NEUROMUSCULAR PERFORMANCE DURING LIFESPAN: ASSESSMENT METHODS AND EXERCISE INTERVENTIONS

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NEUROMUSCULAR PERFORMANCE DURING LIFESPAN: ASSESSMENT METHODS AND EXERCISE INTERVENTIONS

Topic Editors:
Oliver Faude, University of Basel, Switzerland
Lars Donath, German Sport University Cologne, Germany

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EDITORIAL
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Editorial: Neuromuscular Performance During Lifespan: Assessment Methods and Exercise Interventions

Oliver Faude[1]* *and Lars Donath*[2]

[1] Department of Sport, Exercise and Health, University of Basel, Basel, Switzerland, [2] Department of Intervention Research in Exercise Training, German Sport University Cologne, Cologne, Germany

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Editorial on the Research Topic

Neuromuscular Performance during Lifespan: Assessment Methods and Exercise Interventions

Neuromuscular performance can be regarded as the ability of the neuromuscular system to functionally control and drive movements by an appropriate integration, coordination and use of sensory feedback, reflex activity, central motor drive, muscle recruitment pattern, muscular excitation-contraction coupling, and energy availability (Faude et al., 2017). Proper neuromuscular performance enables the human organism to maintain stability and posture within the gravitational field in static and dynamic situations, to generate an appropriate amount of force for a given motor task or to co-ordinate limb movements to protect body structures and to avoid tissue damage, respectively. This definition can be used as a widely applicable umbrella term that subsumes a variety of different dimensions of physical function that determines the way to assess and train neuromuscular performance in different populations and settings.

It is accepted within the scientific community that a well-developed capacity of the neuromuscular system is highly relevant for fitness, sports, and health during the whole lifespan. In early years, the appropriate development of the neuromuscular capacity supports the acquisition of basic movement and motor skills and, thus, contributes to sports competency (Logan et al., 2012). Proper neuromuscular performance development may lead children and adolescents into an active and healthy lifestyle. Furthermore, the capacity of the neuromuscular system is fundamental to achieve peak sports performance in late adolescence and young adulthood (Granacher et al., 2016). In this regard, there is also convincing evidence that injury risk can be reduced by exercise interventions targeting particularly the neuromuscular system (Rössler et al., 2014). During later stages of life, a well-trained neuromuscular capacity enables people to stay active and healthy as well as maintaining the ability to fulfill the job requirements (Jakobsen et al., 2015). In elderly people, neuromuscular fitness can minimize the risk and rates of falling up to 50% (Sherrington et al., 2017). During the later stages of life, the capacity of the neuromuscular system remains relevant to deal with the demands of activities of daily life (ADL) and, thus, to stay mobile and independent as long as possible (McPhee et al., 2016).

Although the relevance of neuromuscular performance is widely recognized, there is a large diversity in assessment methods, cross-sectional associations between relevant outcome measures and potentially efficacious exercise interventions. Whereas, the aerobic capacity or muscular strength are very similarly assessed from childhood to older age, for instance, by conducting a VO_{2max} or one-repetition maximum test, respectively, there is no such uniform assessment method for neuromuscular performance measures. The diversity of assessment methods is mirrored by

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David Jiménez-Pavón, University of Cádiz, Spain

***Correspondence:**
Oliver Faude
oliver.faude@unibas.ch

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a large heterogeneity of intervention approaches. This phenomenon might be attributed to an anticipated task-specificity of neuromuscular co-ordination and adaptations with a lack of far transfer effects (Kümmel et al., 2016). A valid comparison of neuromuscular performance during different stages of the lifespan is, therefore, challenging and remains difficult.

Against the aforementioned background, the main aim of our Research Topic entitled "Neuromuscular performance during lifespan: assessment methods and exercise interventions" was to compile original research articles, systematic reviews and meta-analyses on topics related to neuromuscular performance during lifespan from a cross-sectional and longitudinal perspective. The scientific articles cover "neuromuscular performance" as a broad concept referring to different assessment methods and exercise interventions targeting neuromuscular performance in mainly healthy populations of different ages during lifespan.

In total, 19 scientific articles from 109 authors were finally accepted for publication (for an overview see **Table 1**). Of those, one article is an opinion paper, one a meta-analysis, one an open data set, and the remaining papers report original data. Thereof, 12 articles referred to data from cross-sectional studies and four studies applied longitudinal exercise training interventions. Seven articles referred to the neuromuscular domain of strength, strength-endurance, or muscular power, six dealt with gait or running analysis, four with balance and posture, and one with a general motor control topic and a methodological issue, each.

In the strength and power domain, Kemmler et al. report cross-sectional data of maximum isokinetic knee extension and flexion strength in 362 non-athletic male adults within an age range from 19 to 91 years. The authors found a small decrease in maximum isokinetic leg strength of 0.15% (knee extension) to 0.5% (knee flexion) per year up to the age of 50–60 years and, thereafter an accelerated annual loss of about 1.3%. These results can be used as normative values for healthy male populations of different age groups and point toward the need for sarcopenia prophylaxis in men beginning in the 5th decade in order to address the accelerated muscle decline of advanced age. Nikolaidis et al. and Franchi et al. present normative data for specific athletic populations. Franchi et al. show maximal eccentric hamstring strength data of 170 competitive alpine skiers from the under-15 age group up to elite level skiers. The authors arrive at the conclusion that these data emphasize the relevance of considering maturation when interpreting eccentric hamstring strength data and that such data may support injury prevention approaches. Nikolaidis et al. analyzed force-velocity characteristics, muscle strength, and flexibility in female recreational marathon runners and their relationship with age, race time, and anthropometric characteristics. Although anaerobic power and neuromuscular fitness of female marathon runners are not directly linked to race performance, these findings may be useful for strength and conditioning coaches to monitor the training of athletes, particularly from a more general health-related physical fitness perspective. With regard to potential mechanisms for age-related changes in muscular power, König et al. examined triceps surae muscle strength and tendon stiffness in middle-aged (40–67 years old) compared to younger (18–30 years of age) healthy male adults. In a strength- and stiffness-matched sub-group, the authors analyzed drop jump performance. The authors conclude that "the reduced muscular power output during lower limb multi-joint tasks seen with aging may be due to age-related changes in motor task execution strategy rather than due to muscle weakness."

Acute intervention effects on strength-related outcomes were studied by Barley et al. as well as Siebert et al.. Barley et al. reported that acute dehydration by 3.2% of body mass resulted in impaired muscle strength-endurance and increased fatigue perception in combat sports athletes. Markers of central and peripheral functioning were, however, not altered. Siebert et al. analyzed effects of multidimensional transversal loads around the calf on isometric force during plantar flexions. The results indicate that proprioception may be enhanced and muscle oscillations reduced, which may lead to improved performance. Whether compression garments may impact more complex sports performance remains elusive to date and could be interesting for further research.

Micke et al. conducted a randomized controlled trial on the effects of superimposed whole-body electromyostimulation training vs. traditional strength training on strength and power parameters in male sport students. The results indicate that the combination of dynamic exercises with whole-body electromyostimulation can be as effective as dynamic resistance training alone. A specific "add-on" effect of whole-body EMS cannot be generally assumed.

Van Hooren et al. submitted a highly interesting opinion paper (most viewed paper of this Research Topic as per June 2019) in which they propose a theoretical framework applying dynamical systems theory in order to improve human locomotion. They aimed at, for instance, maintaining gait stability or reducing injury risk. Human locomotion, thereby, can be regarded as a self-organized process based on various attractors on a macroscopic (walking, running), mesoscopic (joint coupling), and microscopic (neural activity and central pattern generators) level, respectively. This framework can serve as a theoretical model, which can be used for studying risk factor models for falls in seniors, for injury prevention in sports or the effectiveness of exercise interventions. Hamacher et al. studied the effects of a number of potentially influencing factors [health and pain status, fear of falling, depression, cognition performance, physical activity, proprioception (joint position sense), peripheral sensation, balance performance, and muscular fitness] on gait stability in 102 adults older than 65 years of age. The authors concluded that the ability to recover from small perturbations may be related to physical activity, peripheral sensation, and pain status. Kluft et al. observed that older adults do not necessarily select a certain motor strategy, which is associated with their physical abilities. They rather have an imprecise perception of their physical abilities during unexpected stepping down movements. It is of crucial interest to further evaluate whether this inappropriate selection of a motor strategy can explain accidental falls in older adults. In this line of reasoning, we previously proposed an agility-based approach to study fall risk and prevention in older adults (Donath et al., 2016). In this regard, Lichtenstein et al.

TABLE 1 | Overview of published research papers categorized by different domains of neuromuscular performance, population, sample size, and study design.

Domain	Population	*N* (Age)	Design	Authors
STRENGTH, FORCE, AND POWER				
Maximal eccentric hamstring strength	Youth/elite skier	170 (14–22 years)	X-sectional	Franchi et al.
Mechanical power output and jumping	Adults	55 (18–67 years)	X-sectional	König et al.
WB-EMS (leg power, jumps, and sprints)	Young adults	19 (25 years)	Intervention (8 weeks)	Micke et al.
Transversal calf muscle loading	Young adults	15 (26 years)	X-sectional	Siebert et al.
Neuromuscular function and dehydration	Youth combat athletes	14 (25 years)	X-Sectional	Barley et al.
Force-velocity and flexibility	Female runner	33 (40 years)	X-sectional	Nikolaidis et al.
Changes of maximal leg strength	Adults to seniors	362 (19–91 years)	X-sectional	Kemmler et al.
GAIT, RUNNING, STEPPING, AND AGILITY				
Modular control of running	Adults	135 (30 years)	Open Data Set	Santuz et al.
Walking based agility testing	Seniors	66 (69 years)	X-sectional	Lichtenstein et al.
Stepping down motor strategies	Seniors	21 (71 years)	X-sectional	Kluft et al.
Sensor based attractor analyses	Clinical seniors	43 (72 years)	X-sectional	Byrnes et al.
Factors affecting gait stability	Seniors	102 (72 years)	X-sectional	Hamacher et al.
Systems theory and locomotion	–	–	Opinion Article	Van Hooren et al.
BALANCE AND POSTURE				
Correlations between balance tasks	Children to seniors	9,353 (6–93 years)	Meta-analysis	Kiss et al.
Reactive balance training	Adults	39 (24 years)	Intervention (4 weeks)	Krause et al.
Interference and balance task learning	Adults	69 (25 years)	Intervention (2 weeks)	Giboin et al.
Athletic training and body perception	Adolescents	67 (14–20 years)	Intervention (6 years)	Ludwig et al.
OTHERS				
Force output and network dynamics	Older adults	47 (55 years)	X-sectional	Gölz et al.
Intermuscular coherence/EMG	Adults	18 (26 years)	X-sectional	Mohr et al.

WB-EMS, whole-body electromyostimulation.

evaluated the reliability and validity of a newly developed test parcour (ACE, Agility Challenge for the Elderly). The walking-based ACE test and its sub-domains (stop and go, cutting, spatial orientation) differentially reflect cardiocirculatory and neuromuscular domains with cardiocirculatory fitness and gait speed contributing most to overall performance. The test, thus, can be useful for documenting changes due to exercise interventions in seniors.

Santuz et al. provided an open-access data set (available at Zenodo; doi: 10.5281/zenodo.1254380) with EMG activities recorded during treadmill running in 135 healthy young adults of both genders. According to the authors, this data set (i) can be used as a prime source for expanding the representation of human motor control due to the large data set, (ii) it can support scientists from multiple disciplines (e.g., musculoskeletal modeling, robotics, neuroscience, sport science), and (iii) it can be used to train students and scientists with muscle synergy extraction methods.

Byrnes et al. analyzed the attractor variability and pattern for acceleration gait data in healthy adults compared to patients with symptomatic lumbar spinal stenosis. There was a difference in attractor patterns between healthy people and patients, but also a large variation in the attractor gait data within groups. Although the attractor used in this study reflects pathology, the variability is too large for a reliable application in this patient population.

The most viewed original article (June 2019) is a randomized controlled trial on the efficacy of reactive balance training as compared to conventional balance training on neuromuscular and kinematic parameters which are relevant with regard to fall risk. Krause et al. found enhanced reflex activity in the leg muscles and improved neuromuscular timing and accuracy in both groups, potentially resulting in more efficient segmental stabilization during fall risk situations. In perturbed situations, the effects were more pronounced and effect sizes larger in the group which performed reactive balance training, pointing toward a high specificity of balance training adaptations. Similarly, the results of Giboin et al. suggest that balance training adaptations are specific and transfer effects are unlikely. Within two different experiments, the authors observed that neither additional intra- nor intersession tasks interfere with the learning of a novel balance task. A meta-analysis of Kiss et al. compared the correlations between different types of balance performance (static, dynamic, proactive, and reactive) in healthy populations during lifespan. The authors observed merely small correlations between any type of balance performance, again indicating that balance is rather task-specific than a "general" ability. This result underpin the need of multiple tests or exercises for the assessment and training-induced changes of balance performance.

Ludwig et al. reported longitudinal data on the development of body posture over a 6-year period in 67 adolescents from age 14 to 20 years. A sub-group performed a specific posture training twice per week in addition to all other physical and sports activities. The results of this long-term longitudinal study suggest that "additional athletic training of 2 h per week including elements for improved body perception seems to have the potential to improve body posture in symptom free male adolescents and young adults."

Although it is well-known that continuous deliberate practice throughout lifespan leading to motor expertise can delay age-related deteriorations in motor skills, relatively little is known about underlying mechanisms. Whether acquired motor expertise leads to a higher initial status from which the age-related decline starts or whether motor expertise results in qualitative differences in motor output and neural processing was the main research question of Gölz et al. A relevant topic from a healthy aging perspective. The authors studied the precision of fine motor control in experts compared to novices with regard to the execution of a dynamic force control task. The obtained results suggest that motor performance of experts is more precise, less variable, and more complex. This finding points toward a better performance and adapted organization of sensorimotor control.

Electromyographical (EMG) assessment of muscle activity can give important insight into the co-ordination and synchronization of muscles during human movements and, consequently, may enable an understanding of muscle performance. Mohr et al. evaluated whether the choice of electrode configuration (monopolar vs. bipolar) or amplifier technology (potential vs. current) have a relevant effect on intermuscular EMG coherence between two muscles (mm. vastus lateralis and medialis) during stable and unstable squatting. The authors indeed found large methodological differences that can explain inconsistencies, which were observed in the scientific literature and should be considered in future studies on muscle activity.

Beside aerobic capacity, a well-developed neuromuscular capacity can be considered a crucial motor performance domain across all age-groups at different performance levels and settings. Different entities refer to "neuromuscular capacity" ranging from strength, power, balance and gait to stepping, running, and agility performance. Numerous behavioral (e.g., postural sway, strength, power, jumping height) and mechanistical (e.g., EEG, EMG) methodological approaches have been developed to "parameterize" neuromuscular performance and capacity, respectively. Standard tools to measure those entities in a valid and reliable manner are highly required and need to be developed and justified based on the intended goals and intentions. The majority of the published papers of this Research Topic are of cross-sectional nature underpinning that walking, stepping and gait are specific motor tasks that involve different strategies on spinal and supra-spinal level. To explain the large variability and specificity of these complex motor tasks theoretical frameworks and one data set have been published. Established behavioral surrogates have been linked with physical and sports performance as well as fall risk and were investigated in both cross-sectional and longitudinal study designs: thereby, for example, the association between performance and dehydration, eccentric strength and maturation, athletic training and body perception, mechanical power and jumping performance, strength and transversal calf muscle load, force velocity and flexibility as well as agility, and neuromuscular performance have been found to be interesting to investigate more detailed. Few studies investigated neuromuscular interventions. Unfortunately, most of these studies lasted <8 weeks. Although interventional studies are methodologically challenging, more studies over longer period of time (e.g., 1 year) are mandatorily needed. This is particularly important as we justify task-specific neuromuscular adaptations mostly based on short time frames of around 12 weeks or less. However, based on more than 9,000 included subjects within the meta-analyses and more than 1,000 investigated subjects within the original papers, the relevance of neuromuscular performance investigated from different perspectives using different approaches across all age groups can be considered very high. The findings of this Research Topic emphasize that neuromuscular motor performance tasks from upright standing to downhill skiing are complex and specific tasks that reflect different neuromuscular requirements. Keeping the different measured or targeted entities of "neuromuscular performance" in mind, it appears challenging to derive gold standard tests to assess neuromuscular performance in general, based on the available papers of this Research Topic. It remains still difficult to compare several tools used in different papers to assess neuromuscular performance as generalizable physical capacity. Future research still needs to rationalize and conceptualize, which outcome needs to be tested and trained in the light of the population, setting, and intended goals. This information should than have impact on interventional study designs that could also focus on primary (hard) endpoints such as fall, disease, mortality, performance, maturation, and aging. These findings should then be incorporated into modeling frameworks that could provide a more holistic and integrative understanding of complex neuromuscular motor tasks.

AUTHOR CONTRIBUTIONS

All authors listed have made a substantial, direct and intellectual contribution to the work, and approved it for publication.

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Intermuscular Coherence Between Surface EMG Signals Is Higher for Monopolar Compared to Bipolar Electrode Configurations

Maurice Mohr[1], Tanja Schön[1,2], Vinzenz von Tscharner[1] and Benno M. Nigg[1]*

[1] Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, AB, Canada, [2] University of Applied Sciences Technikum Wien, Vienna, Austria

Introduction: The vasti muscles have to work in concert to control knee joint motion during movements like walking, running, or squatting. Coherence analysis between surface electromyography (EMG) signals is a common technique to study muscle synchronization during such movements and gain insight into strategies of the central nervous system to optimize neuromuscular performance. However, different assessment methods related to EMG data acquisition, e.g., different electrode configurations or amplifier technologies, have produced inconsistent observations. Therefore, the aim of this study was to elucidate the effect of different EMG acquisition techniques (monopolar vs. bipolar electrode configuration, potential vs. current amplifier) on the magnitude, reliability, and sensitivity of intermuscular coherence between two vasti muscles during stable and unstable squatting exercises.

Methods: Surface EMG signals from vastus lateralis (VL) and medialis (VM) were obtained from eighteen adults while performing series of stable und unstable bipedal squats. The EMG signals were acquired using three different recording techniques: (1) Bipolar with a potential amplifier, (2) monopolar with a potential amplifier, and (3) monopolar electrodes with a current amplifier. VL-VM coherence between the respective raw EMG signals was determined during two trials of stable squatting and one trial of unstable squatting to compare the coherence magnitude, reliability, and sensitivity between EMG recording techniques.

Results: VL-VM coherence was about twice as high for monopolar recordings compared to bipolar recordings for all squatting exercises while coherence was similar between monopolar potential and current recordings. Reliability measures were comparable between recording systems while the sensitivity to an increase in intermuscular coherence during unstable vs. stable squatting was lowest for the monopolar potential system.

Discussion and Conclusion: The choice of electrode configuration can have a significant effect on the magnitude of EMG-EMG coherence, which may explain previous inconsistencies in the literature. A simple simulation of cross-talk could not explain the large differences in intermuscular coherence. It is speculated that inevitable errors in the

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Edited by:
Oliver Faude,
Universität Basel, Switzerland

Reviewed by:
Eduard Kurz,
University Hospital in Halle, Germany
Gennaro Boccia,
Università degli Studi di Torino, Italy

****Correspondence:***
Maurice Mohr
mmaurice@ucalgary.ca

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alignment of the bipolar electrodes with the muscle fiber direction leads to a reduction of information content in the differential EMG signals and subsequently to a lower resolution for the detection of intermuscular coherence.

Keywords: muscle synchronization, surface electromyography, motor unit synchronization, motor unit control, motor control, quadriceps muscle, squatting exercise

INTRODUCTION

The vasti muscles have to work in concert to control knee joint motion and maintain balance of the body during movements such as walking, running, and squatting. Coherence analysis between surface EMG signals from synergistic muscles is a common technique to study intermuscular synchronization and gain insight into strategies of the central nervous system to control the execution of such motor tasks (Farmer et al., 1993; Semmler, 2002). Specifically, previous researchers have used EMG-EMG coherence analyses to elucidate the functional role of intermuscular synchronization, e.g., by investigating its task-dependent property for different motor tasks (Gibbs et al., 1995; Huesler et al., 1998; Kilner et al., 1999; Clark et al., 2013; van Asseldonk et al., 2014; von Tscharner et al., 2014; Mohr et al., 2015; Reyes et al., 2017) or changes in coherence during fatiguing exercises (Boonstra et al., 2008; Kattla and Lowery, 2010; Chang et al., 2012; McManus et al., 2016). These studies suggest that the neuromuscular system adjusts the degree of intermuscular synchronization based on the physical and possibly psychological demands of the movement task. However, some disagreement exists regarding the direction of change in intermuscular synchronization between different movement tasks. For example, higher and lower coherence has been reported to be necessary for balancing movements, which require individual muscle control compared to movements that are stable and require synergistic muscle control (Gibbs et al., 1995; Mohr et al., 2015; Reyes et al., 2017).

When comparing the observed EMG-EMG coherence between multiple studies, it is obvious that the magnitude of coherence as well as the frequency bands where coherence is present can be vastly different. Conceptually, there are three reasons for why previous studies show a large variation in coherence outcomes: First, different EMG recording systems were used (e.g., monopolar vs. bipolar EMG), second, different EMG signal processing techniques were applied, and/or third, the investigated motor tasks and involved muscles were governed by different neuromuscular control strategies leading to different levels of intermuscular synchronization. While many discrepancies between studies can likely be explained by the second and/or third aspect, some studies show considerable differences in coherence despite using the same analysis approaches and despite investigating the same muscles during similar tasks. Therefore, this study will address the first aspect – the influence of the EMG recording system on intermuscular coherence.

For example, Chang et al. (2012) showed that the intermuscular coherence between the vastus medialis and lateralis during a single-leg step-up task is generally lower than 0.5 across frequencies and muscle pairs. In contrast, Mohr et al. (2015) reported EMG-EMG coherence between the vasti muscles during a single-leg squat of generally higher than 0.5 and for a wider range of frequencies up to 80–100 Hz. The major difference between these studies is the use of bipolar and monopolar EMG recording systems, respectively. The rationale for the use of a monopolar over a bipolar EMG amplifier is twofold: First, monopolar EMG avoids the inherent limitation of bipolar EMG systems that the bipolar electrodes must be aligned with the muscle fiber direction. Second, due to differential amplification, bipolar EMG leads to a higher spatial selectivity while monopolar surface EMG provides a more 'global' view on the activity of a muscle (De Luca and Merletti, 1988). Although high spatial selectivity of bipolar EMG may be beneficial when trying to investigate the behavior of individual motor units (Reucher et al., 1987), global information on the activity of two muscles may be desired when investigating intermuscular synchronization at a whole muscle level.

The underlying concept of the bipolar technique is to detect the same motor unit action potentials twice but spatially shifted along the muscle. Then, differential amplification of these two signals leads to a reduction of noise that is common to both electrodes while the signal of interest is retained, i.e., the differential of the summed motor unit action potentials (Basmajian, 1985). This concept relies on the assumption that bipolar electrodes can in fact be aligned with the muscle fiber direction. However, most muscles fibers are oriented at a three-dimensional pennation angle with respect to the aponeurosis and the skin surface, which may change as a function of joint position and muscle force (Wickiewicz et al., 1983; Friederich and Brand, 1990; Rutherford and Jones, 1992; Merletti et al., 2001; Rainoldi et al., 2001). Even if the investigator can achieve a good alignment of the electrodes before the measurement, the assumption that the bipolar electrodes remain aligned with the muscle fiber direction during movements that involve muscle length and force changes does not hold. Bipolar electrode alignment error can alter the amplitude and frequency content of the differential EMG signal in an unknown and unpredictable way (von Tscharner, 2014), which may reduce the ability of this technology to detect intra- and intermuscular coherence.

In contrast, monopolar EMG measurements do not require electrode alignment, represent the entire information about motor unit activity near the measurement point and may thus be more suitable to study intermuscular synchronization. Accordingly, monopolar EMG has been successfully applied to resolve the task-dependent property of intermuscular synchronization between isometric and dynamic squats with a high sensitivity (Cohen's d = 2.3, re-computed from Mohr et al., 2015). To the best knowledge of the authors, the effect of monopolar vs. bipolar EMG measurements on the analysis of EMG-EMG coherence is currently unknown. Based on the above argument, however, it is speculated that the disruption of information in bipolar EMG recordings may lead to a lower resolution to detect high EMG-EMG coherence between muscles and explain the discrepancies between previous studies. Despite these possible advantages of monopolar surface EMG, the technique is not commonly used in biomechanical and neuromuscular investigations. This is due to the susceptibility of monopolar surface EMG to noise from stray-potentials, movement artifacts, and possibly cross-talk due to low spatial selectivity (De Luca and Merletti, 1988; von Tscharner et al., 2013), which may compromise the reliability of a monopolar system.

In addition to using a monopolar electrode configuration, Mohr and colleagues obtained EMG signals via a recently developed current amplifier in contrast to the classic EMG potential amplifier (von Tscharner et al., 2013). The main difference is that the current amplifier injects or withdraws charges at the skin surface above the active muscle to keep all measurement points at ground potential while the potential amplifier relies on a potential at the skin surface with respect to the ground electrodes. The concept of the current amplifier has the advantage that inter-electrode currents are avoided, which enables EMG measurements during conditions when the inter-electrode impedance is largely reduced, e.g., when sweat builds up on the skin or even during extreme conditions such as during underwater measurements (Whitting and von Tscharner, 2014). Furthermore, current measurements may be more sensitive to the EMG signals at higher frequencies compared to measurements from potential amplifiers and may therefore be more suitable to explore coordinated motor unit activity at frequencies beyond the typically investigated beta and gamma bands (>60 Hz) (von Tscharner et al., 2013).

In summary, there are well-founded arguments for the use of monopolar amplifiers and current measurements instead of the traditional bipolar potential measurements when investigating EMG-EMG coherence. However, the effect of monopolar vs. bipolar electrode configurations and potential vs. current EMG recording techniques on the magnitude and frequency of intermuscular coherence has not been systematically investigated. Similarly, it is unknown whether one of these EMG recording techniques can more reliably detect intermuscular coherence or if one is more sensitive to detecting a change in intermuscular coherence between different movement tasks.

Therefore, the first objective of this study was to compare intermuscular coherence of vastus lateralis and medialis surface EMG signals during a dynamic, bipedal squatting task between three different EMG recording techniques: Bipolar potentials, monopolar potentials, and monopolar currents.

The second objective was to compare these three techniques regarding their reliability when repeatedly assessing a stable squatting task and their sensitivity to detecting a change in intermuscular coherence between squatting on a stable vs. unstable surface.

It was hypothesized that:

(1) VL-VM intermuscular coherence would be higher for monopolar EMG signals compared to bipolar signals, and
(2) All three recording techniques would be sensing a lower VL-VM intermuscular coherence during unstable compared to stable squatting although monopolar systems would show a reduced reliability between similar squatting trials.

MATERIALS AND METHODS

Participants

Eighteen healthy, male (n = 14) and female (n = 4) participants (mean ± SD; age 26 ± 5 y; height 175 ± 6 cm; mass 69 ± 7 kg) volunteered to participate in this study. This study was carried out in accordance with the guidelines of the University of Calgary's Conjoint Health Research Ethics Board. The protocol was approved by the University of Calgary's Conjoint Health Research Ethics Board (#REB17-0210). All subjects gave written informed consent in accordance with the Declaration of Helsinki.

Study Design

Each participant completed a total of six squatting trials, three trials were recorded with the monopolar current amplifier system and three trials were recorded with the monopolar potential amplifier system (**Table 1**). The bipolar potential signals were computed from the monopolar signals following data acquisition (see section "EMG potential measurements"). The order of recording systems was balanced randomized. The order of trials was kept constant and consisted of two trials of squatting on a stable surface and one trial of squatting on an unstable surface. The protocol included two trials of stable squatting to determine the between-trial reliability of intermuscular coherence and one trial of unstable squatting to determine the sensitivity of the three systems.

Squatting Tasks

During each trial, participants performed a series of squats down to a knee flexion angle of 70 degrees (0 degrees represents full extension) for a duration of 90 s. The distance between the participants' feet was self-selected and kept constant throughout all trials but had to be at least shoulder wide apart (**Figure 1**). Stable squatting trials were performed on the laboratory floor while unstable squatting trials were completed on the flat side of a BOSU ball (**Figure 1**). For all trials, the squatting speed was set to 20 squats per minute and controlled for by using a metronome at 40 bpm yielding a total of 30 squats per trial that were used for data analysis. In order to ensure consistent knee flexion angles at the lowest squat position, participants were given visual real-time feedback from a one-dimensional electrogoniometer (Biometrics Ltd., United Kingdom) taped across the anterior side of their

TABLE 1 | Design of experimental procedures.

Amplifier	Configuration	Trial 1	Trial 2	Trial 3
Potential	Monopolar	Stable	Stable	Unstable
	Bipolar			
Current	Monopolar	Stable	Stable	Unstable

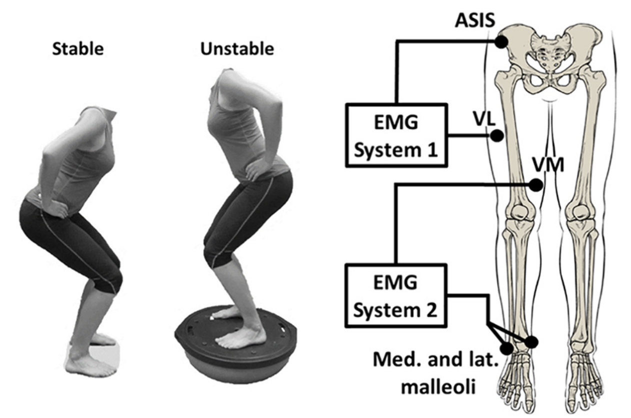

FIGURE 1 | Stable vs. unstable squat (left); conceptual EMG recording set-up (right).

knee joint. Each participant was given one initial practice trial to familiarize with the equipment and squatting speed. For each following trial, the EMG recording system was started once the participant had found the correct squatting rhythm.

EMG Electrode Placement

In order to obtain surface EMG signals from VM and VL, the skin surface above the muscles was shaved, slightly abraded with sand paper and cleaned with alcohol wipes to ensure high signal conductivity. Two Ag-AgCl electrodes (10 mm diameter, 20 mm inter-electrode distance, Norotrode Myotronics-Noromed Inc., United States) in a bipolar configuration were placed over the muscle bellies of VM and VL using the following procedure. First, the electrode positions and orientations on VM and VL were located and marked according to EMG sensor locations described in SENIAM guidelines (Hermens et al., 1999). Next, an ultrasound machine was used to verify that the marked electrode locations were within the proximal-distal and medio-lateral boundaries of the muscles while the participants where performing a static squat at 45 degrees of knee flexion.

EMG Recording Systems

Electromyography recordings of each muscle were obtained using two separate recording systems with separate ground electrodes, data acquisition cards (12-bit A/D converter, National Instruments, Austin, TX, United States), and battery powered laptops. Thus, the systems consisted of two electronically separated circuits to avoid hardware-based crosstalk (Mohr et al., 2015). In system 1, EMG signals of VL were recorded with reference to two ground electrodes placed side by side on the right anterior superior iliac spine. In system 2, EMG signals of VM were recorded with reference to two ground electrodes placed on the medial and lateral malleoli (**Figure 1**). Two ground electrodes were used in each system to improve the stability of the ground potential and to further reduce the resistivity to the returning currents. Each electrode was connected to an extension lead and then fixed in place using adhesive stretch tape. This step was necessary to ensure that the electrode-skin connection was kept constant throughout the protocol when switching between the current and potential measurements. The two recording systems of VL and VM were synchronized using a custom-built device that simultaneously transmitted a pulse to both systems upon pressing a button at the beginning and end of each measurement.

EMG Potential Measurements

Two monopolar EMG potentials were recorded from each muscle using a total of four differential amplifiers at a sampling frequency of 2400 Hz with a hardware-based bandpass filter between 10–500 Hz (Biovision, Wehrheim, Germany). The positive input of the amplifiers was connected to one of the two electrodes placed on each muscle and the negative input was connected to the respective ground. In this configuration, one can use a differential amplifier to record monopolar EMG potentials. Bipolar EMG potentials for VM and VL were computed following data acquisition by calculating the difference between the two monopolar EMG potentials obtained from each muscle (proximal – distal electrode). This approach was selected to compare intermuscular coherence between monopolar and bipolar EMG potential measurements that were obtained from the same squatting trial. A pilot experiment was conducted where bipolar EMG potentials directly recorded from the VL with a single differential amplifier were compared with the computed bipolar EMG potentials as explained above. The power spectra of a 60 s isometric squat were virtually identical between the two methods, thus verifying the validity of the approach.

EMG Current Measurements

Monopolar EMG currents were recorded from the proximal electrode on each muscle using a previously described and validated current amplifier at a sampling frequency of 2400 Hz and a hardware-based bandpass filter between 10–500 Hz (von Tscharner et al., 2013; Mohr et al., 2015).

EMG Signal Analysis

Filtering

Goniometer data were low-pass filtered (cut-off frequency of 1 Hz) using a wavelet-based filter method. The 60 Hz line-frequency contamination was removed from all monopolar EMG signals by applying a line-frequency averaging method and a line filter. In short, this procedure allows to subtract the average line-frequency contamination from the EMG signal without inducing a notch in the EMG power spectrum at 60 Hz (see von Tscharner et al., 2013 for further details). Removing the line-frequency from the signals avoided an artificial intermuscular coherence at 60 Hz. The lowest frequency that was considered for this analysis was 10 Hz, which is given by the 10–500 Hz bandpass filter of the EMG amplifiers and by the notion that the power density function of the surface EMG signal has negligible contributions below 10 Hz (Merletti, 1999).

Sequencing

For each squatting trial, the signals were separated into 30 sequences of 4096 samples (1.7 s) according to peaks in the goniometer signal that represented the time points of highest knee flexion, i.e., the deepest positions during the squats. While these sequences contained the majority of the EMG power during the squats (**Figure 2**), the exact sequence size facilitated using a fast Fourier transform (FFT) during the analysis.

Power and Coherence

The FFT of the unrectified EMG signals was computed for each data sequence, leading to a frequency resolution of 0.6 Hz. The power spectra for each muscle and trial were determined by multiplying the FFT of each sequence with its complex conjugate and averaging across all data sequences. Intermuscular coherence as a function of frequency λ (coherence spectrum) between VL and VM EMG signals for one given squatting trial was computed from the average cross-spectra normalized by the corresponding power spectra across s = 30 data sequences (Rosenberg et al., 1989):

$$\text{coherence}(\lambda) = \frac{|\overline{F_{VL_s}(\lambda) \cdot F_{VM_s}(\lambda)^*}|^2}{\overline{(F_{VL_s}(\lambda) \cdot F_{VL_s}(\lambda)^*)} \cdot \overline{(F_{VM_s}(\lambda) \cdot F_{VM_s}(\lambda)^*)}} \quad (1)$$

For each trial and participant, the average coherence was computed as the mean of the coherence spectrum between 10–60 Hz. The frequency range of 10–60 Hz was chosen since the coherence in this range was highest across all trials and participants and since it spans frequencies in the beta (15–30 Hz) and gamma (30–60 Hz) bands, at which intermuscular coherence is typically reported in the literature (Clark et al., 2013; Marchis et al., 2015; Pizzamiglio et al., 2017).

To assess the possible influence of cross-talk between the vasti muscles on the level of intermuscular coherence measured with different recording systems, a simple simulation was performed. From previous studies it was estimated that in the monopolar electrode configuration, there may be an additional 10% of cross-talk compared to the bipolar configuration due to the absence of spatial filtering (Farina et al., 2002). Therefore, a pair of simulated monopolar VL and VM signals was computed from the respective bipolar EMG signals by adding the VL signal multiplied by a factor of 0.1 to the VM signal and vice versa. The coherence analysis as described above was then repeated for these computed signals with simulated cross-talk.

EMG Intensity

In order to determine whether the level of VL and VM muscle excitation changed between the stable and unstable squatting condition, the overall EMG intensity was determined using a wavelet transform. In short, a filter bank of 30 non-linearly scaled wavelets specifically designed for EMG analysis was used to decompose the raw EMG signals into the time-dependent power in each of the frequency bands (wavelets) (von Tscharner, 2000). For this analysis, powers from twenty wavelets with center frequencies between 10–300 Hz were summed to derive the total power. The square root of the total power yields the total EMG intensity, a close approximation of the frequently used EMG root mean square (von Tscharner, 2000). The overall EMG intensity of VL and VM representing the average level of muscle excitation was calculated as the sum of the total EMG intensity for each individual squat. For each recording system separately, the overall EMG intensities were normalized to the maximum overall EMG intensity obtained across all 90 squats (3 trials of 30 squats). Finally, the normalized EMG intensities were averaged across the 30 squats for each trial to derive one normalized, mean overall EMG intensity for each trial, system, muscle and participant.

Statistical Analysis

For each trial (stable 1, stable 2, unstable) and recording technique (bipolar potential, monopolar potential, monopolar current), the mean and standard deviation of the power and coherence spectra, average coherence values, and normalized overall EMG intensities were computed across 16 participants. In addition, the mean and standard deviation of the average coherence values for the simulated monopolar signals were determined to investigate a possible influence of cross-talk. Two male participants had to be excluded from the analysis as they were not able to perform the unstable squatting trials without help from the investigator. A two-way repeated measures ANOVA with the within-subject factors ‘trial’ and ‘recording technique’ was performed to detect significant main and interaction effects on the average coherence. Mauchly’s test of sphericity was used to test the assumption of sphericity. If the assumption of sphericity was violated, the Huynh–Feldt correction was used and reported. Bonferroni-corrected *post hoc* tests were carried out to determine pairwise comparisons of coherence between individual trials and recording techniques. To investigate a possible effect of squatting technique on the level of muscle excitation, separate two-way repeated measures ANOVAs with the within-subject factors ‘trial’ and ‘recording technique’ were performed for the overall EMG intensities of the two muscles VL and VM. All statistical tests were carried out at a significance level of 0.05 using IBM SPSS statistics (v. 24; SPSS Inc., Chicago, IL, United States).

The reliability of the average VL-VM coherence was determined for each recording technique using the first and second stable squatting trial. Relative reliability was computed using the intra-class correlation coefficient [ICC, model 3 (two-way mixed effects, absolute agreement), type 1] and the corresponding 95% confidence intervals (Shrout and Fleiss, 1979; McGraw and Wong, 1996; Koo and Li, 2016). Absolute reliability was determined using the standard error of measurement (SEM) according to equation (2):

$$\text{SEM} = \sqrt{MS_{\text{E}}} \quad (2)$$

where MS_{E} is the error term obtained from the ANOVA table of the ICC calculations (Eliasziw et al., 1994). The SEM represents the random error of the obtained scores in comparison to the ‘true’ scores in the original units of measurement with the assumption that there is no systematic bias between the measurements. To decide whether an observed change in the obtained scores can be considered ‘true’ change, the SEM can be used to derive the minimal detectable change (MDC), according to equation (3):

$$\text{MDC} = 1.96\sqrt{2}\,\text{SEM} \quad (3)$$

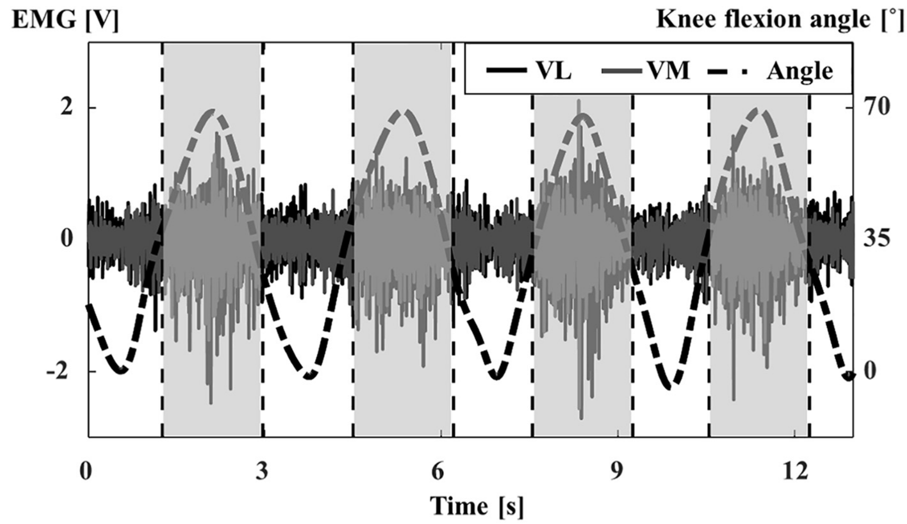

FIGURE 2 | Procedure to separate individual data sequences according to peaks in the knee flexion angle. VL (black, solid line) and VM (gray, solid line) filtered EMG signals recorded with the monopolar potential system and corresponding knee flexion angle (black, dash-dot line). Dashed vertical lines to the left and right of peaks in the knee flexion angle trace indicate the boundaries of individual data sequences (shaded) that were used for further analysis.

The sensitivity of the three recording systems to a change in the average VL-VM coherence when changing from the stable to the unstable squatting condition was assessed according to Cohen's *d* as a measure of effect size. Specifically, the effect size for each system was determined as the mean of the differences between the two conditions (unstable – stable) divided by the standard deviation of the differences. Values for Cohen's *d* of greater than 0.8 represent large effects (Cohen, 1992).

RESULTS

Mean Power and Coherence Spectra

Figure 3 displays the average power spectra and coherence spectra for all recording techniques and the stable and unstable squatting condition. The average VL-VM coherence spectra are clearly reduced when obtained from bipolar compared to monopolar recordings (**Figures 3G–I**). Although the coherence spectra of monopolar potentials and currents generally show a similar shape, the spectra obtained from monopolar currents demonstrate a higher coherence for frequencies above 80 Hz. In addition, it can be observed that the coherence during unstable squatting is larger compared to stable squatting, particularly for frequencies below 40 Hz.

While the power spectra for VL and VM show a similar pattern, the spectra show a different shape when comparing the bipolar and monopolar recording techniques. Specifically, for monopolar recordings the power spectra demonstrate a pronounced peak in the frequency range of 30–50 Hz, which is much less visible in spectra obtained from bipolar recordings. It is also within this frequency band, that a high coherence was observed in the coherence spectra. Similarly, the magnitude of this 30–50 Hz peak is reduced during the unstable compared to the stable squatting condition (**Figures 3A–F**).

Average Coherence

There were significant main effects of 'trial' [$F(1.44,21.59) = 18.24$, $p < 0.001$] and 'recording technique' [$F(3,45) = 61.3$, $p < 0.001$] on the average coherence. There was no significant interaction term between 'trial' and 'recording technique' [$F(3.86,57.9) = 1.79$, $p = 0.144$].

Regarding the first study objective, the *post hoc* comparisons indicated that VL-VM intermuscular coherence was significantly reduced by more than 50% for bipolar potential measurements compared to monopolar potential and monopolar current measurements for each individual squatting trial. There were no significant differences in average coherence between the monopolar potential and current measurements. Further, there were no significant differences in average coherence between the bipolar measurements and the simulated monopolar signals with added cross-talk (**Figure 4A**).

Coherence was significantly higher during the unstable squatting trial compared to both stable squatting trials for the bipolar potential and monopolar current measurements. Despite a similar trend, there were no statistically significant differences in the average coherence between squatting trials for the monopolar potential measurements (**Figure 4B**).

Regarding the second study objective, the reliability and sensitivity of the three recording systems are displayed in **Table 2**. For all recording systems, there were no average differences between the first and second stable squatting trial (**Figure 4B**). For both potential measurements, the lower bounds of the ICC 95% confidence intervals were above 0.75, indicating good relative reliability (Koo and Li, 2016). The monopolar current measurement showed excellent relative reliability [ICC = 0.98

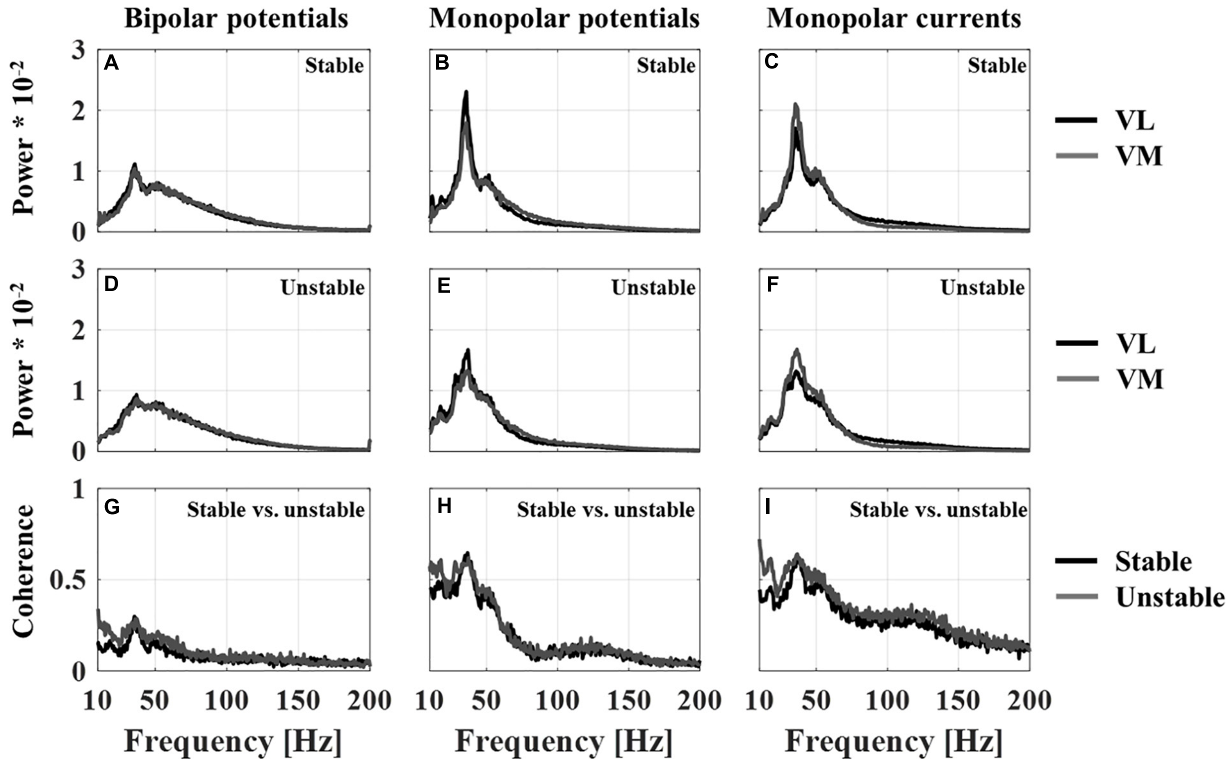

FIGURE 3 | Average of the normalized power spectra across all subjects (*n* = 16) for vastus lateralis and medialis for each recording technique (BP – bipolar potentials, MP – monopolar potentials, MC – monopolar currents) for the stable squat **(A–C)** and unstable squat **(D–F)**; Average of the coherence spectra across all subjects (*n* = 16) between vastus lateralis and medialis for each recording technique and squatting condition **(G–I)**.

(0.94,0.99)]. The MDC in average VL-VM coherence during squatting was between 0.06 for the bipolar potentials and 0.07 for the two monopolar systems. Only for the monopolar current measurements, the mean difference between the unstable and stable squatting condition (trial 3 – trial 1) exceeded the MDC. Similarly, the monopolar current measurements showed the highest effect size (Cohen's d = 1.34), followed by the bipolar (0.91) and monopolar potentials (0.63).

EMG Overall Intensity

Figure 5 shows the average, normalized overall EMG intensity of VL and VM during all three squatting trials. For both muscles, there was a significant interaction effect between 'trial' and 'recording technique' on the overall EMG intensity [VL: $F(2.78, 41.75) = 3.07$, $p = 0.041$; VM: $F(3.27,49.02) = 4.39$, $p = 0.007$]. *Post hoc* comparisons showed that on average, there were no significant differences in EMG intensity between the stable and unstable squatting condition for the bipolar system. For all monopolar recordings, there was a small trend for an average percentage increase of about 10% during the unstable compared to stable squatting condition. The average increase in overall EMG intensity in the unstable vs. stable condition only reached statistical significance for the current measurements of the vastus medialis (trial 1 vs. 3, $p = 0.028$). The large standard deviations in **Figure 5** indicate that there was a high degree of variability between the individuals regarding which of the three squatting trials showed the highest overall EMG intensity.

DISCUSSION

This study aimed to investigate the effect of different EMG recording techniques on the magnitude, reliability, and sensitivity of intermuscular coherence during dynamic squatting tasks. Our first hypothesis that intermuscular coherence would be higher when computed from monopolar compared to bipolar EMG signals was confirmed by the result that the average coherence for bipolar potential measurements was significantly reduced compared to the monopolar potential and current measurements. The second hypothesis that all three systems would be sensitive to a lower VL-VM coherence during unstable vs. stable squatting was not supported by the findings that (1) the monopolar potential recordings showed low sensitivity to the change in coherence between the two squatting conditions and (2) the average coherence was in fact higher during unstable compared to stable squatting.

Recording Techniques – Bipolar vs. Monopolar

The reduction in coherence for bipolar compared to monopolar EMG recordings may have two possible reasons: (1) the reduction

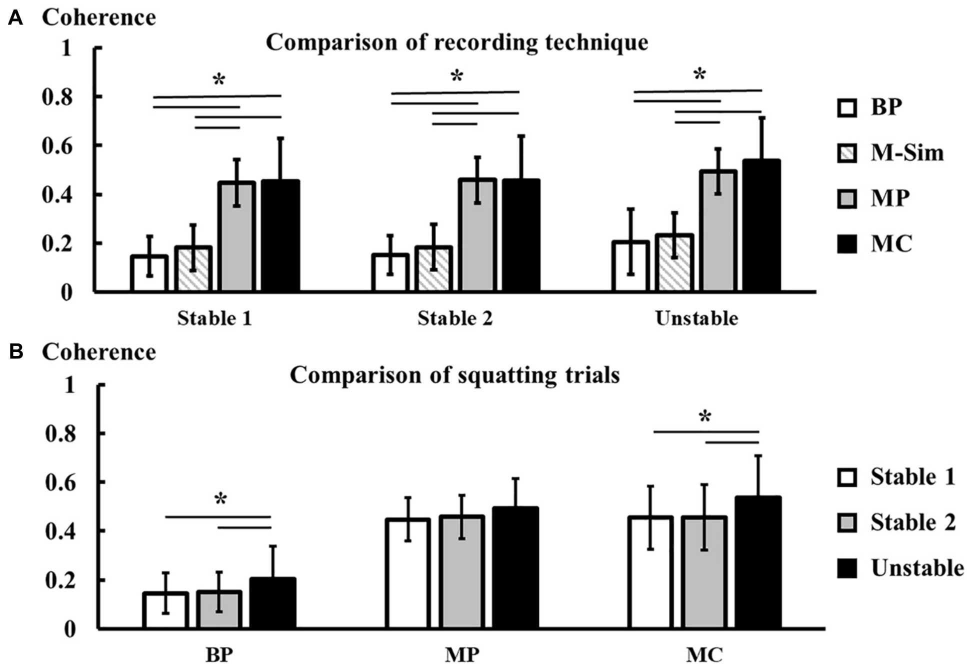

FIGURE 4 | Comparison of average coherence (mean ± SD, n = 16) between recording techniques (BP – bipolar potentials, M-Sim – simulated monopolar potentials, MP – monopolar potentials, MC – monopolar currents) **(A)**, and comparison of average coherence between squatting trials **(B)**. Asterisks mark statistically significant differences between conditions at α = 0.05.

or disruption of amplitude and frequency information within the bipolar signals that arose from electrode alignment and position errors and subsequent differential amplification, and (2) less cross-talk in bipolar recordings due to spatial filtering and higher spatial selectivity.

When a muscle is activated, motor unit action potentials travel along the muscle fibers, starting from the innervation zone and ending at the muscle insertion or origin (Basmajian, 1985). For bipolar EMG measurements, two adjacent electrodes are applied between innervation zone and tendon and in alignment with the muscle fiber direction to detect the same motor unit action potentials twice but spatially shifted along the muscle. Subtracting the signals from these two monopolar EMG signals yields a single-differential bipolar EMG signal, which has the advantage that common noise under both electrodes is reduced (Gallina et al., 2013). Consequently, bipolar EMG measurements have been state-of-the-art in investigating muscle activation patterns during movement and have frequently been used to study intermuscular synchronization (Gibbs et al., 1995, 1997; Kilner et al., 1999; Halliday et al., 2003; Boonstra et al., 2008; Kattla and Lowery, 2010; Reyes et al., 2017). During movements such as squatting, however, the fiber direction and location of the innervation zone of the quadriceps muscles with respect to the electrode position on the skin are functions of the knee angle and quadriceps muscle force (Rutherford and Jones, 1992; Gallina et al., 2013). Similarly, the innervation zone has been shown to move with respect to the skin as a function of knee angle (Rainoldi et al., 2000; Gallina et al., 2013). In consequence, bipolar electrodes cannot be properly aligned with the muscle fiber direction and the bipolar EMG signal will likely represent a combination of (1) the differential between propagating motor unit action potentials from the same motor units recorded twice at different locations along the muscle fiber direction and (2) the differential between motor unit action potentials that originate from different motor units (von Tscharner et al., 2013). The ratio of these differentials depends on the geometry

TABLE 2 | Reliability and sensitivity of average coherence outcomes.

System	Reliability				Sensitivity		
	Relative		Absolute		Unstable – Stable		Cohen's d
	ICC	95% CI	SEM	MDC	Mean	SD	
Bipolar potential	0.93	(0.81,0.97)	0.02	0.06	0.06	0.07	0.91
Monopolar potential	0.92	(0.80,0.97)	0.03	0.07	0.05	0.07	0.63
Monopolar current	0.98	(0.94,0.99)	0.03	0.07	0.08	0.06	1.34

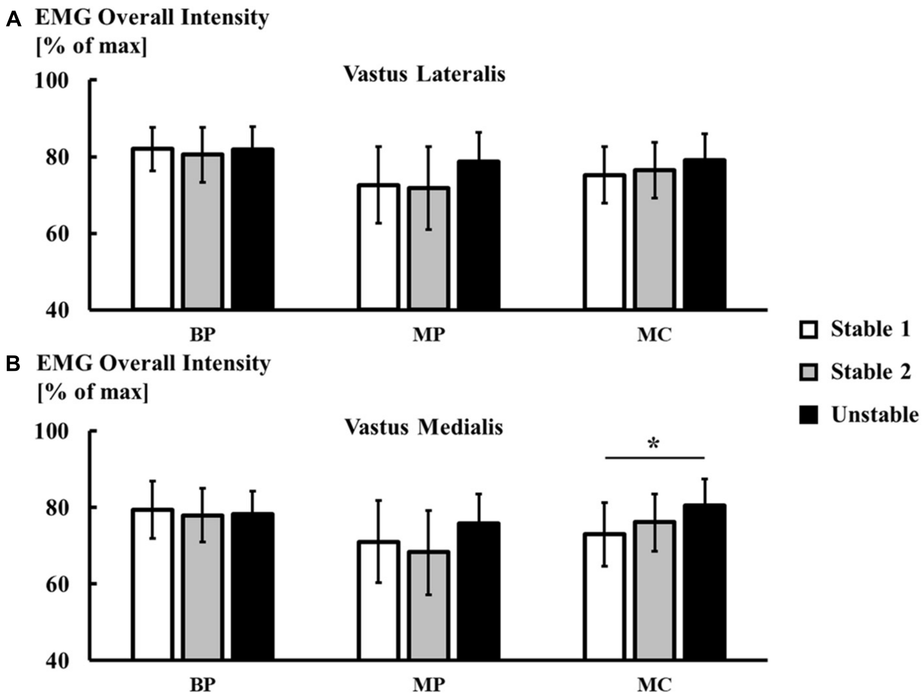

FIGURE 5 | Comparison of normalized overall EMG intensity (mean ± SD, n = 16) between recording techniques (BP – bipolar potentials, MP – monopolar potentials, MC – monopolar currents) for the vastus lateralis **(A)**, and vastus medialis **(B)**. Asterisks mark statistically significant differences between conditions at α = 0.05.

of the muscle as well as electrode placement and will change with the pennation angle throughout a movement. It is well known that a portion of motor units within one muscle are synchronized in time, i.e., intramuscular synchronization, with an accuracy of more than 5 ms (Sears and Stagg, 1976; Bremner et al., 1991; De Luca et al., 1993). Therefore, it is likely that signals recorded by the electrodes in a bipolar measurement setup are highly correlated and are either eliminated or at least disrupted in an unspecific and unpredictable way by the common mode rejection of the amplifier (von Tscharner, 2014). In consequence, only the signals that are uncorrelated between adjacent electrodes, i.e., not synchronized, are retained in bipolar measurements.

Evidence for this assumption can be seen in the average power spectra that were obtained using the monopolar and bipolar recording systems in this study. The monopolar recordings demonstrate a pronounced peak in the power spectrum between frequencies of 30–50 Hz. This 40 Hz peak in the EMG power spectrum of dynamic tasks has been observed previously and has been connected to rhythmic bursts of clustered motor unit activity, where multiple motor units are firing within a short time window of 10 ms (Yao et al., 2000; Maurer et al., 2013; Asmussen et al., 2018). If the two electrodes of a bipolar amplifier are recording motor unit action potentials from different motor units that are virtually firing at the same time, the common mode rejection would likely remove a significant amount of this information and explain why the 40 Hz peak is absent or much reduced in amplitude in the power spectra obtained from bipolar recordings.

Both intra- and intermuscular synchronization of motor units as measured by EMG-EMG coherence have been speculated to originate primarily from common, or shared inputs of the corticospinal tract to the respective motoneuron pools (Bremner et al., 1991; Farmer et al., 1993; Lowery et al., 2007). In consequence, the degree of intra- and intermuscular synchronization is most likely correlated. If the bipolar EMG signal from one muscle only contains information about uncorrelated motor unit activity, as described above, it will be more difficult to detect intermuscular synchronization between different muscles. In contrast, monopolar EMG recordings contain the entire signal information and inherently do not need to be aligned with the muscle fiber direction. Therefore, it is speculated that the reduced or disrupted information within the bipolar EMG signal is the theoretical basis for the reduced VL-VM coherence in comparison to monopolar signals seen in this study.

The second possible origin of the difference in intermuscular coherence observed between the recording systems is a varying influence of cross-talk. Cross-talk ratios of about 10% were reported between the VL and VM during isometric knee extensions (Farina et al., 2002). Specifically, it has been shown that cross-talk was reduced in surface EMG signals from the thigh muscles when they were obtained using a double-differential vs. a single-differential recording technique (De Luca and Merletti, 1988; Farina et al., 2002). The reduction of cross-talk is most likely due to an increase in spatial selectivity of the EMG system when using double-differential amplification (Reucher et al., 1987). Albeit not systematically investigated to date, it can be speculated that monopolar surface EMG signals may thus contain more cross-talk components compared to bipolar single-differential signals. Cross-talk between EMG recordings of adjacent muscles can artificially inflate EMG-EMG coherence and should therefore be carefully addressed as a potential confounding factor in this study (Grosse et al., 2002; Halliday et al., 2003). There are three reasons why the influence of cross-talk on the findings of this study is likely small. First, the EMG measurement set-up of this study was carefully designed to record VL and VM signals using two electronically separated circuits with separate grounds to exclude the possibility of hardware-based cross-talk (Mohr et al., 2015). Second, when measuring intermuscular EMG-EMG coherence, significant cross-talk between the two muscles of interest typically leads to a resulting coherence spectrum that shows high values across a broad range of frequencies, spanning almost the entire EMG bandwidth (Grosse et al., 2002; Halliday et al., 2003). This was not observed for any of the individuals tested in this study. Third and most importantly, we used a simple simulation to investigate the possible influence of an additional 10% of cross-talk components in monopolar compared to bipolar EMG signals. Although on average, the simulated signals show a slightly higher coherence compared to the bipolar signals (see **Figure 4A**, BP vs. M-Sim), this difference was not statistically significant and cannot explain the large increase in coherence from the bipolar to the monopolar recording systems. In summary, while the presence of cross-talk can not be completely excluded in this study, cross-talk was not a major confounding factor in the comparison of monopolar vs. bipolar EMG systems.

Recording Techniques – EMG Intensity

A second difference between the recording systems was observed for the level of muscle excitation according to the overall EMG intensity during the stable and unstable squatting exercise. While the bipolar EMG measurements of the vasti muscles did not show an average change in EMG intensity between the movement conditions, the monopolar recording systems showed a small, average increase in EMG intensity for both VL and VM during the unstable squat. This is in accordance with a previous study showing no or only a small percentage increase ($<10\%$) in thigh muscle activity when switching from squatting on a stable to squatting on an unstable surface (Anderson and Behm, 2005). The discrepancy between the monopolar and bipolar recording systems could originate from additional synchronized inputs that the vasti muscles received from the central nervous system during the unstable squat as suggested by the corresponding increase in VL-VM coherence during this exercise. Such synchronized motor unit activity would increase the overall EMG intensity (Yao et al., 2000; Asmussen et al., 2018) but may not be detected by the bipolar EMG recording system due to the elimination or reduction of common input signals as explained above.

Recording Techniques – Potentials vs. Currents

A third difference between the recording systems was observed between the coherence spectra obtained using the monopolar current compared to the potential amplifiers. Specifically, the current amplifier detected a higher magnitude of coherence for frequencies above 80 Hz compared to the potential amplifiers. The presence of high-frequency intermuscular coherence has been reported between EMG signals for upper and lower limb muscles (Chang et al., 2012; Marchis et al., 2015; Mohr et al., 2015; Pizzamiglio et al., 2017). Intermuscular coherence within the gamma band (30–60 Hz) and higher frequencies has been speculated to represent a coupled, descending motor command to muscles involved in movement tasks that require dynamic modulation of muscle force for error correction – such as squatting on an unstable surface in the current study (Pizzamiglio et al., 2017). For these force modulations, it may be preferable for the central nervous system to primarily activate fast motor units due to their ability to generate higher forces and faster conduction velocities (Milner-Brown et al., 1973; Wakeling and Syme, 2002; Hodson-Tole and Wakeling, 2009). In parallel, it has been suggested that faster motor units generate motor unit action potentials that contribute high-frequency components to the EMG signal (Wakeling and Rozitis, 2004), which could explain the second, smaller peak in the coherence spectra at frequencies above 100 Hz seen in this study (see **Figures 3H,I**). However, direct evidence for a preferential recruitment of fast, large motor units for a mixed fiber type muscle is currently not available. Nevertheless, it is unlikely that this second high-frequency coherence peak seen for the current recordings is due to a measurement artifact but it is unclear why this peak is much reduced or absent in the potential recordings.

Previously, von Tscharner et al. (2013) had observed that the monopolar current amplifier is more sensitive in detecting EMG signal power at high frequencies, which could be a reason for the higher EMG-EMG coherence at these frequencies in the current recordings. However, **Figure 3** does not show an obvious difference between the average power spectra of monopolar current and potential amplifiers at frequencies above 80 Hz. It may be that synchronized, fast motor units only have a negligible contribution to the average EMG power, which is dominated by frequencies below 80 Hz, but that they still contribute to the coherence spectrum, which is independent of signal amplitude. Further research is required to understand why the current amplifier may be more sensitive in resolving motor unit action potentials at higher frequencies.

Stable vs. Unstable Squat

Both the bipolar potential and monopolar current system showed an average increase in VL-VM coherence during the squat on the unstable BOSU balance trainer compared to the stable squat. Albeit not statistically significant, the monopolar potential system also showed an increase toward a higher VL-VM coherence during unstable squatting. In parallel, there was no difference in intermuscular coherence between the first and second trial of stable squatting, demonstrating the absence of a possible learning effect. For both the bipolar potential and monopolar current system, the increased VL-VM coherence during unstable squatting was equal to or exceeded the respective MDC. In combination, these findings suggest that the neuromuscular strategy to control the vasti muscles changed when adding an unstable surface to the squatting exercise.

While all three recording systems indicated an average increase in VL-VM coherence between the two movement conditions, the bipolar potential and monopolar current systems were more sensitive compared to the monopolar potential system. Therefore, if researchers are interested in studying a change in intermuscular coherence between two different tasks, the bipolar potential system or monopolar current system seem to be more suitable than the monopolar potential technique.

The squatting movement on the BOSU balance trainer was selected as a task that is comparable to squatting on a stable surface in terms of joint kinematics and net force while demanding a greater involvement of the individual quadriceps muscles in maintaining postural stability. The result of a higher coherence during unstable squatting was not expected since previous investigators have reported a reduction in intermuscular coherence when performing a task that requires more individual muscle control compared to a task that requires more synergistic muscle control (Mohr et al., 2015; Reyes et al., 2017). For example, Reyes et al. (2017) demonstrated a reduction in intermuscular beta-band coherence (15–30 Hz) between a finger and a thumb muscle during a task where participants pinched an unstable spring compared to a task where a stable cylinder was compressed with a matched force. Furthermore, musicians who require more individual control of finger muscles showed a lower degree of motor unit synchronization within a finger muscle compared to weight lifters who have trained to use their finger muscles in synergy (Semmler and Nordstrom, 1998). It is questionable, however, whether vastus medialis and lateralis in this study were in fact controlled more individually by the central nervous system during the squat on the BOSU balance trainer compared to the stable squat. Anderson and Behm compared the general level of EMG intensity of vastus lateralis as well as of lower leg and core muscles between squatting on a stable vs. unstable surface (Anderson and Behm, 2005). Vasti EMG intensity was not significantly different between the two squatting conditions, which corroborates the result of this study, whereas the EMG intensity of the core and lower leg muscles was increased by up to 50% during the unstable condition. This indicates that the role of the quadriceps in maintaining postural stability during the unstable squat is small in relation to core and lower leg muscles. As a consequence, it may not be appropriate to compare the current findings with previous studies that investigate individual muscle control paradigms. Instead, the authors speculate that during both squatting exercises, the vasti muscles act as prime movers and were thus controlled as a functional unit by the central nervous system (De Luca and Erim, 2002; Anderson and Behm, 2005). During the unstable squat, the motor units of the vasti muscles may have received additional, intermittent and synchronized inputs to achieve small adjustments in the knee flexion angle trace while squatting on the BOSU ball. Furthermore, these intermittent bursts of activity may have disturbed the rhythmic, clustered motor unit activity related to the 40 Hz peak in the VL and VM power spectra and, thus explain the reduced magnitude of this peak during the unstable squat in **Figure 3**. In support of this argument, Gibbs et al. (1995) showed that the motor unit synchronization between two synergistic lower leg muscles as measured by a cross-correlation analysis was higher during a balancing standing task compared to a regular standing task and compared to voluntary contractions while lying down. It was suggested that the increase in motor unit synchronization may originate from a greater involvement of the vestibular system, specifically that the muscles received synchronized inputs from increased activity in vestibulospinal neurones. The authors speculate that a similar neuromuscular mechanism could explain the finding of higher VL-VM intermuscular coherence during the balancing task in this study.

Reliability and Sensitivity

The question remains if one of the EMG recording techniques, bipolar vs. monopolar, is more suited to investigate EMG-EMG coherence as a measure of intermuscular synchronization. A higher coherence score alone does not necessarily indicate that the monopolar system is more suitable. Therefore, reliability and sensitivity analyses were performed to give further insight into this question. Comparing all three systems, it was observed that the coherence obtained from the monopolar currents showed the highest relative reliability between two stable squatting trials as well as the highest sensitivity when changing to unstable squatting with a large effect size of greater than one. The monopolar potential measurements, however, showed a low sensitivity and could not resolve the increase in coherence when changing between squatting conditions. This could be because monopolar potential recordings are more susceptible to stray potentials in the measurement environment and electrical noise that could contaminate the signals and reduce the system sensitivity (von Tscharner et al., 2013). The bipolar system showed good relative reliability and resolved a large effect between the stable and unstable squat, although with a lower sensitivity compared to the monopolar current system.

Therefore, when studying EMG-EMG intermuscular coherence to investigate the relative change in intermuscular synchronization between two or more movement conditions, both the bipolar potential and monopolar current systems seem to be suitable while the monopolar potential system should not be used. A monopolar current technique may be preferable over the traditional bipolar technique if (1) the muscles of interest are far enough apart that cross-talk between monopolar electrodes has a minor influence, and (2) the movement of interest does not involve impacts, e.g., walking or running. The latter would induce large motion artifacts in monopolar EMG measurements, which would produce a misleading EMG-EMG coherence.

When studying the magnitude of intermuscular coherence as a measure of the absolute degree of intermuscular synchronization between two muscles for a certain individual or a group of individuals, monopolar EMG recordings on the one hand may provide a more 'global' view on correlated motor unit activity at the whole muscle level. On the other hand, bipolar EMG recordings, particularly in combination with additional spatial filtering techniques or when applied as multi-electrode arrays, may provide better information on the behavior and synchronization of individual motor units. Whether one or the other technique better represents the physiological origin of correlated motor unit activity, i.e., the strength of common inputs to the motor neuron pools of two muscles, should be the focus of future studies.

CONCLUSION

This study investigated the effect of three different surface EMG recording systems on the coherence between the raw EMG signals of vastus medialis and lateralis during bipedal squatting on stable and unstable surfaces. When EMG signals were obtained with the traditional bipolar potential amplifier, the magnitude of intermuscular coherence between 10–60 Hz was less than half compared to the coherence based on monopolar signals. This may be a consequence of disrupted information about motor unit activity contained in the bipolar EMG signals as a result of the elimination of common signals by the differential bipolar amplifier. A simple simulation of additional cross-talk in monopolar signals could not explain this substantial difference in coherence between the recording systems. When comparing squatting exercises on a stable and unstable surface, only the bipolar potential and monopolar current system resolved an increase in intermuscular coherence for the unstable surface, with a larger effect size for current measurements. The monopolar potential system showed low sensitivity to the change in the movement condition and should therefore not be used to determine intermuscular coherence. If cross-talk plays a minor role and in the absence of movement artifacts, both bipolar potential and monopolar current measurements are suited to study changes in intermuscular coherence as an indicator of varying levels of intermuscular synchronization between different conditions.

DATA AVAILABILITY STATEMENT

The data (raw data set/processed data sheet) for this study can be found in the Mendeley data repository (Mohr et al., 2018).

AUTHOR CONTRIBUTIONS

MM and VvT conceived and designed the experiments. TS performed the experiments. TS, MM, and VvT analyzed the data. VvT and BN contributed reagents, materials, and analysis tools. BN supervision. MM wrote the first manuscript draft. MM, TS, VvT, and BN reviewed, edited, and accepted final manuscript version.

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ORIGINAL RESEARCH
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Neuromuscular and Kinematic Adaptation in Response to Reactive Balance Training – a Randomized Controlled Study Regarding Fall Prevention

*Anne Krause[1,2], Kathrin Freyler[1], Albert Gollhofer[1], Thomas Stocker[3], Uli Brüderlin[3], Ralf Colin[3], Harald Töpfer[3] and Ramona Ritzmann[1]**

[1] Department of Sport Science, University of Freiburg, Freiburg, Germany, [2] Institute of Training and Computer Science in Sport, German Sport University Cologne, Cologne, Germany, [3] Department of Mechatronics, University of Applied Sciences, Esslingen, Germany

Edited by:
Lars Donath,
German Sport University Cologne,
Germany

Reviewed by:
Martin Behrens,
University of Rostock, Germany
Eling D. de Bruin,
Karolinska Institutet (KI), Sweden

***Correspondence:**
Ramona Ritzmann
ramona.ritzmann@sport.uni-freiburg.de

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Slips and stumbles are main causes of falls and result in serious injuries. Balance training is widely applied for preventing falls across the lifespan. Subdivided into two main intervention types, biomechanical characteristics differ amongst balance interventions tailored to counteract falls: conventional balance training (CBT) referring to a balance task with a static ledger pivoting around the ankle joint versus reactive balance training (RBT) using externally applied perturbations to deteriorate body equilibrium. This study aimed to evaluate the efficacy of reactive, slip-simulating RBT compared to CBT in regard to fall prevention and to detect neuromuscular and kinematic dependencies. In a randomized controlled trial, 38 participants were randomly allocated either to CBT or RBT. To simulate stumbling scenarios, postural responses were assessed to posterior translations in gait and stance perturbation before and after 4 weeks of training. Surface electromyography during short- (SLR), medium- (MLR), and long-latency response of shank and thigh muscles as well as ankle, knee, and hip joint kinematics (amplitudes and velocities) were recorded. Both training modalities revealed reduced angular velocity in the ankle joint ($P < 0.05$) accompanied by increased shank muscle activity in SLR ($P < 0.05$) during marching in place perturbation. During stance perturbation and marching in place perturbation, hip angular velocity was decreased after RBT (P from TTEST, $P_t < 0.05$) accompanied by enhanced thigh muscle activity (SLR, MLR) after both trainings ($P < 0.05$). Effect sizes were larger for the RBT-group during stance perturbation. Thus, both interventions revealed modified stabilization strategies for reactive balance recovery after surface translations. Characterized by enhanced reflex activity in the leg muscles antagonizing the surface translations, balance training is associated with improved neuromuscular timing and accuracy being relevant for postural control. This may result in more efficient segmental stabilization during fall risk situations, independent of the intervention modality. More pronounced modulations and higher effect sizes after RBT in stance perturbation point toward specificity of training adaptations, with an emphasis on the proximal body segment for RBT. Outcomes underline the benefits of balance training with a clear distinction between RBT and CBT being relevant for training application over the lifespan.

Keywords: reflex, electromyography, posture control, balance, conventional balance training, kinematics, lifespan, reaction

INTRODUCTION

Perturbation-related falls in response to slips or trips are major causes (>60%) of injuries over the lifespan (Gallagher et al., 1984; Winter, 1995; Rubenstein, 2006). As a consequence, affected individuals suffer from physical impairments, reduced autonomy, and a constrained quality of life (Tinetti, 1994; Rubenstein, 2006; Heinrich et al., 2010). Fall scenarios and related injuries among children (Gallagher et al., 1984), adults (Timsina et al., 2017), and seniors (Alexander et al., 1992) constitute a major public health problem and have gained socioeconomic importance due to high clinical and consequential costs (Miller et al., 2000; Stevens et al., 2006). Hence, scientific debates about efficient countermeasures move into focus (Granacher et al., 2011a).

Comparing populations of high-risk fallers to non-fallers, beyond cognitive and strength deficits, factors as the following have been empirically identified as predisposing a person to a greater fall incidence: a decreased ability to stabilize postural equilibrium (Arampatzis et al., 2008), deteriorated balance recovery (van Dieën et al., 2005; Karamanidis and Arampatzis, 2006), undersized timing, and extent of the postural response (Tang and Woollacott, 1998). In particular, distinctly declined neuromuscular activity (Tang and Woollacott, 1998; Granacher and Gollhofer, 2005; Gehring et al., 2014), smaller peak knee displacement after translation (Horak et al., 2005) or rotation of the support surface (Bakker et al., 2006), increased joint torques and angular velocities in gait (Lee and Kerrigan, 1999; Lord et al., 2001; Dobkin and Dobkin, 2003) have been determined in fallers. In other words, not the age itself, but rather the overall level of movement control seems to be the limiting factor to break one's fall. This can be verified in both children who lack adult-like maturity of their joint control (Ganley and Powers, 2005) and in elderly who lose acquired skills as a result of progressive aging-induced degradation (Sawers et al., 2017). Independent of the age category, fall prevention programs have been established to counteract the falls and diminish consequential costs (Granacher et al., 2011a).

Scientific debates dealing with ***fall prevention*** outlined paradigms, including balance training, to counteract postural instability through more effective compensatory muscle activation in young and old sub-samples (Granacher et al., 2006; Karlsson et al., 2013; Ungar et al., 2013). Recently, a novel type of balance training was introduced: reactive balance training (RBT) is an intervention *simulating* the fall situation itself by the application of unpredictable, random, multi-directional displacements of the support surface (Shumway-Cook et al., 2003; Obuchi et al., 2004; Bieryla and Madigan, 2011; Granacher et al., 2011b, 2012; Mansfield et al., 2015). It was shown that crucial adaptive skills for resisting falls can be acquired rapidly among young and older adults through a single session of RBT with exposure to slips on a movable platform (Bhatt and Pai, 2008; Pai et al., 2010; Bhatt et al., 2011) and transfer effects persist beyond the laboratory for fall situations encountered in daily living (Bhatt et al., 2006). However, evidence for the effectiveness of RBT applied over several weeks is still limited and varies greatly regarding perturbation stimuli in simulated fall risk paradigms (Fitzgerald et al., 2000, 2002; Shumway-Cook et al., 2003; Obuchi et al., 2004; Shimada et al., 2004; Hurd et al., 2006; Mansfield et al., 2010; Bieryla and Madigan, 2011; Bierbaum et al., 2013). Besides its effectiveness, insights into the specific neuromuscular modulations after such forms of balance training are still lacking. Nonetheless, the knowledge about those modulations is the basis to develop further recommendations for RBT as a possible fall avoidance training. Furthermore, fall preventive adaptations are further merely assessed in indirect measures, such as reduced time to stabilize equilibrium (Shumway-Cook et al., 2003; Bieryla and Madigan, 2011) and modified reactions to a stimulus, which are assessed by means of frequency and contact time (Mansfield et al., 2010) concomitant with changed neuromuscular activation for regaining equilibrium (Obuchi et al., 2004). Although conjunctions with fall (Shimada et al., 2004) and injury prevention (Hurd et al., 2006), or even with returning to physical activities within the rehabilitation process are assumed (Fitzgerald et al., 2000, 2002), fundamental evidence about the associated functional benefits and underlying neuromuscular mechanisms for avoidance of falls is missing.

In contrast, conventional balance training (CBT) performed on unstable surfaces and convex devices has been validated to improve postural stability (Granacher et al., 2006; Gruber et al., 2007a,b; Taube et al., 2008; Freyler et al., 2014) and to elicit functional and neuromuscular adaptations beneficial for fall avoidance, such as augmented muscle strength (Bruhn et al., 2006). This includes explosive force and rate of force development (Gruber and Gollhofer, 2004; Taube et al., 2007) as well as modified muscle activity, such as increased activation or reduced co-contraction by means of improved muscle coordination, induced by neural adaptations within the central nervous system (Granacher et al., 2009; Nagai et al., 2012; Oliveira et al., 2013; Behrens et al., 2015). These neuronal adaptations were specified by higher amplitudes and shorter latencies in the early reflex responses in the shank muscles, resulting in augmented ankle joint stiffness (Granacher et al., 2006) and reduced fall frequency (Madureira et al., 2007) in response to postural perturbations.

Comparing RBT to CBT from a biomechanical point of view, stabilizing torques are shifted from distal to proximal. According to the pendulum model (Winter, 1995; Schmitt, 2003), "punctum fixum" and "punctum mobile" are exchanged from the support surface (CBT) to the center of mass itself (RBT), challenging the subject to maintain equilibrium above an unstable moving support surface (RBT) instead of oscillating around a fixed ledger (CBT) (Maki and McIlroy, 2006) (**Figure 1**). As a consequence, RBT requires an accurate repositioning of the center of mass (COM) utilizing rapid and appropriate neuromuscular responses to regain a stable body position after surface translation. Thus, RBT may challenge reactive postural stability more than CBT (Horstmann and Dietz, 1990; Yim-Chiplis and Talbot, 2000) and may be more effective as an intervention to counteract falls (Freyler et al., 2016).

Based on the aspects mentioned above, the rationale of the study was to compare neuromuscular and kinematic adaptations of RBT and CBT in terms of the relevant fall risk factors. Adaptations were measured during functional tasks of stance and marching in place perturbation before and after the training intervention. Focus was set on muscular activation patterns characterized by phase-specific reflex parameters indicated as short- (SLR), medium- (MLR), and long- (LLR) latency responses following the onset of perturbation (Horak and Nashner, 1986; Diener et al., 1988; Gollhofer and Rapp, 1993). While SLR mainly comprises Ia afferent reflex responses, information during MLR is also transmitted via the midbrain and brainstem. Latest responses (LLR) encompass transcortical pathways through the cerebral cortex (Jacobs and Horak, 2007). RBT might simulate a fall risk situation, which is why differentiated neuromuscular and kinematic effects were expected to be more pronounced in RBT than CBT. It was hypothesized that improvements induced by RBT in dynamic stabilization after perturbation would ameliorate neuromuscular control for slip recovery characteristics, comprising (a) an elevated reflex activity in the shank and thigh muscles and (b) enhanced kinematic segmental stabilization to compensate for the disturbing stimulus during stance and marching in place perturbation.

MATERIALS AND METHODS

Participants

Thirty-nine healthy participants of sport students [24 females (*f*), 15 males (*m*), age 24 ± 3 years] participated in the study. The sample size was estimated by means of a power analysis (test attributes: *F*-test, repeated-measures analysis of variance, within-between factors, $f = 0.25$, medium effect; alpha = 0.05; Power = 0.75) (Faul et al., 2007). Volunteers who performed any other kind of balance training or suffering from acute injuries or neurological irregularities were excluded. We requested a written document from all subjects confirming the absence from any kind of additional balance training apart from this study. All participants, furthermore, gave written informed consent to the experimental procedure, which is approved by the ethics committee of the University of Freiburg (EK Freiburg 16/13) in accordance with the latest revision of the Declaration of Helsinki. Using the concealed allocation procedure, participants were randomly divided up by "matched-pairs" either into a RBT group (RBT-group, 11f/8m, age 24 ± 3 years, height 173 ± 7 cm, weight 67 ± 12 kg) or a control group that performed conventional balance training (CBT-group, 13f/7m, age 24 ± 3 years, height 172 ± 9 cm, weight 67 ± 10 kg, one male drop-out, rate 2.6%). Matched-pairs were determined prior to the interventions based on the postural sway measures (described in detail in the outcome measures). Therefore, subjects who demonstrated almost equivalent performance levels prior to training were randomly allocated either to the CBT-group or to the RBT-group by drawing lots.

Experimental Design

In a randomized control trial, a repeated-measures matched-subject design (subjects and therapists were not blinded; assessors were blinded) was used to ascertain the effect of a 4-week trial of RBT versus CBT on neuromuscular and biomechanical aspects of fall avoidance characteristics, postural reflexes and kinematics in response to perturbation (**Figure 2**). Two protocols were used in a randomized order to assess the ability to compensate for stance and gait disturbances in stumbling situations after the interventions. However, for pre to post measurements, we conducted the same order in each individual to exclude any effects due to preceding tasks. In *protocol 1*, training-induced effects on postural reactions in a static balance setting was investigated, while, in *protocol 2*, a dynamic test setting during locomotion was investigated. Electromyograms (EMGs) of four leg muscles as well as ankle, knee, and hip joint kinematics were recorded during both settings before and after the interventions (**Figure 2**). Prior to each measurement, isometric maximal voluntary contractions were performed for all recorded muscles according to Freyler et al. (2016) for EMG normalization. Training and testing sessions were surveyed, supervised, and documented by the authors.

Training Intervention

For both groups the training interventions lasted 4 weeks and comprised three sessions per week lasting 20 min each. Groups trained in parallel. One session consisted of four parts separated by 1-min breaks, each part comprised four repetitions, respectively. Each repetition lasted 1 min and was divided into 40 s training with a 20 s intermittent break (Taube et al., 2008; Lesinski et al., 2015). Training settings were matched regarding training frequency, number and duration of sets and pauses. RBT was performed on an electromagnetically driven swinging platform (Perturmed, Brüderlin, Germany) generating surface translations in the horizontal plane (eight directions: medio-lateral (ml), antero-posterior (ap), and the diagonals (for a detailed description of the device see Freyler et al., 2015). Meanwhile, the CBT-group trained with conventional balance devices, including unstable surfaces ranging from easy (Airex® Balance-Pad) to more challenging postural demands (Togu® Dynair air cushions ø 33 cm, Aero-Step cushions 51 cm × 37 cm × 8 cm/46 cm × 32 cm × 8 cm, Jumper® 52 cm × 24 cm) of different balance performance complexities (Gruber et al., 2007a; Taube et al., 2008; Lesinski et al., 2015). For both groups, participants were instructed to maintain or to recover stability with their head forward-facing, arms akimbo and the non-standing leg being flexed. If this task was accomplished

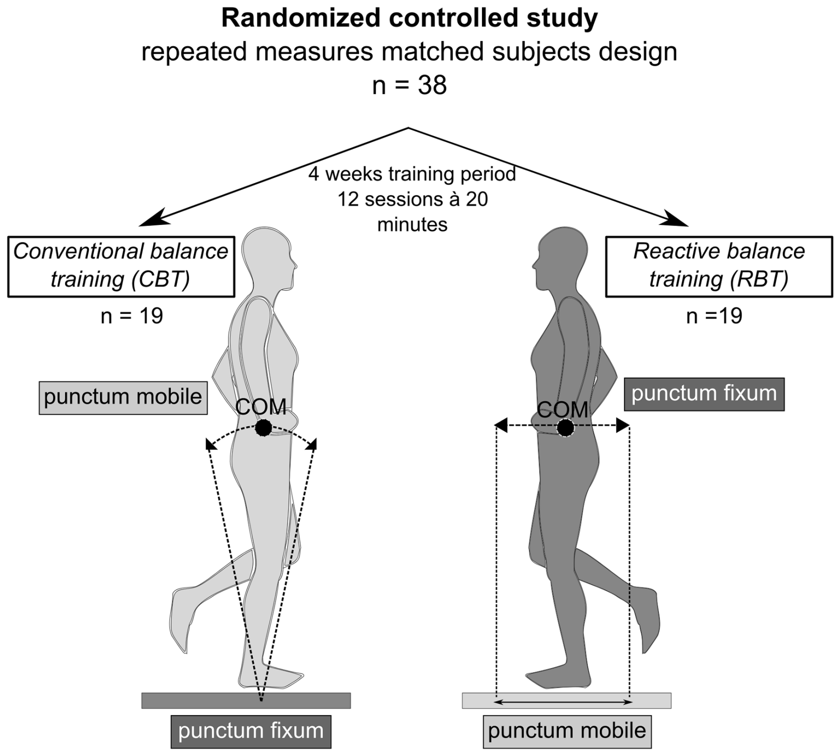

FIGURE 1 | Study overview and models comparing conventional versus reactive balance training. Comparison of the two training paradigms on the basis of pendulum models (Winter, 1995; Schmitt, 2003) with the conventional training (CBT) describing a single inverted pendulum, while participants in the reactive balance training (RBT) group oscillate around the fixation point located within the center of gravity. In this randomized controlled study, baseline data collection was followed by a random "matched-pair" assignment of 39 volunteers to either the CBT-group or RBT-group. Before and after 4 weeks of training, measurements were conducted to assess changes in fall-related risk factors.

easily, the level of difficulty was raised successively and individually within the training period. For RBT, the perturbation program (embracing 16 levels in total) was increased in difficulty in the following order: increase of translation displacement (2–4–6 cm), increase of additional directions (ap–ml– diagonals), and reduction of durations between perturbations (4–2–1 s break). For CBT, support surfaces were varied for an individual increase of difficulty. Subsequently, eyes were closed during training to exclude visual cues for RBT and CBT.

Protocols

To exclude habituation effects prior to measurements, subjects practiced for a period of 10 min in the test conditions. The order of balance protocols and tasks was randomized (by drawing lots) to exclude confounding effects but was controlled in post-assessments referring to pre-measurements.

Protocol 1 – Stance Perturbation

To determine compensatory reactive responses to sudden perturbations, as they occur during stumbling, posterior translations of the support surface were induced randomly to the left leg during static monopedal stance (**Figure 2**). Translations were conducted on a customized platform [Department of Sport Science, University of Freiburg, cf. (Mornieux et al., 2014)] moving horizontally backward with an amplitude of 16.21 $\pm$ 0.04 cm and impulse duration of 140 $\pm$ 3 ms resulting in a mean velocity of 1.22 $\pm$ 0.02 m $\cdot$ s^{-1} and a maximal acceleration of 11.2 $\pm$ 0.5 m $\cdot$ s^{-2}. Fifteen perturbations were induced randomly within a range of 2–4 s (Hall and Jensen, 2002). Participants were asked to sustain balance on one leg (right) and to stabilize equilibrium as quickly as possible. In case of balance loss or using their other leg, trials were repeated.

Protocol 2 – Marching in Place Perturbation

For functional testing, participants marched in place with the left foot stepping on the movable platform. Fifteen displacements were induced randomly after trespassing a light barrier so that the support surface was translated during full weighted foot-contact unexpected in time (Mornieux et al., 2014). Gait pace was controlled and standardized by a metronome (112 beats $\cdot$ min^{-1}). Prior to recording, all participants practiced the task for 10 min so that participants were habituated to the task and marching steps were reliable. Arm were moved in a crisscrossing pattern, and vision was forward-facing. Translations were induced by the same customized platform as described in *protocol 1* with an amplitude of 20.05 $\pm$ 0.05 cm and impulse duration of 150 $\pm$ 2 ms resulting in a mean velocity of 1.34 $\pm$ 0.02 m $\cdot$ s^{-1} and a maximal acceleration of 14.2 $\pm$ 0.8 m $\cdot$ s^{-2} (Granacher et al., 2006; Mornieux et al., 2014). Participants were asked to sustain

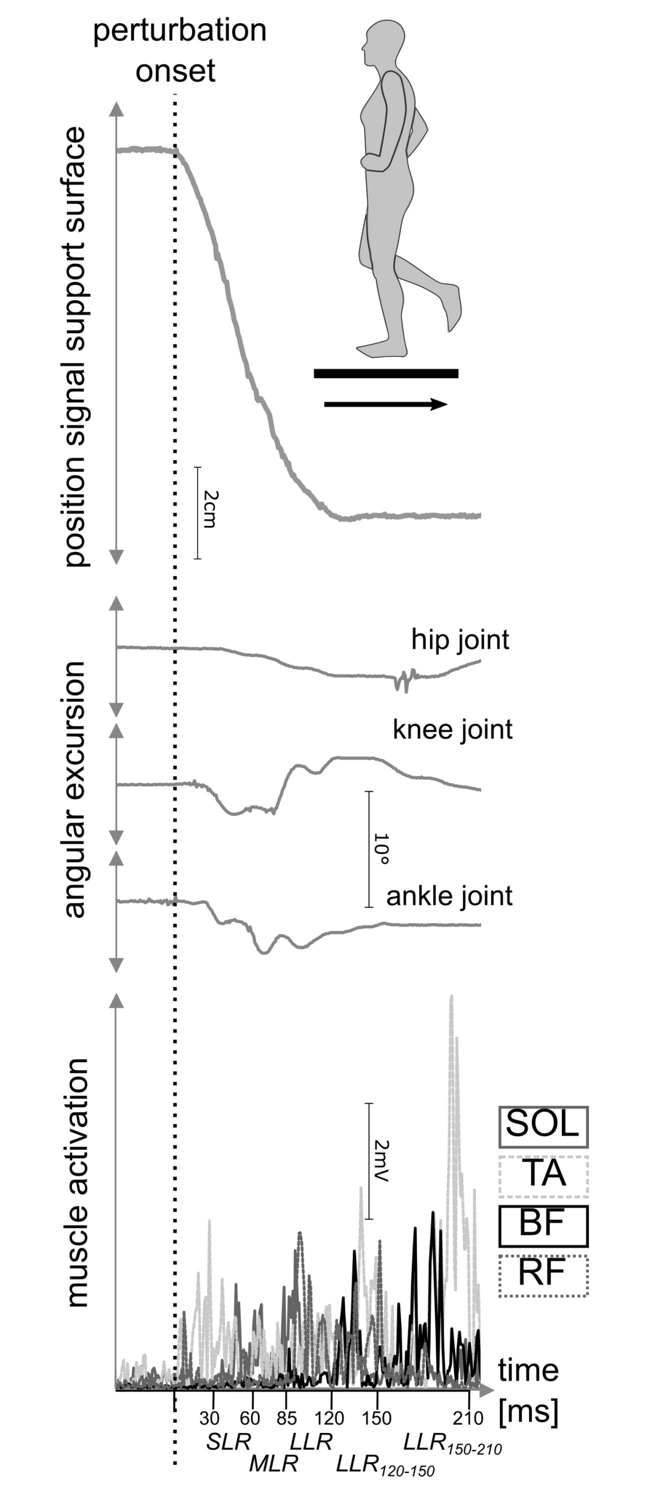

FIGURE 2 | Outcome measures during posterior perturbation of a representative participant. Recording of neuromuscular data (bottom) and kinematic data (middle) during posterior translation of the support surface (top) were carried out. For muscular activation, the soleus (SOL), tibialis anterior (TA), biceps femoris (BF), and rectus femoris (RF) muscles were measured during short- (SLR), medium- (MLR), and long-latency responses (LLR) after perturbation onset. Simultaneously, joint excursions of the ankle, knee, and hip joint were assessed.

balance after perturbation and to stabilize equilibrium as quickly as possible. In case of balance loss, defined as bracing oneself against the wall next to the customized platform, trials were repeated.

Outcome Measures

Neuromuscular Data

According to SENIAM (Hermens et al., 2000), surface EMGs of selected muscles of the shank and the thigh of the left leg [soleus (SOL), tibialis anterior (TA), biceps femoris (BF), and rectus femoris (RF)] were recorded. Bipolar Ag/AgCl surface electrodes (Ambu Blue Sensor P, Ballerup, Denmark; diameter 9 mm, center-to-center distance 25 mm) were placed onto the muscle belly in line with the underlying muscle fibers. A reference electrode was fixed onto the patella. Interelectrode resistance was kept below 5 kΩ by means of shaving, light abrasion, and degreasing of the skin with a disinfectant. EMG signals were transferred via shielded cables to an amplifier (band-pass filter 10 Hz–1 kHz, 1000× amplified) and sampled with 1000 Hz.

Kinematics

Ankle, knee, and hip joint angles were recorded with monoaxial electrogoniometers (Biometrics®, Gwent, United Kingdom). For that purpose, the center of rotation of the goniometer was placed over the rotational axis of the respective joint (ankle: malleolus lateralis, knee: knee joint cavity, and hip: trochanter major) and the two arms (proximal and distal) were aligned in extension of the joint axes (ankle: pointing toward fifth metatarsal and longitudinal axis of the shank, knee: pointing toward malleolus lateralis and trochanter major, and hip: longitudinal axis of the femur and trunk). For details, see Freyler et al. (2016).

Data Processing

For the analysis of neuromuscular data, the EMG of each muscle was rectified, averaged, and integrated [iEMG (mVs)]. iEMG was analyzed regarding the reflex phases after perturbation: SLR (30–60 ms), MLR (60–85 ms), and LLR (85–120 ms) (Grey et al., 2001; Taube, 2006). In addition, the iEMG was calculated for the interval 120 ms until the end of the perturbation (150 ms) and according to Dietz et al. (1989) up to 210 ms after the perturbation onset. Subsequently, all iEMG data were individually normalized to those recorded during maximal voluntary contraction (Halaki and Gi, 2012). An onset latency of each muscle was identified as the first burst >2 standard deviations above the baseline iEMG (Henry et al., 1998). Percentage differences (pre/post) were calculated from normalized values corresponding to baseline data for each participant and subsequently averaged.

Ankle, knee, and hip joint kinematics were expressed as mean joint amplitudes in the time interval during posterior translation [°]. Angular excursion was averaged for each participant and normalized to the neutral position defined at 90° in the ankle joint (longitudinal axis foot/fibula) and 180° in the knee (longitudinal axis fibula/femur) and hip joint (longitudinal axis femur/trunk). The angular velocity of joint excursions (Ω) was assessed as follows: $\Omega = x \cdot t^{-1}$ with x describing the displacement [°] and t the time to maximal excursion [s] in a timeframe of 0–200 ms.

Statistics

To determine statistical differences within the independent variable groups (2, RBT-group versus CBT-group) and time (2, pre versus post) a repeated-measures analysis of variance (rmANOVA) was conducted. Dependent variables were iEMG data (SOL, TA, BF, RF), latencies (SOL, BF, RF) and angular excursion and velocity (ankle, knee, and hip joint). The normality of the data was evaluated with the Kolmogorov–Smirnov test; data followed a normal distribution. If the assumption of sphericity measured by Mauchly's sphericity test was violated, Greenhouse-Geisser correction was used. To detect one-sided effects, additional one-tailed paired student's t-tests (TTESTs) were calculated. The level of significance was defined at $P < 0.05$. "P" indicates the level of significance for rmANOVA, "P_t" describes results of TTESTs. To control for changes in onset latency within the different muscles, a univariate ANOVA was conducted including *post hoc* tests.

Additionally, effect sizes were calculated according to Cohen (d) and by means of Partial Eta squared ($\eta_p{}^2$, see **Tables 1–5**). Reference values are defined as trivial with $d < 0.2$ ($\eta_p^2 < 0.01$), as small with $0.2 < d > 0.5$ ($0.01 < \eta_p^2 > 0.06$), as medium with $0.5 < d > 0.8$ ($0.06 < \eta_p^2 > 0.14$) and as large effects with $d > 0.8$ ($\eta_p^2 > 0.14$) (Cohen, 1988, 1992; Thalheimer and Cook, 2002).

Statistical methods were conducted with the statistics software SPSS 20.0 (SPSS, Inc., Chicago, IL, United States). Group data are presented as mean value ± standard deviation.

RESULTS

Protocol 1 – Stance Perturbation

Neuromuscular Activity

Grand means of the iEMG activity are listed in **Table 1** and illustrated in **Figure 3** (for coefficient of variation cf. **Supplementary Table S1**). Significant time effects for both groups ($P < 0.05$) could be shown for RF in SLR and MLR, indicating higher activation amplitude after the training intervention. For the RBT-group, reflexive BF muscle activity showed a tendency toward augmentation in SLR and LLR ($P_t = 0.06$). In later time frames of the neuromuscular response, iEMG was enhanced for BF ($LLR_{120-150}$) and for SOL for both groups, RBT and CBT ($LLR_{120-150}$, $LLR_{150-210}$, $P < 0.05$). TA was only increased in $LLR_{120-150}$ ($P_t < 0.05$) after RBT and in $LLR_{150-210}$ after CBT ($P_t < 0.05$, **Table 1**). Interaction effects (time × group) were observed for SOL, TA, and RF in $LLR_{120-150}$ [SOL $F(1,17) = 4.85$, $P < 0.05$, TA $F(1,17) = 5.48$, $P < 0.05$, RF $F(1,16) = 10.71$, $P < 0.05$]. Effect sizes varied between trivial up to large effect sizes (**Table 1**).

Muscle Onset Latencies

Onset latencies diminished significantly for the stance perturbations in RF, BF, and SOL in the RBT-group and in SOL for the CBT-group (**Table 2**, for absolute values cf. **Supplementary Table S2**).

Kinematics

Grand means of the ankle, knee, and hip joint kinematics are displayed in **Table 3** and illustrated in **Figure 4**. The rmANOVA revealed significant time effects: ankle angular velocity decreased ($P < 0.05$), whereas knee joint amplitude increased for both groups in response to training ($P < 0.05$). Additionally, hip angular velocity was reduced after RBT only ($P_t < 0.05$).

Significant percentage changes in neuromuscular and kinematic data were more pronounced after RBT compared to CBT, which is illustrated in larger effect sizes ranging between medium to large sizes for RBT and small to medium sizes for CBT (Cohen's d, cf. **Table 3**).

Protocol 2 – Marching in Place Perturbation

Neuromuscular Activity

Changes in iEMG activity in response to training are displayed in **Table 4** and illustrated in **Figure 3** (for coefficient of variation cf. **Supplementary Table S1**). The rmANOVA revealed significant time effects for both groups: SOL iEMG activity increased in SLR only, while RF iEMG activity increased in SLR and MLR ($P < 0.05$). For RBT, muscle activity of RF, BF, and SOL was enhanced in $LLR_{120-150}$ and $LLR_{150-210}$ significantly ($P < 0.05$). For CBT, shank muscle activity increased significantly in $LLR_{120-150}$ and showed a tendency in $LLR_{150-210}$, in the thigh, muscle responses increased in RF, only ($P_t < 0.05$, **Table 4**). Interaction effects (time × group) were observed for BF in $LLR_{120-150}$ [BF $F(1,16) = 6.77$, $P < 0.05$].

Larger effect sizes were reached for CBT versus RBT in the SLR (large versus medium effect sizes). In the MLR, greater effect sizes were evident for RBT (medium to large effect sizes) (**Table 4**). In $LLR_{120-150}$, in almost all muscles, effect sizes reached large effects; in $LLR_{150-210}$ this is still true for SOL for RBT (**Table 4**).

Muscle Onset Latencies

Onset latencies diminished significantly in RF and BF in the RBT-group (**Table 2**). No significant changes were manifested for the CBT group.

Kinematics

Table 5 and **Figure 4** display grand means of the ankle, knee, and hip joint kinematics. The rmANOVA revealed a significant time effect illustrated in reduced ankle angular velocity for both groups ($P < 0.05$). For the RBT-group, hip angular velocity also decreased ($P_t < 0.05$).

Effect sizes of significant percentage changes ranged from medium to large for RBT and from small to large for CBT.

DISCUSSION

The purpose of the study was to identify differences between RBT and CBT and to clarify whether RBT improves kinematic and neuromuscular responses associated with the balance recovery after slips. The outcomes of this study outline ***reflex phase- and segment-specific adaptations*** dependent on the context of the movement task: (a) a facilitation of neuromuscular activation

TABLE 1 | Neuromuscular data during stance perturbation.

	Protocol 1: Stance perturbation					
	Group	**Δ pre/post**	P_t	**d**	**P**	η_p^2
SLR iEMG [%]						
RF	RBT	**+60 ± 93**	**0.01**	0.95	$F(1,16) = 5.390$, $p =$ **0.034**	0.252
	CBT	+46 ± 120	0.06	0.55		
BF	RBT	+88 ± 209	0.06	0.62	$F(1,14) = 3.846$, $p = 0.070$	0.216
	CBT	+45 ± 198	0.19	0.33		
TA	RBT	+1 ± 60	0.48	0.02	$F(1,13) = 0.057$, $p = 0.815$	0.004
	CBT	−5 ± 31	0.25	0.23		
SOL	RBT	−11 ± 39	0.14	0.40	$F(1,16) = 0.038$, $p = 0.847$	0.002
	CBT	+11 ± 57	0.20	0.29		
MLR iEMG [%]						
RF	RBT	+29 ± 70	0.05	0.60	$F(1,16) = 5.892$, $p =$ **0.027**	0.269
	CBT	+22 ± 67	0.09	0.47		
BF	RBT	+62 ± 182	0.11	0.50	$F(1,13) = 1.299$, $p = 0.275$	0.091
	CBT	−6 ± 35	0.25	0.26		
TA	RBT	+12 ± 61	0.23	0.30	$F(1,13) = 0.614$, $p = 0.447$	0.045
	CBT	−3 ± 36	0.35	0.13		
SOL	RBT	−9 ± 45	0.21	0.29	$F(1,16) = 0.006$, $p = 0.939$	<0.001
	CBT	+5 ± 48	0.34	0.15		
LLR iEMG [%]						
RF	RBT	+1 ± 55	0.48	0.02	$F(1,16) = 2.194$, $p = 0.158$	0.121
	CBT	+32 ± 82	0.06	0.56		
BF	RBT	+42 ± 97	0.06	0.64	$F(1,13) = 1.646$, $p = 0.222$	0.112
	CBT	+39 ± 221	0.25	0.25		
TA	RBT	+26 ± 159	0.27	0.24	$F(1,14) = 0.361$, $p = 0.558$	0.025
	CBT	0 ± 43	0.49	0.01		
SOL	RBT	−10 ± 46	0.20	0.30	$F(1,16) = 0.544$, $p = 0.471$	0.033
	CBT	+26 ± 113	0.16	0.34		
$LLR_{120-150}$ iEMG [%]						
RF	RBT	**−17 ± 39**	**0.03**	0.65	$F(1,15) = 4.607$, $\mathbf{p = 0.049}$	0.235
	CBT	**+134 ± 218**	**0.01**	0.90		
BF	RBT	**+123 ± 157**	**<0.01**	1.14	$F(1,16) = 10.713$, $\mathbf{p = 0.005}$	0.401
	CBT	**+39 ± 72**	**0.02**	0.79		
TA	RBT	**+94 ± 129**	**<0.01**	1.06	$F(1,17) = 8.572$, $p = 0.009$	0.335
	CBT	+6 ± 114	0.41	0.08		
SOL	RBT	**+97 ± 105**	**<0.01**	1.35	$F(1,17) = 21.366$, $\mathbf{p < 0.001}$	0.557
	CBT	**+52 ± 80**	**<0.01**	0.95		
$LLR_{150-210}$ iEMG [%]						
RF	RBT	+27 ± 101	0.13	0.39	$F(1,17) = 8.464$, $\mathbf{p = 0.010}$	0.332
	CBT	**+96 ± 140**	**<0.01**	1.00		
BF	RBT	+50 ± 159	0.09	0.46	$F(1,16) = 2.704$, $p = 0.120$	0.145
	CBT	**+31 ± 58**	**0.02**	0.77		
TA	RBT	+14 ± 187	0.38	0.11	$F(1,17) = 2.502$, $p = 0.132$	0.128
	CBT	**+62 ± 81**	**<0.01**	1.11		
SOL	RBT	**+91 ± 93**	**<0.01**	1.42	$F(1,17) = 19.998$, $p <$ **0.001**	0.541
	CBT	**+125 ± 196**	**<0.01**	0.93		

Mean changes in iEMG (%) in the short- (SLR), medium- (MLR), and long-latency responses (LLR) after perturbations during quiet monopedal stance (protocol 1). iEMG data are normalized to baseline values. Significant changes in response to the training intervention (TTEST, P_t) with effect sizes (Cohen's d, d) as well as time interactions (rmANOVA, P) with effect sizes (Partial Eta Squared, η_p^2) are illustrated in bold in the right columns with P_t and $P < 0.05$.

in shank and thigh muscles was accompanied by (b) a reduced muscle onset latency and (c) a decline in angular velocity for the hip and ankle joint. Even though no interaction effects could be observed comparing RBT to CBT for the early reflex phases, effect sizes of RBT were more pronounced for recovery response adaptations with a greater emphasis on the proximal body

TABLE 2 | Onset latency of the electromyograms during stance and marching in place perturbation.

	Group	Δ pre/post [%]	P_t	d	P	η_p^2
Protocol 1: Stance perturbation						
RF	RBT	**−16 ± 13**	**<0.01**	1.70	$F(1,17) = 7.852$, p = 0**.012**	0.316
	CBT	+1 ± 23	0.44	0.05		
BF	RBT	**−11 ± 10**	**<0.01**	1.60	$F(1,17) = 0.151$, $p = 0.703$	0.009
	CBT	+8 ± 31	0.15	0.36		
SOL	RBT	**−5 ± 9**	**0.02**	0.75	$F(1,17) = 18.478$, p **< 0.001**	0.521
	CBT	**−12 ± 12**	**<0.01**	1.44		
Protocol 2: Marching in place perturbation						
RF	RBT	**−14 ± 9**	**<0.01**	2.22	$F(1,17) = 4.621$, **p = 0.046**	0.214
	CBT	+1 ± 25	0.45	0.04		
BF	RBT	**−12 ± 14**	**<0.01**	1.25	$F(1,16) = 2.436$, $p = 0.138$	0.132
	CBT	+8 ± 22	0.06	0.55		
SOL	RBT	−2 ± 8	0.17	0.33	$F(1,17) = 0.004$, $p = 0.947$	<0.001
	CBT	+2 ± 18	0.31	0.17		

Mean changes from pre to post in EMG onset latency (%) for the protocols 1 and 2. iEMG data are normalized to baseline values. Significant changes in response to the training intervention (TTEST, P_t) with effect sizes (Cohen's d, d) as well as time interactions (rmANOVA, P) with effect sizes (Partial Eta Squared, η_p^2) are illustrated in bold in the right columns with P_t and $P < 0.05$.

TABLE 3 | Kinematic data during stance perturbation.

	Protocol 1: Stance perturbation					
	Group	Δ pre/post	P_t	d	P	η_p^2
Amplitude [°]						
hip	RBT	−0.45 ± 1.39	0.14	0.42	$F(1,11) = 0.066$, $p = 0.802$	0.006
	CBT	+0.18 ± 1.25	0.28	0.18		
knee	RBT	+0.63 ± 1.49	0.05	0.51	$F(1,16) = 6.368$, p = **0.023**	0.285
	CBT	**+0.46 ± 1.11**	**0.04**	0.47		
ankle	RBT	+0.91 ± 4.26	0.20	0.25	$F(1,16) = 1.493$, $p = 0.239$	0.085
	CBT	+0.82 ± 4.06	0.20	0.27		
Velocity [degrees · s^{-1}]						
hip	RBT	**−6.66 ± 11.34**	**0.04**	0.65	$F(1,10) = 0.918$, $p = 0.360$	0.084
	CBT	+1.87 ± 13.77	0.29	0.18		
knee	RBT	−2.19 ± 23.29	0.35	0.16	$F(1,16) = 0.180$, $p = 0.677$	0.011
	CBT	+0.77 ± 22.22	0.44	0.05		
ankle	RBT	−10.26 ± 23.71	0.06	0.52	$F(1,14) = 4.627$, p = **0.049**	0.248
	CBT	−3.88 ± 15.72	0.17	0.25		

Depicted are averaged changes of amplitude (°) and velocity (degrees · s^{-1}) of the ankle, knee, and hip joint excursions after perturbations during quiet monopedal stance (protocol 1). Significant changes in response to the training intervention (TTEST, P_t) with effect sizes (Cohen's d, d) as well as time interactions (rmANOVA, P) with effect sizes (Partial Eta Squared, η_p^2) are illustrated in bold in the right columns with P_t and $P < 0.05$.

segment. Thus, our hypotheses are verified with the constraint that modified adaptations occur after both trainings.

Two aspects may be of considerable importance for the interpretation of these findings (Gallagher et al., 1984; Alexander et al., 1992; Timsina et al., 2017): the first one deals with training-induced *neuromuscular enhancement* (Hatzitaki et al., 2005; Granacher et al., 2006) and the second with *multi-segmental joint kinematic* associated with the recovery response for fall avoidance (Cham and Redfern, 2001; Bhatt et al., 2006).

Neuromuscular Enhancement

Both training modalities achieved an enhancement in iEMG activity concomitant with reduced onset latencies in relevant sets of muscles that antagonize the perturbation stimulus. Thereby, marching in place perturbation were compensated by quickly delivered reflex activations (SLR) in the distal body segment. Transmitted via Ia afferent pathways, this has been associated with stiffening of the ankle joint complex leading to an improved recovery of posture in previous research (Granacher et al., 2006). To counteract the torque induced by posterior surface translation, an increase in SOL activation in SLR may have caused the decline in ankle joint velocity to regain stability of the body after unexpected surface translation (Hatzitaki et al., 2005; Granacher et al., 2006). At the same time, enhanced SOL activity without any changes in TA activity might be a result of a training-induced improved intermuscular coordination associated with

TABLE 4 | Neuromuscular data during marching in place perturbation.

	Protocol 2: Marching in place perturbation					
	Group	**Δ pre/post**	P_t	***d***	***P***	η_p^2
SLR iEMG [%]						
RF	RBT	**+43 ± 84**	**0.02**	0.75	*F*(1,17) = 7.853, *p* = **0.012**	0.316
	CBT	**+80 ± 141**	**0.01**	0.83		
BF	RBT	+59 ± 200	0.13	0.43	*F*(1,14) = 2.624, *p* = 0.128	0.158
	CBT	+53 ± 220	0.18	0.35		
TA	RBT	−8 ± 37	0.21	0.30	*F*(1,15) = 0.438, *p* = 0.518	0.028
	CBT	+14 ± 54	0.14	0.37		
SOL	RBT	+46 ± 119	0.06	0.56	*F*(1,17) = 9.339, *p* = **0.007**	0.355
	CBT	**+52 ± 83**	**0.01**	0.90		
MLR iEMG [%]						
RF	RBT	**+26 ± 45**	**0.02**	0.82	*F*(1,16) = 6.968, *p* = **0.018**	0.303
	CBT	**+42 ± 95**	**0.04**	0.64		
BF	RBT	+38 ± 219	0.26	0.25	*F*(1,13) = 0.441, *p* = 0.518	0.033
	CBT	−1 ± 32	0.47	0.03		
TA	RBT	+9 ± 56	0.27	0.23	*F*(1,15) = 2.024, *p* = 0.175	0.119
	CBT	+10 ± 47	0.19	0.31		
SOL	RBT	+38 ± 113	0.09	0.49	*F*(1,17) = 2.938, *p* = 0.105	0.147
	CBT	+36 ± 108	0.09	0.49		
LLR iEMG [%]						
RF	RBT	−12 ± 39	0.12	0.44	*F*(1,14) = 0.086, *p* = 0.773	0.006
	CBT	+15 ± 78	0.24	0.28		
BF	RBT	+66 ± 212	0.11	0.46	*F*(1,14) = 1.941, *p* = 0.185	0.122
	CBT	+14 ± 124	0.34	0.16		
TA	RBT	+27 ± 98	0.14	0.40	*F*(1,16) = 0.828, *p* = 0.376	0.049
	CBT	−5 ± 32	0.26	0.22		
SOL	RBT	+5 ± 76	0.40	0.09	*F*(1,16) = 2.537, *p* = 0.131	0.137
	CBT	+59 ± 145	0.05	0.59		
$LLR_{120-150}$ iEMG [%]						
RF	RBT	**+85 ± 131**	**<0.01**	0.94	*F*(1,17) = 7.146, *p* = **0.016**	0.296
	CBT	+48 ± 189	0.15	0.37		
BF	RBT	**+185 ± 215**	**<0.01**	1.26	*F*(1,16) = 10.519, *p* = **0.005**	0.397
	CBT	−1 ± 158	0.49	0.01		
TA	RBT	+41 ± 256	0.24	0.24	*F*(1,17) = 5.392, *p* = 0.033	0.241
	CBT	**+86 ± 125**	**<0.01**	1.00		
SOL	RBT	**+96 ± 108**	**<0.01**	1.28	*F*(1,17) = 23.150, *p* < **0.001**	0.577
	CBT	**+110 ± 166**	**<0.01**	0.96		
$LLR_{150-210}$ iEMG [%]						
RF	RBT	**+50 ± 104**	**0.03**	0.69	*F*(1,17) = 5.410, *p* = **0.033**	0.241
	CBT	**+122 ± 261**	**0.03**	0.68		
BF	RBT	**+123 ± 214**	**0.01**	0.84	*F*(1,15) = 5.861, *p* = **0.029**	0.281
	CBT	+53 ± 139	0.08	0.55		
TA	RBT	+86 ± 258	0.09	0.48	*F*(1,17) = 4.879, *p* = **0.041**	0.223
	CBT	+46 ± 125	0.07	0.54		
SOL	RBT	**+89 ± 94**	**<0.01**	1.37	*F*(1,17) = 9.773, *p* = **0.006**	0.365
	CBT	+64 ± 190	0.08	0.49		

Mean changes in iEMG (%) in the short- (SLR), medium- (MLR), and long-latency responses (LLR) after perturbations during marching in place (protocol 1). iEMG data are normalized to baseline values. Significant changes in response to the training intervention (TTEST, P_t) with effect sizes (Cohen's d, d) as well as time interactions (rmANOVA, P) with effect sizes (Partial Eta Squared, η_p^2) are illustrated in bold in the right columns with P_t and $P < 0.05$.

a reduced antagonistic co-activation and a greater rate of force development (Behrens et al., 2015).

Concomitant to shank muscle activation, knee extensors (RF) were activated faster and more distinctly in early reflex phases (SLR and MLR) during marching in place perturbation. Augmented neuromuscular activation in the MLR and subsequent reflex responses (LLRs) are known to rely on the involvement of higher centers of the central nervous system

TABLE 5 | Kinematic data during marching in place perturbation.

	Protocol 2: Marching in place perturbation					
	Group	**Δ pre/post**	**P_t**	**d**	**P**	**η_p^2**
Amplitude [°]						
Hip	RBT	+0.02 ± 1.74	0.48	0.02	$F(1,13) = 0.023$, $p = 0.881$	0.002
	CBT	+0.03 ± 1.93	0.48	0.02		
knee	RBT	+0.46 ± 1.39	0.09	0.34	$F(1,13) = 0.087$, $p = 0.773$	0.007
	CBT	−0.11 ± 1.17	0.36	0.09		
ankle	RBT	−0.55 ± 1.35	0.06	0.43	$F(1,15) = 1.729$, $p = 0.208$	0.103
	CBT	−0.07 ± 0.94	0.39	0.05		
Velocity [degrees · s^{-1}]						
Hip	RBT	**−5.98 ± 12.44**	**0.05**	0.50	$F(1,13) = 1.026$, $p = 0.330$	0.073
	CBT	−0.33 ± 15.13	0.47	0.03		
knee	RBT	−5.32 ± 23.38	0.18	0.27	$F(1,10) = 1.179$, $p = 0.303$	0.105
	CBT	+2.25 ± 20.31	0.36	0.11		
ankle	RBT	**−12.13 ± 16.00**	**<0.01**	0.99	$F(1,14) = 6.074$, $p =$ **0.027**	0.303
	CBT	−4.16 ± 13.85	0.13	0.26		

Depicted are averaged changes of amplitude (°) and velocity (degrees · s^{-1}) of the ankle, knee, and hip joint excursions after perturbations during marching in place perturbation (protocol 2). Significant changes in response to the training intervention (TTEST, P_t) with effect sizes (Cohen's d, d) as well as time interactions (rmANOVA, P) with effect sizes (Partial Eta Squared, η_p^2) are illustrated in bold in the right columns with P_t and $P < 0.05$.

such as the midbrain and brainstem (MLR) or even the motor cortex (LLRs) and thus, can generate specified reaction to the stimulus (Jacobs and Horak, 2007). Knee extensors are known to be predominantly activated in the eccentric phase of initial foot contact for impact absorption and/or energy storage in the elastic elements (Lacquaniti et al., 2012). Enhanced knee extensor activation, in terms of fall prevention, could lead to segmental stabilization (Horak and Nashner, 1986; Schillings et al., 2000; Gruber and Gollhofer, 2004), a consolidation of COM vertically above the support surface (Pfusterschmied et al., 2013), and might, therefore, finally reduce the risk of falling during slips and stumbles.

With an emphasis on the later reflex phases (120 ms after perturbation), next to muscle activation in SOL-RF, also TA and BF activities were raised for both groups. While early reflex responses are provided by Ia afferent pathways, supraspinal pathways are involved in later reflex responses (Taube et al., 2008). Thus, greatest muscle activation in later reflex phases might point toward a strategy induced by the motor cortex to counteract the perturbation stimulus by means of fall avoidance.

Phase- and segment-specific distinctions for neuromuscular control of postural responses after perturbation are manifested. Significant interaction effects and greater effect sizes for RBT versus CBT indicate intervention-dependent adaptations. Training-specific distinctions exist for RF activity, which are more pronounced during stance perturbation after RBT, a training that simulates fall characteristic situations associated with the following paragraph.

Joint Kinematics and Multi-Segmental Strategy

The second aspect associated with fall avoidance deals with segmental joint kinematics. Despite the complex nature of falls, studies show that falls can be significantly reduced by reducing fall risk factors (Rubenstein, 2006). In the current investigation, enhanced muscle activation was accompanied by reduced angular velocity during stance (ankle joint) and marching in place perturbation (ankle and hip joint). Augmented ankle and hip joint velocities have been identified to be distinctive in fallers and vice versa, reduced hip and ankle joint velocities in non-fallers (Lee and Kerrigan, 1999; Lord et al., 2001). With reference the aforementioned articles, our results can be interpreted as follows: participants frequently exposed to the fall situation in training – as it occurs during RBT – may learn from the risk situation (Pai et al., 2010; Bhatt et al., 2011), adapting their motor behavior (Bhatt et al., 2011) by activating skeletal muscles appropriately to counteract postural deterioration and restrict joint movement and velocity (Cham and Redfern, 2002; Bhatt et al., 2006; Pai et al., 2010; Parijat and Lockhart, 2012; McCrum et al., 2014). Thereby, significantly elevated knee deflections – well pronounced after RBT and CBT – are in line with evidence during stance perturbation (Di Giulio et al., 2013) and slipping situations in locomotion (Cham and Redfern, 2001; Bhatt et al., 2006). Knee joint flexion led to a lowering of COM height and an immediate postural unloading of the perturbed foot (Sawer et al., 2017); both are associated with a rapid reacquisition of a stable COM during unpredictable slips known to be essential to secure postural control and to reduce fall risk (Schillings, 2005; Bhatt et al., 2006). Inter alia, a smaller knee deflection has been identified in patients with postural instabilities (Horak et al., 2005; Bakker et al., 2006) and is associated with enhanced joint rigidity (Chmielewski et al., 2005) and reduced range of motion (Hatzitaki et al., 2005).

These findings with greater effect sizes after RBT indicate that this training caused a pronounced shift in multi-segmental organization using a stiff ankle joint for immediate compensation, a deflected knee for the reacquisition of a stable

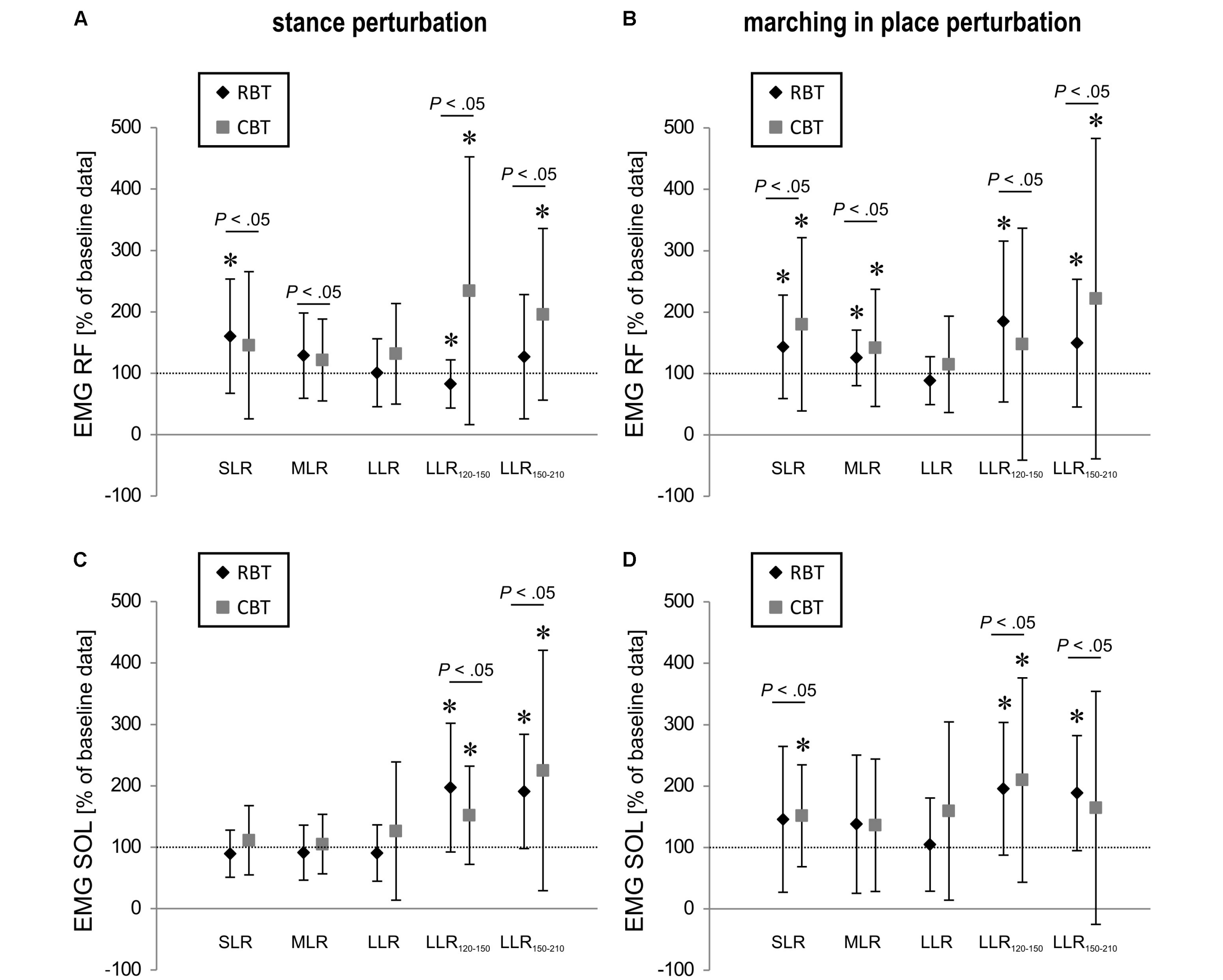

FIGURE 3 | Neuromuscular data of leg muscle activity. Changes of iEMG (in %) of the RF and soleus (SOL) muscles in the short- (SLR), medium- (MLR), LLR as well as 120–150 ms ($LLR_{120-150}$) and 150–210 ms ($LLR_{150-210}$) after stance perturbation **(A,C)** and after marching in place perturbation **(B,D)** are shown. Data are normalized to baseline values (the horizontal dashed line marks baseline values obtained before the interventions). Mean values are illustrated for both training groups – the RBT group and the CBT group. Significant pre/post differences are marked with * (P_t < 0.05), while significant time interactions are illustrated with bars (P < 0.05).

COM, and reduced hip joint velocities to control COM and trunk movements to achieve fast balance recovery and safe body equilibrium.

Distinction RBT and CBT and Functional Relevance

With neuromuscular control being the relevant aspect to determine the quality of postural response, adaptations of both interventions might be associated with a reduced fall incidence that could be relevant for fallers across the lifespan (Rubenstein, 2006; Bhatt and Pai, 2008; Granacher et al., 2010; Bieryla and Madigan, 2011). During stance perturbation, RBT as well as CBT enhanced RF activity and concomitantly reduced joint velocities in the proximal limb segment (knee and hip), but greater effects could be observed after RBT. This is in accordance with earlier investigations, emphasizing an intervention-specific adaptation due to specific training tasks (Freyler et al., 2016). The musculature of the proximal segments is well known to generate compensatory forces to restore equilibrium after slips and stumbles (Horak and Nashner, 1986; Hall and Jensen, 2002; Hatzitaki et al., 2005), and participants benefit from a two-segmental strategy to restore postural equilibrium and stabilize the trunk to prevent falling (Tang and Woollacott, 1998; Hall and Jensen, 2002; Schillings, 2005). During marching in place perturbation, early activation of distal *and* proximal activity is achieved after both training interventions, but effect sizes are even greater after CBT. Although this finding surprised us, comparable results for the shank musculature have been reported in more static paradigms in other studies

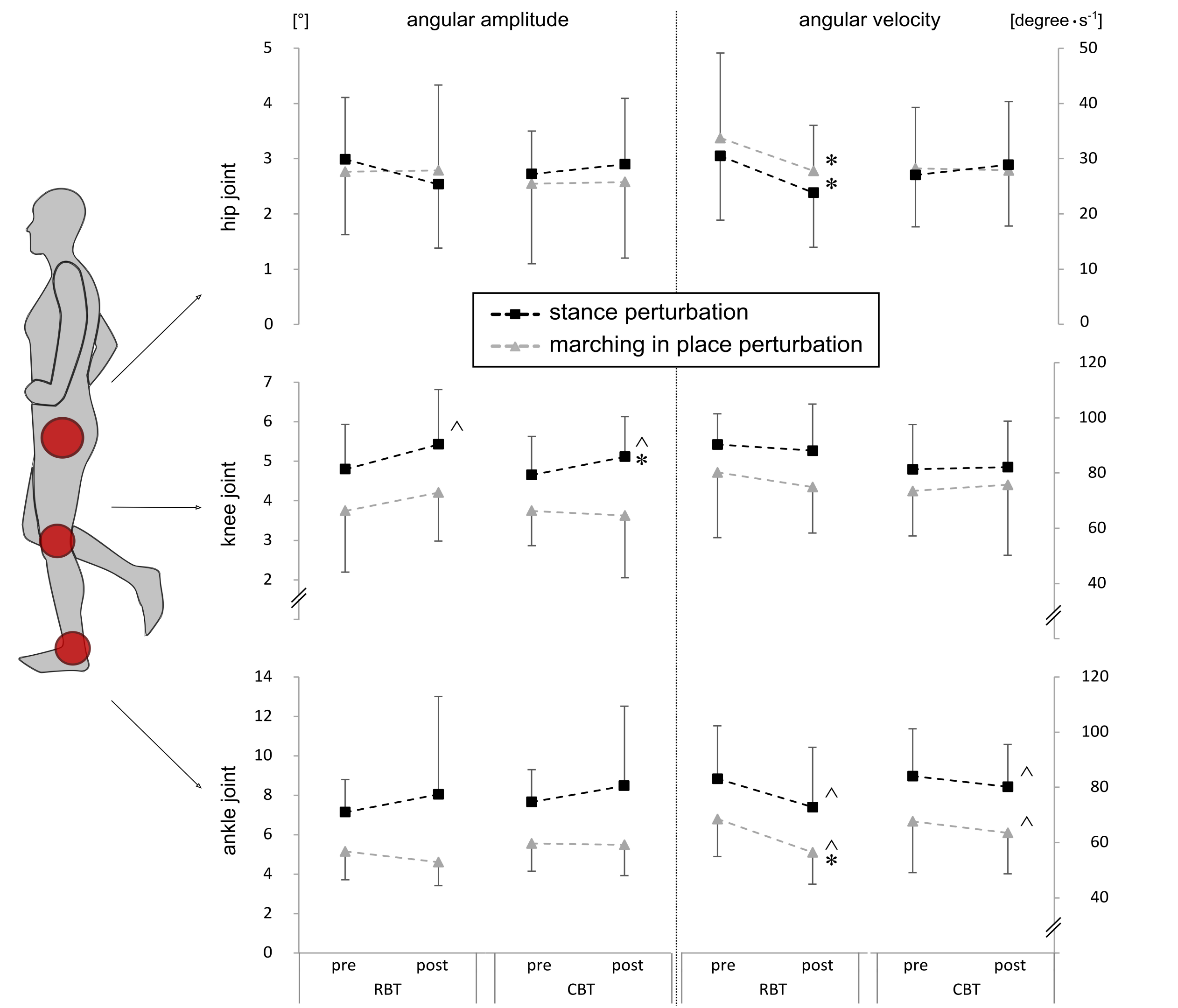

FIGURE 4 | Kinematic data of lower limb joints. Joint kinematics, such as amplitude [°] (left column) and velocity (degrees · s^{-1}) (right column), for the ankle, knee, and hip joint during stance perturbation (black rectangles, *protocol 1*) and marching in place perturbation (gray triangles, *protocol 2*) are shown. Mean values are illustrated for both training groups – the RBT group and the CBT group. Significant time effects are marked with ˆ ($P < 0.05$) and significant group comparisons (TTESTs) with * ($P_t < 0.05$).

(Gruber and Gollhofer, 2004; Taube et al., 2007; Freyler et al., 2016). Thereby, it can be supposed that CBT purposefully acts on the distal body segment (Freyler et al., 2016) making use of the knee joints that allows the stabilization of the COM within critical trajectories, and thus may reduce fall incidence (Pai et al., 2010; Bhatt et al., 2011).

Limitations

For a conclusive statement, it is crucial to consider the limitations of the study. Even though the methodological approach in the current paper was carefully chosen based on previous evidence, three limiting aspects could not be ruled out.

First, fall avoidance has to be differentiated from fall incidence: The population investigated is known to have great skills regarding motor control. Thus, it is an ideal population to investigate fall avoidance during stumbling situations. However, actual fall incidences were not observed which is why a differentiation of kinematic and neuromuscular characteristics cannot be provided based on current results.

Second, simulation of stumbling executed in laboratory paradigms as close to everyday life situations as possible is always biased. For instance, we had to inform participants about possible stumbling situations due to ethical reasons. Thus, they knew that a fall risk situation could occur. However, participants were not aware about the mechanical attributes of the surface translation and the onset of the perturbation occurred randomly so that anticipatory muscle activity could still be excluded.

Third, we compared two – due to evidence – *effective* training regimens for postural control. Therefore, both trainings demonstrated improved neuromuscular and kinematic adaptations. The effectiveness of both trainings might be the reason for missing interaction effects in fastest reflex responses. Even though the comparison of RBT and CBT is restricted to the comparison of effect sizes, this could still point toward differential adaptations.

CONCLUSION

Both training modalities improved reactive balance recovery after perturbations with adaptations being task-specific. Medium to large effect sizes were observed for neuromuscular and kinematic responses for RBT in response to stance perturbations, a task similar to the one which had been trained. This study provides basic evidence that neuromuscular control can be acquired rapidly by frequently reproducing the unexpected nature of real-life slipping situations within 4 weeks. It can be concluded that with repeated exposure to simulated slips, the central nervous system learns to choose a more effective muscle synergy and segmental organization to achieve fast balance recovery. Therefore, in dependence on the respective balance demands and independently of the stimulus itself, the participant can create a situation-specific postural stabilization strategy, and thus may have reduced the incidence of falling after RBT. While the current investigation is limited to neuromuscular adaptation in a healthy population, future studies across the lifespan might benefit from the current results by means of basic scientific evidence.

Conclusively, this study might set an essential cornerstone for further fall prevention investigations across the lifespan. Future investigations are needed which investigate, if especially high-risk fallers such as children and the elderly could benefit from RBT as a special form of CBT by counteracting age- or disease-associated degeneration of neuromuscular and kinematic strategies.

AUTHOR CONTRIBUTIONS

All authors made substantial contributions to the conception or design of the work, the acquisition, analysis, and interpretation of data for the work. Further they contributed drafting the work and revising it critically, they helped with the final approval of the version to be published and made the agreement to be accountable for all aspects of the work in ensuring that questions related to the accuracy or integrity of any part of the work are appropriately investigated and resolved.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2018.01075/full#supplementary-material

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Inertial Sensor-Based Gait and Attractor Analysis as Clinical Measurement Tool: Functionality and Sensitivity in Healthy Subjects and Patients With Symptomatic Lumbar Spinal Stenosis

S. Kimberly Byrnes[1], Corina Nüesch[1,2], Stefan Loske[1,3], Andrea Leuenberger[1], Stefan Schären[3], Cordula Netzer[3] and Annegret Mündermann[1,2,4]*

[1] Clinic for Orthopaedics and Traumatology, University Hospital Basel, Basel, Switzerland, [2] Department of Biomedical Engineering, University Hospital Basel, Basel, Switzerland, [3] Clinic for Spinal Surgery, University Hospital Basel, Basel, Switzerland, [4] Department of Clinical Research, University Hospital Basel, Basel, Switzerland

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Lars Donath,
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***Correspondence:**
Annegret Mündermann
annegret.muendermann@unibas.ch

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Objective: To determine if the attractor for acceleration gait data is similar among healthy persons defining a reference attractor; if exercise-induced changes in the attractor in patients with symptomatic lumbar spinal stenosis (sLSS) are greater than in healthy persons; and if the exercise-induced changes in the attractor are affected by surgical treatment.

Methods: Twenty-four healthy subjects and 19 patients with sLSS completed a 6-min walk test (6MWT) on a 30-m walkway. Gait data were collected using inertial sensors (RehaGait) capturing 3-dimensional foot accelerations. Attractor analysis was used to quantify changes in low-pass filtered acceleration pattern (δM) and variability (δD) and their combination as attractor-based index ($\delta F = \delta M^* \delta D$) between the first and last 30 m of walking. These parameters were compared within healthy persons and patients with sLSS (preoperatively and 10 weeks and 12 months postoperatively) and between healthy persons and patients with sLSS. The variability in the attractor pattern among healthy persons was assessed as the standard deviation of the individual attractors.

Results: The attractor pattern differed greatly among healthy persons. The variability in the attractor between subjects was about three times higher than the variability around the attractor within subject. The change in gait pattern and variability during the 6MWT did not differ significantly in patients with sLSS between baseline and follow-up but differed significantly compared to healthy persons.

Discussion: The attractor for acceleration data varied largely among healthy subjects, and hence a reference attractor could not be generated. Moreover, the change in the attractor and its variability during the 6MWT differed between patients and elderly healthy persons but not between repeated assessments. Hence, the attractor based on low-pass filtered signals as used in this study may reflect pathology specific differences in gait characteristics but does not appear to be sufficiently sensitive to serve as outcome parameter of decompression surgery in patients with sLSS.

Keywords: IMU, 6MWT, gait variability, gait changes, decompression surgery

INTRODUCTION

Human function is determined by the status of the neuromusculoskeletal system. Specifically, the interrelationship between structural aspects of the musculoskeletal and neuromuscular systems determine performance characteristics that are critical for facilitating normal movement conditions of the entire physiological range (Komi, 1984). However, many orthopedic diseases or conditions are associated with an abnormal, asymmetric, or variable gait pattern (Pirker and Katzenschlager, 2017). For instance, lumbar spinal stenosis (LSS)—a degenerative narrowing of the lumbar spinal canal—can influence mobility because of neuromuscular impairment. LSS associated radiating leg pain or pain in the lower back and/or the buttocks (Kreiner et al., 2013) frequently leads to a compromised ability to walk (Tong et al., 2007) resulting in abnormal or variable gait patterns. Symptoms in conditions affecting the neuromusculoskeletal system can be present at all times or intermittently, appear suddenly or have a creeping appearance. Hence, studying the effects of neuromuscular impairments such as those caused by LSS on ambulatory function during prolonged walking or specific functional tests such as the 6-min walk test (6MWT) provides important insights into normal and pathological neuromuscular function and performance. Yet, detailed knowledge on normal function, gait patterns, and their variability in a healthy population is a prerequisite for elucidating pathological function and gait patterns.

Patients with symptomatic LSS (sLSS) adopt strategies to avoid pain when performing daily activities such as walking that may manifest as changes in kinematic and kinetic gait parameters, and sLSS is often treated surgically by decompression surgery to relieve pain and improve mobility (Adachi et al., 2003). Patients with sLSS walk slower and with greater trunk sway, and their gait is less symmetric and generally more variable than in healthy persons (Suda et al., 2002). Moreover, cadence, stride length, gait speed, and symmetry increased and gait variability decreased in patients with sLSS treated with decompression surgery (Toosizadeh et al., 2015). However, to date it is unknown if these altered gait patterns are stable or if they change during continued walking, for instance, due to the onset of pain. The 6MWT is a standardized test first used to identify the submaximal level of functional capacity in patients with cardio-pulmonary diseases (Enright, 2003). Today, the 6MWT is commonly used for evaluating surgeries with pre- and postoperative measurements but can also capture the progress of therapeutic intervention, and has been used to assess changes in gait pattern and variability (ATS Committee on Proficiency Standards for Clinical Pulmonary Function Laboratories, 2002). Instrumented gait analysis may aid in the diagnosis by objectively revealing specific gait changes during a 6MWT associated with sLSS and monitoring rehabilitation processes after surgery.

Traditional instrumented gait analysis capturing kinematic and kinetic parameters is costly and time consuming. In recent years, gait analysis based on inertial sensors or measurement units have received increasing attention. The effective and convenient handling of inertial sensors compared to traditional multi-camera three-dimensional gait analysis may simplify and increase the efficiency of evaluating and interpreting gait data and hence is attractive for future clinical use (Tao et al., 2012). While inertial sensor data allow calculating kinematic parameters, alternative analyses based on the measured acceleration data have also been investigated. For instance, acceleration data collected by inertial sensors is considered a valid parameter for quantifying human movement (Godfrey et al., 2008). Specifically, acceleration data can be evaluated using attractor analysis, and changes in acceleration pattern and variability between conditions can be calculated and compared on an individual or group level (Vieten et al., 2013). Recently, attractor analysis has been used as an index to describe movement variability and motor fatigue in patients with multiple sclerosis (Sehle et al., 2014).

While attractor analysis of acceleration gait data may be a valuable tool for clinical applications, to date the variability in the attractor among healthy persons is unknown and data on orthopedic populations are lacking. Based on the literature, the following research questions arise:

- Is the attractor for acceleration gait data similar among healthy persons and can a reference attractor be defined?
- Does the attractor for acceleration gait data change during the 6MWT in patients with sLSS?
- Are the exercise-induced changes in the attractor in patients with sLSS greater than in healthy persons?
- Are the exercise-induced changes in the attractor affected by surgical treatment?

Answering these questions will lay the foundation for the potential use of attractor analysis of acceleration gait data as clinical tool in the assessment of diseases and conditions affecting gait and for gaining further insight into their pathomechanisms.

MATERIALS AND METHODS

Participants

Twenty-four older healthy participants [15 female; mean $\pm$ 1 standard deviation, age: 59.9 $\pm$ 10.5 years; body mass index (BMI): 24.0 $\pm$ 3.5 kg/m^2; **Table 1**] were recruited from

TABLE 1 | Mean (1 standard deviation) demographic information of participants.

Parameter	Healthy	Patients	*P*-value
Sex (male/female)	9/15	11/8	
Age (years)	59.9 (10.5)	73.8 (5.3)	**<0.001**
Height (cm)	168.5 (9.7)	167.8 (9.2)	0.797
Weight (kg)	68.5 (14.8)	75.8 (9.3)	0.069
BMI (kg/m^2)	24.0 (3.5)	27.1 (4.1)	**0.010**
ODI (%)			
Baseline		27.9 (16.9)	
10-week follow-up		8.5 (13.0)	
12-month follow-up		11.4 (13.5)	
6MWD			
Baseline	410.7 (64.3)	361.4 (100.9)	
10-week follow-up		397.5 (90.5)	
12-month follow-up		400.4 (87.4)	
D (left foot, baseline) within-subject	0.99 (0.66)[a]	1.38 (0.96)[a]	
D (left foot, baseline) between-subject	3.16	3.62	

ODI, Oswestry Disability Index; 6MWD, 6-min walk distance; D, standard deviation of the cycles toward the attractor; [a]standard deviation of all D-values of all subjects or patients. Bold values, statistically significant difference between groups (P < 0.05).

local health and sports clubs. Exclusion criteria for healthy participants were: previous surgeries or joint replacements that could influence the gait pattern; the use of walking aids; and neurological or mental disorders. Nineteen patients (8 female; age: 73.8 ± 5.3 years; BMI: 27.1 ± 4.1 kg/m^2; **Table 1**) diagnosed with sLSS and scheduled for decompression surgery were included in this study. All patients were recruited between May and August 2016 before their scheduled decompression surgery. Exclusion criteria were: BMI above 35 kg/m^2; the use of walking aids; the inability to walk for 6 min; and neurological or mental diseases. This study was approved by the regional ethics committee and performed according to the Declaration of Helsinki. All participants were informed about the study protocol and provided written consent.

Experimental Methods

All participants completed a gait analysis with an inertial sensor gait analysis system while walking up and down a 30-m hallway for 6 min (6-min walk test, 6MWT). Healthy participants completed one gait analysis. Patients completed gait analysis on the day before decompression surgery and 10 weeks and 12 months after surgery to assess the influence of surgery on gait function. At each assessment, patients also completed the Oswestry Disability Index (ODI) Questionnaire to record their pain and perceived functional disability.

Gait Analysis During the 6MWT

For all gait analysis sessions, participants were instructed to walk up and down the same 30-m hallway for 6 min. Acceleration data were collected by the RehaGait® system (Hasomed GmbH, Magdeburg, Germany) comprising seven inertial measurement units (each comprising a triaxial accelerometer (±16 g); a triaxial gyroscope (±2000°/s); and a triaxial magnetometer (±1.3 Gs); sampling rate 400 Hz) and software provided by the manufacturer. The inertial sensors were placed on the pelvis and bilaterally on the feet, the shank, and the thigh. In our study, gait data of the first and last 30 m of the 6MWT were examined regarding changes in gait patterns and gait variability and the influence of walking exercise on gait function. After test completion, raw acceleration data were exported in csv format from the manufacturer's software. The distance walked during the 6MWT (6-min walk distance 6MWD) was recorded as a measure of gait performance.

Oswestry Disability Index Questionnaire (ODI)

The ODI is a self-administered valid and reliable questionnaire used to evaluate and plan further therapy and treatment options in patients with lower back pain. The ODI reflects important aspects of functional pain-related disability in activities of daily life captured by 10 items (pain intensity, personal care, lifting, walking, sitting, standing, sleeping, sexual life, social life, traveling). The sum of scores is presented in percent (0–20%—minimal disability, to 81–100%—patient bed-bound or claiming to be extremely limited by their symptoms). Changes in the ODI score can be used to monitor the patient's progression and is commonly used by physical therapists for therapy planning and patient outcome (Vianin, 2008).

Computational Methods

Attractor analysis was performed according to Vieten et al. (2013) on the raw acceleration data captured by the foot sensors of the RehaGait® system. The raw acceleration vectors were low-pass filtered at 4.5 Hz using a 4th order Butterworth filter (Woltring, 1990). The built-in software of the RehaGait® provided gait events that were used to cut the three-dimensional acceleration vectors into single strides at heel-strike. Consecutive strides were depicted as limit-cycles, where each cycle represented one stride. The attractor itself represents the mean cycle of all strides and is calculated as:

$$\begin{aligned}\overline{A}_{a,C}\left(\tau_j\right) &= \frac{1}{n}\sum\nolimits_{i=1}^{n}\overline{a}_{a,C}\left(i\cdot\tau_j\right)+\frac{1}{n}\sum\nolimits_{i=1}^{n}\overline{b}_{a,C}\left(t=i\cdot\tau_j\right)\\ &\approx \frac{1}{n}\sum\nolimits_{i=1}^{n}\overline{a}_{a,C}\left(i\cdot\tau_j\right).\end{aligned}$$

(C: beginning or end of walking test; a: right or left foot)

Three attractor parameters were defined to describe the acceleration data:

- δM describes the change in acceleration pattern between two conditions (here the first and the last 30 m of the 6MWT), and represents the difference between two attractors (greater δM corresponds to a greater change in acceleration pattern between the two conditions)

$$\delta M=\sqrt{\frac{1}{m\cdot v^2}\sum\nolimits_{j=1}^{m}\left[\left(\overline{A}_{r,B}\left(\tau_j\right)-\overline{A}_{r,E}\left(\tau_j\right)\right)^2+\left(\overline{A}_{l,B}\left(\tau_j\right)-\overline{A}_{l,E}\left(\tau_j\right)\right)^2\right]}$$

(m: number of values within the attractor; v: average speed; B: beginning; E: end of walking test; r: right foot; l: left foot)

- δD describes the change in variability around the attractor and represents the change in acceleration variability between conditions (greater δD corresponds to a greater change in variability)

$$\delta D = \sqrt{\frac{1}{m}\sum_{j=1}^{m}\left[\left(\overline{D}_{r,B}(\tau_j) - \overline{D}_{r,E}(\tau_j)\right)^2 + \left(\overline{D}_{l,B}(\tau_j) - \overline{D}_{l,E}(\tau_j)\right)^2\right]}$$

- δF is considered the attractor-based index and is the product of δM and δD.

A reference attractor was generated by calculating the mean of all attractor vectors of the control group. The standard deviation between the attractor vectors and the reference attractor was calculated to assess the between-subject variability.

Statistical Analysis

All statistical analysis and calculations were performed in SPSS Version 21 (IBM Corporation, Armonk, NY). The data was tested for normality using the Shapiro-Wilk-Test. The Mann-Whitney-U-Test was used to detect differences in attractor parameters and 6MWDs between patients and healthy subjects. Repeated measures analysis of variance (ANOVA) was performed to detect differences in attractor parameters, ODI score and 6MWD between assessments (baseline, 10-week and 12-month follow-up) within patients. Bonferroni *post-hoc* tests were applied to detect specific differences between the time points. The level of significance was set to .05 for all tests.

RESULTS

During the 6MWT healthy subjects walked on average a distance of 410.7 ± 64.3 m. The values of the attractor parameters and therefore the changes in acceleration pattern and variability during the 6MWT were small. The acceleration pattern and the variability around the attractor within subjects were similar for the first and last 30 m of the 6MWT (**Figure 1**). The attractor patterns between subjects differed greatly. The reference attractor of all healthy subjects and their individual attractors are shown in **Figure 2**. The between-subject variability around the reference attractor was about three times higher than the variability around the attractor within subjects (**Table 1**).

The acceleration patterns and the attractor for the first and last 30 m of the 6MWT at baseline, 10-week follow-up and 12-month follow-up for one patient are shown in **Figure 3**. Overall, the patients showed greater changes in their gait pattern during the 6MWT than healthy persons reflected by higher δM values at all assessments compared to healthy persons (**Table 2**). This difference was statistically significant at the 12 month-follow-up ($P = 0.008$).

The change in acceleration variability (δD) between the first and last 30 m of the 6MWT differed between patients and healthy persons (**Table 2**). δD values in patients were higher than those in healthy persons ($P < 0.05$ for all). The attractor-based index δF and combined value for gait quality and style (Vieten et al., 2013) was significantly greater in patients than in healthy persons at all assessments ($P < 0.05$ for all). Within the patient group, the attractor parameters did not differ significantly between assessments. Nonetheless, at the 10-week follow-up all three parameters tended to be higher than before surgery. The changes in acceleration pattern from the first to the last 30 m of the 6MWT and change in acceleration variability tended to be greater after surgery than at baseline.

The difference in mean 6MWD between patients and healthy persons decreased from 49.3 m pre-operatively to 13.2 m at 10 weeks to 10.3 m at 12 months although these differences were not significant (**Tables 1**, **2**). These differences in gait performance, parameters, and variability persisted after correcting for age and BMI.

DISCUSSION

The objectives of this study were to determine if the attractor for acceleration gait data is similar among healthy persons defining a reference attractor; if exercise-induced changes in the attractor in patients with sLSS are greater than in healthy persons; and if the exercise-induced changes in the attractor are affected by surgical treatment. Our results showed that the attractor for acceleration gait data varies largely among healthy subjects, and hence a reference attractor cannot be defined.

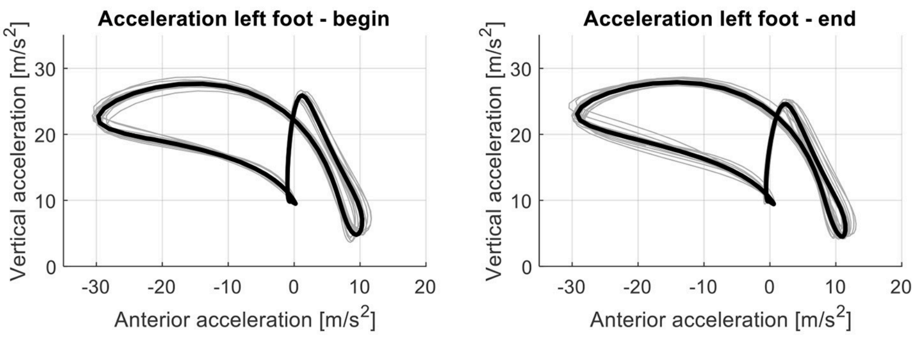

FIGURE 1 | The attractor (black line; mean of acceleration loops of all steps taken during 30 m) and acceleration loops of all steps taken at the beginning (first 30 m; left graph) and end (last 30 m; right graph) of the 6MWT (gray lines) for the left foot exemplary in one healthy subject.

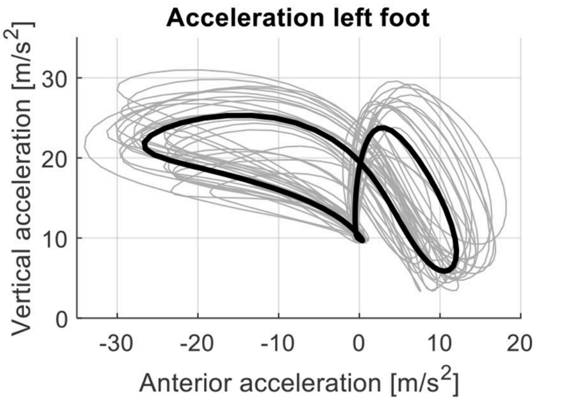

FIGURE 2 | The reference attractor (black line; mean of individual attractors) and individual attractors of all healthy persons ($N = 24$; gray lines) for the left foot. Individual attractors were based on acceleration data of the foot during the first minute of the 6MWT. Differences in patterns between subjects remained after correcting for walking speed.

Moreover, the change in the attractor and its variability during the 6MWT differed between patients and elderly healthy persons but not between repeated assessments. Hence, the attractor may reflect pathology specific differences in gait characteristics but does not appear to be sufficiently sensitive to serve as outcome parameter of decompression surgery in patients with sLSS.

Interestingly, we observed a large variability in the attractor for acceleration gait data among healthy subjects that was much greater than the intra-individual variability in acceleration patterns. Inter-subject variability in attractor patterns remained unaffected by normalization to walking speed. This result is particularly interesting because walking faster requires higher foot acceleration after the stationary period during ground contact. However, our results showed that foot acceleration patters are highly individual and may in fact be considered a person's unique "foot acceleration print." Moreover, acceleration patterns and their variability were similar at the beginning and end of the 6MWT indicated by small attractor parameters. These results suggest that acceleration patterns are stable throughout a non-fatiguing exercise in healthy persons. Based on these results we postulate that

- there is no reference attractor on foot acceleration data characterizing normal gait; and
- individual attractor patterns on foot acceleration may be a unique characteristic of a person's gait.

This result is in agreement with a recent study by Broscheid et al. (2018) who have shown that the fundamental walking pattern described by the acceleration attractor does not change with rehabilitation or after a single training session. The fact that correcting attractor patterns for walking speed did not reduce the variability in patterns between persons leads to the speculation that this pattern may be invariant to changes in walking speed associated with aging (Frimenko et al., 2015). This result may open up new opportunities in identifying groups of individuals who respond differently to intervention or being at higher risk for incurring an injury or disease. Factors that could contribute to the large variability include parameters that cannot be influenced such as sex or body height and parameters that can be modified such as body mass or parameters of neuromuscular performance (e.g., muscle strength or muscle coordination). For instance, Akalan et al. (2016) have shown that iliopsoas muscle group weakness resulted in related hip joint velocity reduction and stiff-knee gait during walking in healthy persons. However, the paradigm of a unique "foot acceleration print" similar to the previously proposed gait print (Broscheid et al., 2018) warrants further investigation.

At all assessments, the change in acceleration variability and in the attractor index during the 6MWT was greater in patients with sLSS than in healthy subjects. This result indicates that a 6MWT is sufficient to elicit functional changes in patients with sLSS. Changes in gait stability during the 6MWT have been previously reported in patients after stroke (Iosa et al., 2012). In another study on the same subjects and patients, we did not observe changes in spatiotemporal gait parameters and gait asymmetry during the 6MWT. Hence, it appears that attractor analysis on foot acceleration data is more sensitive for detecting changes in gait patterns during a relatively short functional gait test than traditional gait parameters. The attractor analysis implemented in our study was based on previous research of Sehle et al. (2014) who defined a fatigue index for patients with multiple sclerosis. In their study, patients were asked to walk on a treadmill until complete exhaustion, which occurred in less than 30 min in all patients, while healthy subjects were asked to walk for 30 min. The attractor index in our patients was below the fatigue index cut-off for motor fatigue ($\delta F = 4$) reported by Sehle et al. (2014). Because we limited the walking exercise to 6 min based on the widely accepted use of the 6MWT in clinical cohorts (Gao et al., 2014; Dunn et al., 2015; O'Brien et al., 2016; Keilani et al., 2017; Withers et al., 2017), it is remarkable that we still observed greater changes in attractor variability in patients than in healthy persons. It is unknown if patients in our study would have experienced even greater changes in attractor variability if they would have continued to walk until exhaustion or inability to continue due to sLSS symptoms. However, smaller attractor index values in our population are coherent with clinical observations of poorer gait function after exhaustion in patients with multiple sclerosis and motor fatigue than in patients with sLSS.

On average, patients with sLSS benefitted from decompression surgery: the ODI score decreased significantly from baseline to 10 weeks after decompression surgery and remained unchanged until 12 months after surgery. Higher ODI scores indicate less mobility and more pain. Thus, on average the decompression surgery reduced pain and symptoms. Similar results were found by McGirt et al. (2015), where the ODI improved significantly 12 months post-operatively. Moreover, the 6MWD improved significantly after decompression surgery in patients with sLSS from values below those of healthy persons to values similar to those in healthy persons. These results are

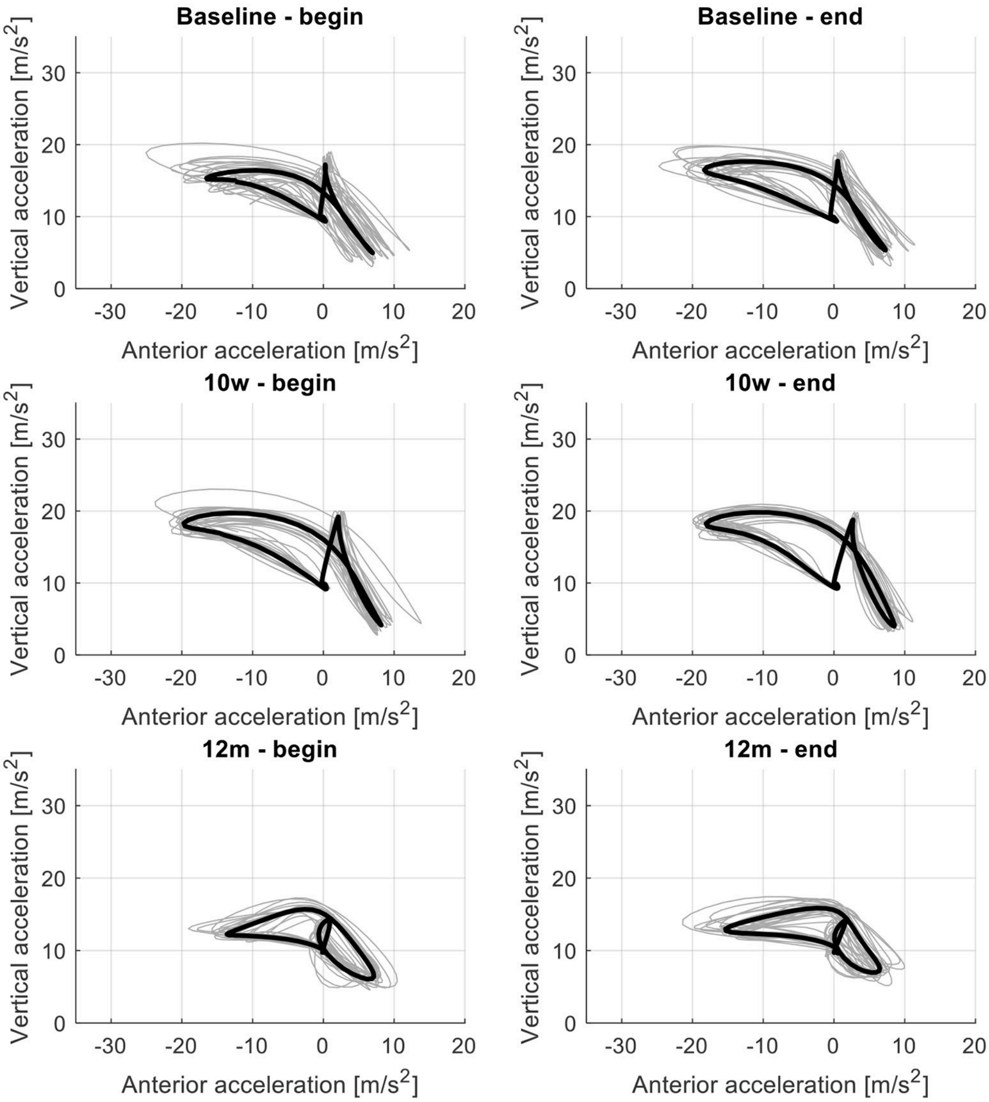

FIGURE 3 | The attractor (black line; mean of acceleration loops of all steps taken during 30 m) and acceleration loops of all steps taken at the beginning (first 30 m; left graph) and end (last 30 m; right graph) of the 6-min walk test (gray lines) for the left foot for the assessments at baseline **(Top)**, 10-week follow-up **(Middle),** and 12-month follow-up **(Bottom)** exemplary for one patient with sLSS.

in agreement with reports of improved functional capacity after decompression surgery (Smuck et al., 2018). This result demonstrates the importance of decompression surgery for regaining quality of life (Zarghooni et al., 2017).

In contrast to improvements in gait performance, exercise-induced changes in acceleration patterns and variability did not differ between assessments. We had expected that the exercise-induced changes in acceleration pattern and variability would decrease after decompression surgery. There are several possible explanations for this result. Because of the long duration of symptoms (at least 6 months) patients may have adopted a gait pattern preoperatively to compensate for pain characterized by greater variability that was unaffected by decompression surgery. Moreover, the long duration of symptoms and the claudication was likely associated with compromised muscle strength, impairment in trunk extensor muscle endurance, leg strength, leg strength asymmetry, or passive knee and ankle range of motion (Schmidt et al., 2017). Schmidt et al. also showed that impairment in trunk extensor muscle endurance, leg strength, leg ROM, and asymmetry of strength and ROM are associated with performance-based mobility (Schmidt et al., 2017).

The lack of statistically significant differences in exercise-induced changes in acceleration patterns and variability over time may be attributed to the heterogeneity of our patient population. We enrolled patients in our study independent of the type of sLSS. Hence, future studies may not only further explore the association between neuromuscular attributes

TABLE 2 | Mean values and statistical results of the attractor analysis in healthy persons and patients with sLSS.

		Norm (N = 24)	**δM**	**δD**	**δF**	**6MWD (m)**		
			2.33	**0.79**	**2.10**	**410.7**		
		Patients (N = 19)	***P*-value Norm vs. Patients**				***P*-value patients**	**Post-hoc**
δM	Pre	2.66	0.127				0.087	0.839[a]; 0.344[b];
	10w	2.99	0.169					1.00[c]
	12m	3.08	**0.008**					
δD	Pre	0.96		**0.007**			0.347	0.634[a]; 0.488[b]; 1.00[c]
	10w	1.24		**0.018**				
	12m	1.26		**0.001**				
δF	Pre	2.57			**0.007**		0.067	0.353[a]; 0.113[b]; 1.00[c]
	10w	3.70			**0.026**			
	12m	3.72			**<0.001**			
6MWD (m)	Pre	361.4				0.050	**0.017**	**0.002**[a]; **0.028**[b]; 1.00[c]
	10w	397.5				0.741		
	12m	400.4				0.501		
ODI (%)	Pre	27.9					**<0.001**	<0.001[a]; 0.001[b];
	10w	8.5						0.810[c]
	12m	11.4						

Statistically significant results are shown in bold (P < 0.05). δM, change in acceleration pattern; δD, change in acceleration variability; δF, attractor based index describing change in acceleration pattern and quality; pre, preoperative; 10w, 10-week follow-up; 12m, 12-month follow-up; 6MWD, 6-minute walk distance; ODI, Oswestry Disability Index score; [a]Difference between 10-week follow-up and baseline; [b]Difference between 12-month follow-up and baseline; [c]Difference between 12-month follow-up and 10-week follow-up.

and exercise-induced changes in acceleration patterns and variability but also relate this association to location of claudication based on medical imaging. Moreover, although not reported here, comorbidities present in some patients may have influenced their gait patterns independent of the limitations caused by the LSS. Further, the age of the patients may have played a role because geriatric patients experience skeletal and muscular changes (Marzetti and Leeuwenburgh, 2006) that can influence the patients' gait and neuromuscular attributes and therefore influence attractor patterns. However, including age in our statistical models did not affect the results. Nonetheless, this possibility could be further explored to investigate the pathomechanism of sLSS. Moreover, in contrast to treadmill walking assessed by Sehle et al. (2014), we assessed foot accelerations during overground walking. For instance, Bizovska et al. (2018) have shown that stride time variability and short-term Lyapunov exponents of acceleration data in all directions are greater for overground than for treadmill walking. However, it remains unknown if the change in acceleration variability would also depend on the walking condition.

In summary, we explored the use of attractor analysis on low-pass filtered acceleration data to assess changes in acceleration patterns and variability during a defined walking exercise. We presented answers to our research questions:

- Is the attractor for acceleration gait data similar among healthy persons and can a reference attractor be defined? We observed a large variability in attractors of acceleration patterns among healthy subjects precluding the definition of a reference attractor.
- Does the attractor for acceleration gait data change during the 6MWT in patients with sLSS? The change in the attractor and its variability during the 6MWT did not differ between repeated assessments.
- Are the exercise-induced changes in the attractor in patients with sLSS greater than in healthy persons? In general, patients had greater exercise-induced changes in acceleration patterns than healthy persons.
- Are the exercise-induced changes in the attractor affected by surgical treatment? Exercise-induced changes in acceleration patterns were not affected by decompression surgery.

Hence, overall patients had a less stable gait than healthy persons pre- and postoperatively. Multiple factors may play an important role for the difference between patients and the control group, such as the age of participants or other coexisting diseases, or secondary changes of sLSS including compromised neuromuscular performance. Readjusting gait patterns after surgery may require more than 12 months as indicated by a trend toward increasing changes in acceleration patterns post-operatively.

ETHICS STATEMENT

This study was carried out in accordance with the recommendations of Swissethics and Ethics Committee Northwest/Central (EKNZ). The protocol was approved by the Swissethics and Ethics Committee Northwest/Central (EKNZ). All subjects gave written informed consent in accordance with the Declaration of Helsinki.

AUTHOR CONTRIBUTIONS

SB, AM, and CNü conceived the study. SB, AL, and SL collected the data. SB and CNü processed the data. SB, AM, and CNü analyzed the data. All authors were involved in data interpretation. SB wrote the manuscript. All authors critically revised the manuscript and approved of the final version.

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frontiers in Physiology

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Impact of Multidirectional Transverse Calf Muscle Loading on Calf Muscle Force in Young Adults

Tobias Siebert[1], Manuel Eb[1], David S. Ryan[2], James M. Wakeling[2] and Norman Stutzig[1]*

[1] Department of Motion and Exercise Science, University of Stuttgart, Stuttgart, Germany, [2] Department of Biomedical Physiology and Kinesiology, Simon Fraser University, Vancouver, BC, Canada

It has been demonstrated that unidirectional transversal muscle loading induced by a plunger influences muscle shape and reduces muscle force. The interaction between muscle and transversal forces may depend on specific neuromuscular properties that change during a lifetime. Compression garments, applying forces from all directions in the transverse plane, are widely used in sports for example to improve performance. Differences in the loading direction (unidirectional vs. multidirectional) may have an impact on force generating capacity of muscle and, thus, on muscle performance. The aim of this study was to examine the effect of multidirectional transversal loads, using a sling looped around the calf, on the isometric force during plantarflexions. Young male adults (25.7 ± 1.5 years, n = 15) were placed in a prone position in a calf press apparatus. The posterior tibial nerve was stimulated to obtain the maximal double-twitch force of the calf muscles with (59.4 and 108.4 N) and without multidirectional transverse load. Compared to the unloaded condition, the rate of force development (*RFD*) was reduced by 5.0 ± 8.1% (p = 0.048) and 6.9 ± 10.7% (p = 0.008) for the 59.4 and 108.4 N load, respectively. No significant reduction (3.2 ± 4.8%, p = 0.141) in maximum muscle force (F_m) was found for the lower load (59.4 N), but application of the higher load (108.4 N) resulted in a significant reduction of F_m by 4.8 ± 7.0% (p = 0.008). Mean pressures induced in this study (14.3 and 26.3 mm Hg corresponding to the 59.4 and 108.4 N loads, respectively) are within the pressure range reported for compression garments. Taking the results of the present study into account, a reduction in maximum muscle force would be expected for compression garments with pressures ≥26.3 mm Hg. However, it should be noted that the loading condition (sling vs. compression garment) differs and that compression garments may influence other mechanisms contributing to force generation. For example, wearing compression garments may enhance sport performance by enhanced proprioception and reduced muscle oscillation. Thus, superposition of several effects should be considered when analyzing the impact of compression garments on more complex sport performance.

Keywords: transverse load, muscle compression, human gastrocnemius, muscle contraction dynamics, compression garments, muscle pressure

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****Correspondence:***
Tobias Siebert
tobias.siebert@inspo.uni-stuttgart.de

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INTRODUCTION

Neuromuscular performance depends on the properties of the neuromuscular system, which change during the lifespan, as well as on the interaction of the neuromuscular system with external forces from the environment. The majority of knowledge about skeletal muscle force generation is based on experiments on muscle fiber preparations (Gordon et al., 1966; Edman, 1988; Ranatunga et al., 2007) and isolated muscles (Hill, 1938; Katz, 1939; Scott et al., 1996; Siebert et al., 2015) that are freed from the neighboring tissue. In this isolated situation, the muscle or muscle fiber can deform freely without any external forces (except for the gravitational force). This is in contrast to the *in vivo* situation where muscles are surrounded by other muscles, connective tissue, and bones. These neighboring tissues may transfer forces to the muscle, modifying muscle architecture (Wick et al., 2018) and the force generation in the longitudinal (in the direction of the line of action, **Figure 1**, x-axis) direction. Examination of the interaction of muscles with external forces is important for a better understanding of movement generation and control. Myofascial force transmission via fascial connections between neighboring muscles has been shown to influence longitudinal muscle force (Carvalhais et al., 2013; Bernabei et al., 2015; Yucesoy et al., 2015). Furthermore, transverse compressive forces (perpendicular to the line of action) may be transferred between muscles during contraction thereby potentially influencing the longitudinal muscle force (Reinhardt et al., 2016).

External forces from outside the body can also act on muscles in addition to forces transferred from neighboring structures. Kinesio taping influences passive muscle shape and architecture (Pamuk and Yucesoy, 2015) and changes at least the initial conditions of a contraction. Muscle compression induced by transverse forces commonly occurs in daily life (for instance when wearing compression garments and during sitting) and the impact of compression garments on human performance is of great interest. Strength and power performance after fatigue seem to recover at a faster rate with the use of compression garments, for instance by increasing venous return and removal of metabolites (Hill et al., 2013), limitation of oedema (Partsch, 2012), and increased oxygen delivery to the tissue (Bringard et al., 2006). However, the effect of compression garments on acute sporting performance and muscle force is currently under discussion (Heneghan et al., 2012).

Studies on isolated rat muscles (Siebert et al., 2014a,b, 2016) revealed that muscle compression in unilateral transverse direction (**Figure 1**, z-axis) induced by a plunger reduced muscle work and force (up to 15%) performed in the longitudinal direction as well as the rate of force development (*RFD*) (up to 35%). Unidirectional transverse muscle loading may also be experienced during sitting. In contrast, compression garments (e.g., compression pants) induce external forces that act on the muscle from all directions in the transverse plane (multidirectional transverse loading, **Figure 1**, y-z plane). These different loading situations (unidirectional vs. multidirectional loading) lead to different passive muscle deformations and may subsequently lead to different contraction dynamics.

Experimental studies examining the impact of multidirectional transverse loading on muscle force and deformation are rare. Wakeling et al. (2013) reported reduced muscle thickness and fascicle pennation during cyclical isotonic plantarflexions with elastic compression bandages around the calf muscles. Azizi et al. (2017) constrained the muscle by a rigid tube and observed reduced muscle shortening and work. However, both studies didn't measure directly the amount of multidirectional transverse loading and its effect on muscle force.

Muscle deformations during contraction are strongly influenced by muscle architecture as well as active and passive muscle properties, which change dramatically during aging (Siebert et al., 2017). Thus, mechanical interaction between muscle and transversal loads might be age dependent. Here we induced a specific multidirectional transverse loading and measured its impact on muscle force and *RFD* in young adults. This was achieved by a sling that was looped around the human calf and loaded with two different weights (59.4 and 108.4 N) during electrically induced plantarflexions. It was hypothesized that multidirectional transversal muscle loading leads to a reduction in longitudinal muscle force.

MATERIALS AND METHODS

Subjects

Fifteen male young adults (height: 179 $\pm$ 6 cm; weight: 77.6 $\pm$ 7.2 kg; BMI: 24.1 $\pm$ 1.2; age: 25.7 $\pm$ 1.5 years) participated in this study. All subjects were informed about the risks of the experiments and gave their written consent. The study protocol was approved by the ethical committee of the university hospital of Tuebingen and conducted according to the latest declaration of Helsinki.

Measurement of Muscle Force and Lifting Height of the Load

We measured the force generated by the calf muscles during a double-twitch stimulation applied to the posterior tibial nerve. The double-twitch force was used as it is highly reproducible (Stutzig and Siebert, 2016) and commonly used to examine contraction dynamics in human experiments (Stutzig and Siebert, 2015a,b). The cathode was fixed in the popliteal fossa as close as possible to the posterior tibial nerve and the anode (5 $\times$ 10 cm) was fixed on the thigh approximately 2 cm proximal to the patella. A trigger box (DG2A, Digitimer, Herfordshire, United Kingdom) and a high current stimulator (DS7AH Digitimer, Herfordshire, United Kingdom) generated the electrical paired stimuli (pulse interval 10 ms, pulse duration: 1 ms). The maximal stimulation intensity was assessed using a ramp protocol. Starting at 10 mA, the current was increased by 10 mA every 10 s until the double-twitch force did not further increase. A 3D force plate (Type 9260 AA3, Kistler Instrumente AG, Winterthur, Switzerland, sample rate: 1,000 Hz) was used to measure the double-twitch force during stimulation. The vertical displacement of the load was recorded using a draw wire sensor (SX50, WayCon, Taufkirchen, Germany, sample rate: 1,000 Hz), which was connected to the load (**Figure 1**). The lift height Δh

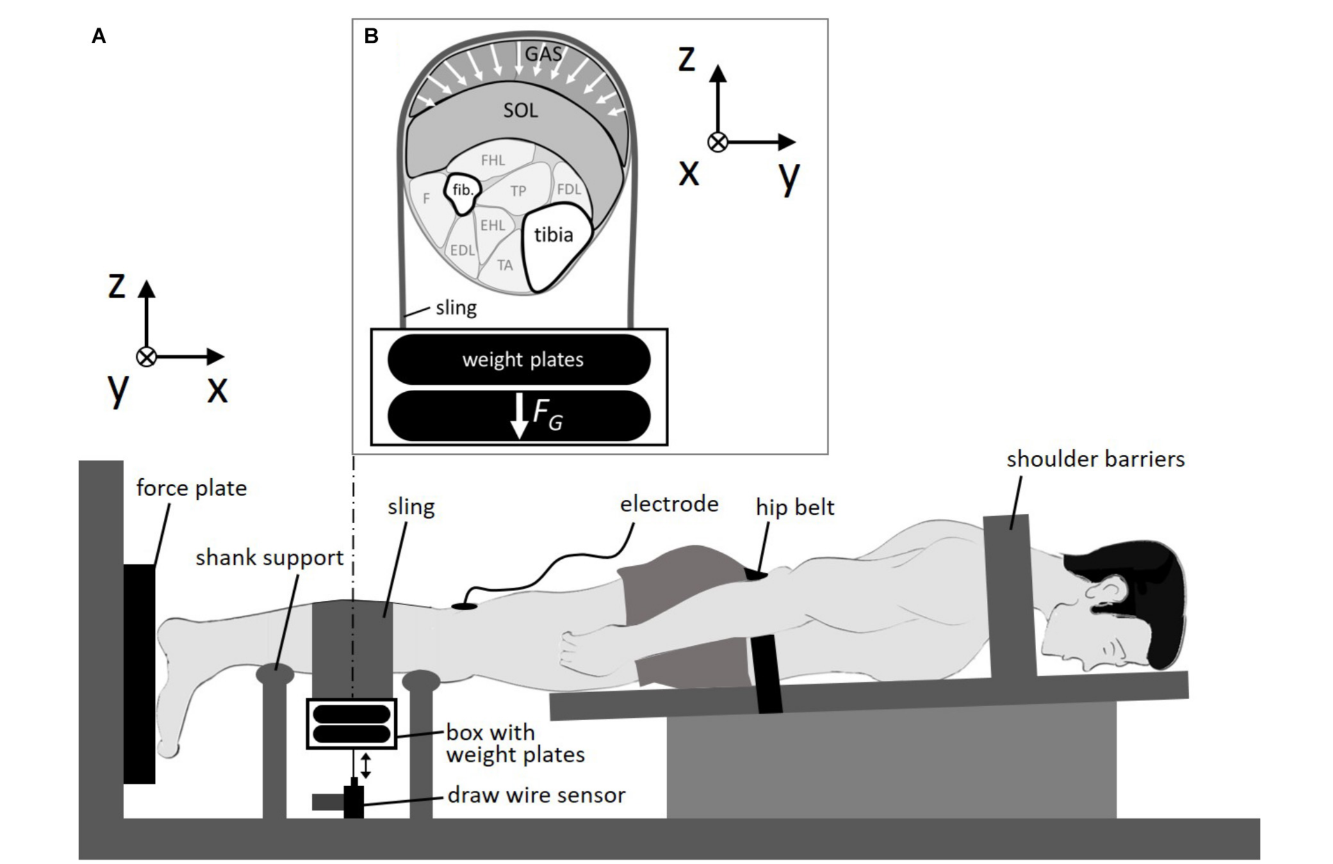

FIGURE 1 | Experimental setup. **(A)** Subject was fixed in a prone position. The sling was looped around the calf and loaded by weight plates. **(B)** Cross section of the right calf. Calf muscles were compressed by the sling in multiple directions in the transversal plane due to gravitational forces (F_G) of the weight plates. GAS: *M. gastrocnemius*; SOL, *M. soleus*; fib., fibula; FHL, *M. flexor hallucis longus*; FDL, M. flexor *digitorum longus*; TP, *M. tibialis posterior*; F, *M. fibularis*; EDL, *M. extensor digitorum longus*; EHL, *M. extensor hallucis longus*; TA, *M. tibialis anterior*; x, line of action/longitudinal direction; y-z plane, transversal plane.

of the load was calculated as the difference between its maximal height during contraction and its initial height in the passive state when the passive muscle was compressed. Work performed to lift the load W_{lift} was calculated by multiplying Δh and the gravitational force (load 1:59.4 N, load 2:108.4 N).

Experimental Protocol

At first the subjects performed a standardized warm-up consisting of 5 min of running on a treadmill (with 12 km/h), 3 sets of 10 calf raises (bilateral), and 10 repetitive calf jumps (bilateral). Afterward, the subjects were instructed to lie prone in a calf press apparatus with the right foot attached to a 3D force plate (**Figure 1**). The left foot was placed and secured beside the force plate. The stimulation electrodes were attached on the skin and the subject was immobilized in this position (full extended knee, ankle angle at 90°) with shoulder barriers and with a hip belt. Subjects then performed a further warm-up consisting of 10 submaximal (increasing from about 30 to 90% maximum voluntary contraction) isometric plantarflexions. First stimulation was performed about 5 min after the warm up.

The experiments started with five paired stimuli, applied to the posterior tibial nerve every 10 s at rest for the pretest baseline. Then, a leather sling (width: 15 cm, length: 75 cm) was looped around the calf at its most prominent bulge (**Figure 1**) covering the gastrocnemius muscle belly for the most part (about 70%). The longitudinal distance of the proximal border of the sling to the popliteal fossa (where the cathode was fixed) was about 7 cm. An aluminum box was hung on both ends of the sling for storage of the metal weights (5 or 10 kg). The draw wire sensor was connected via a hook to the bottom of the box. Due to the pulling force (3 N) of the draw wire sensor, and the deadweight of the aluminum box (0.5 kg) and sling (0.25 kg), loading of the box with 5 and 10 kg resulted in transverse loads of 59.4 and 108.4 N, respectively. We performed two series of loaded experiments (load 1:59.4 N, load 2:108.4 N) each consisting of 5 paired stimuli. In between the two series a rest time of 1 min was set without transverse muscle loading. Finally, five double-twitches every 10 s were performed without transverse muscle loading (post-test) to assess possible conditional changes during the experiment.

Data Analysis

A moving average filter (window length: 11 samples) was used for smoothing the force and kinematic data. The resultant force

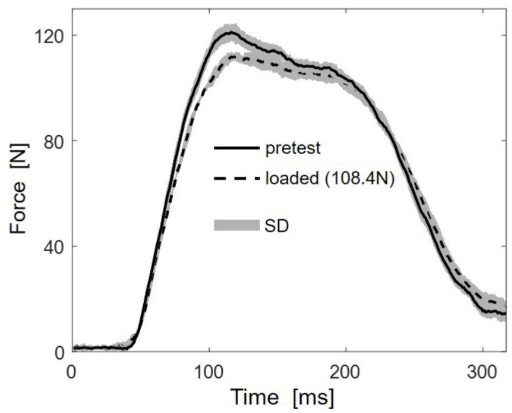

FIGURE 2 | Representative (subject 10) mean force-time traces of five stimulations for the highest load (108.4 N, dashed line) and the five unloaded reference contractions (pretest, solid line). Standard deviations (SD) are shown as gray areas.

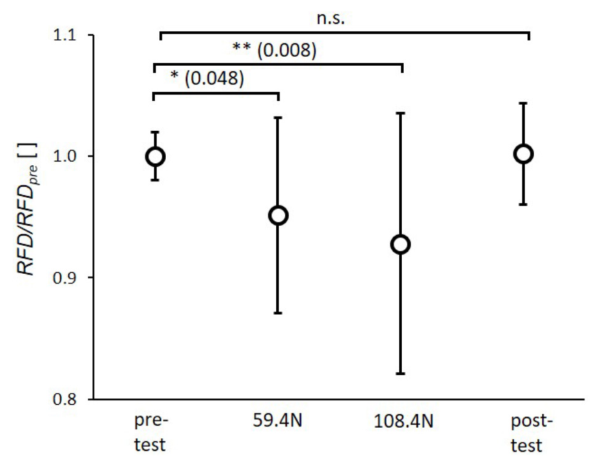

FIGURE 3 | Mean (circles) and standard deviation of rate of force development (*RFD*) for pretest, loading conditions (59.4 and 108.4 N) and post-test. For each subject *RFD* values of single twitches were normalized by mean subject specific rate of force development of pretest (RFD_{pre}). $^{*}p < 0.05$; $^{**}p < 0.01$; $^{***}p < 0.001$.

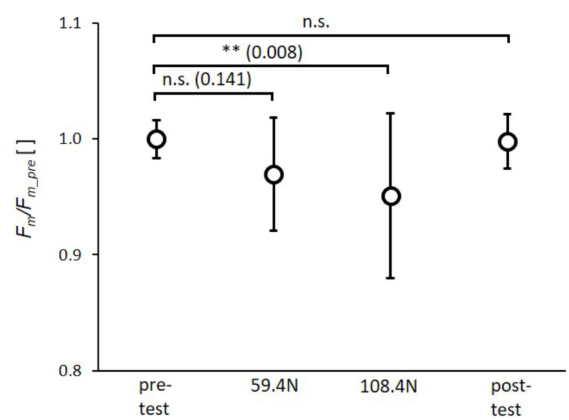

FIGURE 4 | Mean (circles) and standard deviation of maximal force (F_m) for pretest, loading conditions (59.4 and 108.4 N) and post-test. For each subject F_m values of single twitches were normalized by mean subject specific maximal force of pretest (F_{m_pre}). $^{*}p < 0.05$; $^{**}p < 0.01$; $^{***}p < 0.001$.

was calculated from the measured 3D force components for each double-twitch and used to quantify the maximal double-twitch force (F_m) and the *RFD* using custom-made Matlab scripts (MATLAB R2013a, The MathWorks, Inc., Natick, MA, United States). *RFD* was calculated as the force difference between 0.1 and 0.9 F_m divided by the corresponding required time. The mean and standard deviation of a stimulation series (e.g., 5 doublets at rest) was assessed and used for further statistical analyses.

Statistical Analyses

Data are presented as mean ± standard deviation. No indication of deviation from normal distribution was found using the Shapiro–Wilk Test. As lifting height of the load was measured for the loaded conditions only, a *t*-test for dependent samples was performed to test for differences between 59.4 and 108.4 N load. One-way repeated measures ANOVA (rANOVA) were conducted for the force parameters *(*F_m*, RFD*) to test for differences between conditions (pretest, 59.4 N load, 108.4 N load, and post-test). The level of significance was set at $p < 0.05$. Significant main effects or interactions of the rANOVA were probed using Bonferroni *post-hoc* tests. Effect size was determined using partial eta squared (η_p^2) and classified as follows: low $\eta_p^2 = 0.01$, medium $\eta_p^2 = 0.06$, and large $\eta_p^2 = 0.14$ (Cohen, 1988).

RESULTS

Multidirectional transverse muscle loading affected the force generation during plantarflexion. **Figure 2** shows the mean force-time curve of 5 stimulations for the higher load (108.4 N) and the mean of 5 unloaded reference contractions. The force-time curves indicate reduced *RFD* and reduced maximal force.

The statistical analysis showed significant effects of transverse loading for *RFD* ($p < 0.001$, $\eta_p^2 = 0.33$). The *post hoc* tests revealed a significant reduction of *RFD* by 5.0 ± 8.1% and 6.9 ± 10.7% between pretest and 59.4 N load and between pretest and 108.4 N load (**Figure 3**), respectively. There were no significant differences between pre- and post-test as well as between 59.4 and 108.4 N loading.

The rAnova revealed significant effects of muscle loading for F_m($p = 0.003$, $\eta_p^2 = 0.28$). The *post-hoc* test revealed a significant reduction in F_m by 4.8 ± 7.0% ($p = 0.008$) for the higher load (108.4 N). The reduction in F_m of 3.2 ± 4.8% induced by the lower load (59.4 N) was not significant (**Figure 4**). Furthermore, there were no significant differences between the 59.4 and 108.4 N load ($p = 0.141$) as well as between pre- and post-test.

Due to problems with the data acquisition of the lift height for one subject, only data of 14 subjects have been analyzed.

During plantarflexion contractions, the load was lifted by 4.6 ± 1.2 mm and by 5.6 ± 2.0 mm for the lower (59.4 N) and the higher (108.4 N) load, respectively. A *t*-test for dependent samples showed no significant differences between the lift heights ($p = 0.064$). The work performed to lift the load was higher ($p < 0.001$) for the 108.4 N load (616 ± 213 mJ) compared to the 59.4 N load (273 ± 70 mJ).

DISCUSSION

The aim of the present study was to examine the effect of multidirectional transverse muscle loading on contraction dynamics. Here we used a sling looped around the calf to apply multidirectional transverse muscle loading to the human calf muscles. Our experimental results demonstrate that muscle force and *RFD* were decreased by 4.7% and 6.9% in the loaded (108.4 N) condition, respectively, compared to the unloaded condition. In general, our observations are consistent with experiments applying unidirectional transversal forces on isolated rat muscles (Siebert et al., 2014a, 2016) and reporting reduced muscle forces, too. As transverse muscle loading (unidirectional and multidirectional) increases intramuscular pressure, a general reduction in muscle force can be explained by this effect (Siebert et al., 2018). However, application of transversal forces (0.64–2.6 N) by a plunger to the rat *M. gastrocnemius medialis* (GM) resulted in a more pronounced reduction of F_m (4.8–12.8%) and *RFD* (20–35%) (Siebert et al., 2014b). This might be due to differences in the normalized amount of loading, the loading condition (multidirectional vs. transversal), or the observed specimen (isolated rat GM vs. human calf muscles).

Normalization of transversal forces (0.64–2.6 N) used in rat experiments by the maximum isometric GM muscle force (F_{im} = 11.2 N; Siebert et al., 2014a) resulted in values of 6–23% of F_{im}. Applying a similar normalization for the transversal forces in this study (F_{im} = 4,500 N for human calf muscles at knee and ankle angles of 180° and 90°; Haxton, 1944; Stutzig and Siebert, 2015a) would yield much lower normalized muscle loading of 1–2% of F_{im} for loads of 59.4 and 108.4 N, respectively. As the reduction in muscle force and *RFD* increase with increasing loads (Siebert et al., 2014b, 2016), lower normalized muscle loading might partially explain lower force and *RFD* reduction in the present study.

Furthermore, differences in the loading condition might influence force-producing mechanisms. In the present study, applied forces act in multiple directions within the transverse (z-y) plane (**Figure 1**) compared to the unidirectional action of forces in z-direction in the plunger experiments. This impacts the muscle shape and architecture during passive muscle loading and changes at least the initial conditions of a contraction. Moreover, uni- and multidirectional transverse muscle loading may result in different deformations of the myofilament grid which might have impact on force generation capacity of cross bridges (Siebert et al., 2018).

Comparison of results of unidirectional and multidirectional transverse muscle loading is even more difficult as different specimens (isolated rat GM vs. human calf muscles) have been observed. For a single muscle, the reduction in muscle force can be mainly described by a simple model approach using a lever to convert transverse force and length change (i.e., lifting height) into longitudinal force and length (Siebert et al., 2014a, 2018). In the present study, multidirectional transverse forces acted on the gastrocnemius muscle (**Figure 1B**) and it is unclear if these forces were transmitted to the underlying *M. soleus* (SOL) as the depth of compression was not ascertained in this study. Therefore, specific contributions of GAS and SOL to calf deformations (and thus lifting height of the load) and reduction in longitudinal muscle force are unknown. In principle, transmission of transversal forces between neighboring muscles is possible (Reinhardt et al., 2016) allowing the different muscles to work together within a muscle package. Additionally, it is likely that calf muscle deformation and lifting height of the load was influenced by deformations of the smaller and deeper-lying FHL, TP, and FDL (**Figure 1B**). However, more precise examinations of 3D muscle architectures of calf muscles and their deformations during contraction are required to gain a better understanding of 3D interactions of synergistic muscles with each other as well as with external transverse forces.

There are only two studies that have applied multidirectional forces in the transverse plane to the muscle, but with deviating experimental approaches. Azizi et al. (2017) limited the radial expansion of isolated palmaris longus muscles ($n = 4$) of leopard frogs. The authors constructed small rigid plastic tubes, fitted to the diameter of the muscle, and placed them around the muscle. Small reductions of about 5% in maximum isometric force were reported, but were not significant (from 1.66 ± 0.36 N to 1.58 ± 0.30 N). Furthermore, the amount of muscle shortening and work output was reduced significantly during isotonic contractions against 25% F_{im} in the constraint condition. Wakeling et al. (2013) applied elastic compression bandages to the human lower leg. This external compression reduced muscle thickness, fascicle pennation and force generating capacity of fascicles, but the isotonic experimental design controlled the plantarflexions to have similar ankle torque and angular velocity. Summarizing, it can be stated, that restriction of radial muscle expansion by application of multidirectional transverse forces (e.g., by rigid tubes, elastic bandages, or slings) affects muscle deformation, muscle architecture and mechanical performance, which potentially influences muscle force and shortening as well as the ability of the muscle to do work in the longitudinal direction.

Measured mean forces of contractions without transverse muscle loading evoked by electrical stimulation are 165 ± 30 N. Considering a gearing of 2.8 (ratio of moment arms of the foot and the calcaneus about the angle joint; Arndt et al., 1998), this corresponds to Achilles tendon forces of 462 ± 84 N. Reported maximum values for Achilles tendon forces are in the range of 489–661 N during cycling at work loads between 88 and 265 W (Gregor et al., 1987) and in the range of 1,320–1,490 N during walking at speeds of 1.1 and 1.8 m/s (Finni et al., 1998). Thus, electrically evoked forces are in the lower range of forces produced during voluntary efforts.

Restriction of Muscle Deformation by Compression Garments and Increased ECM Stiffness

Restriction of muscle deformation by multidirectional transverse forces may be relevant when wearing compression garments as well as when examining the effects of increased stiffness of the collagenous extracellular matrix (ECM) on muscle performance. For example, compression garments are increasingly used by elite and recreational athletes, as they are thought to enhance performance. However, the results concerning increased acute sport performance are controversial (Beliard et al., 2015; Donath and Faude, 2016). Kraemer et al. (1996) found no effect on maximal force or power of the highest vertical jump, but did show enhanced mean force and power over 10 jumps. Mixed results for jumping performance and no effect on sprinting were reported by MacRae et al. (2011) regarding the use of compression garments, while Born et al. (2013) calculated small positive effect sizes for vertical jumping and sprint performance when compression garments were worn. Wearing compression garments had no impact on running (Ali et al., 2007) and cycling performance (Scanlan et al., 2008). Pressures applied at the calf by 12 different compression garments out of 24 studies analyzed in a review article (Beliard et al., 2015) are in the range of 8–39 mm Hg (corresponding to 0.10–0.52 N/cm^2) with a mean value of 19.9 mm Hg (0.26 N/cm^2). These pressures are within the range of mean pressures (transverse load divided by contact area between muscle and sling) of 0.19 $\pm$ 0.02 and 0.35 $\pm$ 0.03 N/cm^2 induced in the present study with loads of 59.4 and 108.4 N, respectively. Taking these results into account, pressure induced by compression garments might be in a range where no effect on muscle force can be expected (**Figure 4**, 59.4 N load corresponding to $\sim$0.19 N/cm^2 $\sim$14.3 mm Hg). Since we found a significant decrease in muscle force of 4.8% for the higher transverse load (**Figure 4**, load 108.4 N), a decrease in muscle force might be expected for compression garments that induce a higher pressure ($\geq$0.35 N/cm^2$\sim$26.3 mm Hg). However, results concerning the impact of compression garments on acute sport performance are controversial, as stated above. Regarding this, it should be noted that compression garments may influence other mechanisms contributing to force generation (which might be negligible in the current study) and counteract a potential reduction in muscle force induced by muscle compression. Wearing compression garments may enhance sport performance through enhanced proprioception (Perlau et al., 1995; Bernhardt and Anderson, 2005), increased muscle blood flow (Broatch et al., 2017), and reduced muscle oscillation (Sperlich et al., 2013) during repetitive exercise, which might reduce muscle pain and fatigue (Kraemer et al., 1998; Doan et al., 2003). Furthermore, measurements of complex voluntary performances (as in cycling, sprinting, jumping) have higher mean variations compared to experiments with electrical muscle stimulation on immobilized subjects. Thus, it might be possible that such small differences as an expected reduction in force of 5% induced by muscle compression cannot be resolved during complex voluntary movements exhibiting variations in force $>$5%.

In addition to compression garments, slings, or elastic bandages, increased stiffness of the ECM restricts muscle deformation. The mechanical properties of the ECM change with age, exercise, or illness (Kjaer et al., 2006; Gao et al., 2008; Lieber and Ward, 2013). Alnaqeeb et al. (1984) reported increased muscle stiffness with age that was closely correlated with the increase in endomysium, perimysium, and with total muscle collagen content. Moreover, ECM production is enhanced after chronic eccentric muscle loading (Heinemeier et al., 2007; Franchi et al., 2017). A stiffer ECM was found in children with spastic cerebral palsy (Smith et al., 2011) and as a consequence of skeletal muscle fibrosis (Lieber and Ward, 2013). Simulations of muscle contractions with increasing ECM stiffness restricted the muscle's ability to expand radially, which in turn compromises muscle shortening and performance (Azizi et al., 2017). Moreover, muscle architecture (e.g., pennation angle) and muscle properties change with age (Alnaqeeb and Goldspink, 1987; Binzoni et al., 2001; Siebert et al., 2017), and depend on the specific type of training (Franchi et al., 2014, 2017) resulting in age and training dependent muscle shape, size, and deformation during contraction. Better understanding of mechanical interaction between contractile tissue and transverse forces (e.g., induced by compression garments or slings) in young healthy adults may help to reveal how an age related deviation (e.g., in ECM stiffness) can compromise performance in older people or in response to neuromuscular pathologies.

AUTHOR CONTRIBUTIONS

TS, NS, JW, and DR conceived and designed the experiments. ME performed the experiments. TS, NS, and ME analyzed the data. TS and DR prepared the figures. All authors interpreted the results, edited, revised, and drafted the manuscript, and approved the final version of manuscript.

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Additional Intra- or Inter-session Balance Tasks Do Not Interfere With the Learning of a Novel Balance Task

Louis-Solal Giboin, Markus Gruber and Andreas Kramer*

Sensorimotor Performance Lab, University of Konstanz, Konstanz, Germany

Background: It has been shown that balance training induces task-specific performance improvements with very limited transfer to untrained tasks. Thus, regarding fall prevention, one strategy is to practice as many tasks as possible to be prepared for a multitude of situations with increased fall risk. However, it is not clear whether the learning of several different balance tasks interfere with each other. A positive influence could be possible via the contextual interference (CI) effect, a negative influence could be induced by the disruption of motor memory during consolidation or retrieval.

Methods: In two 3-week training experiments, we tested: (1) whether adding an additional balance task in the same training session would influence the learning of a balance task [first task: one-leg stance on a tilt-board (TB), six sessions, 15 × 20 s per session; additional task: one-leg stance on a slack line (SL), same amount of additional training]; (2) whether performing a different balance task (SL) in between training sessions of the first task (TB) would influence the learning of the first task. Twenty-six healthy subjects participated in the first experiment, 40 in the second experiment. In both experiments the participants were divided into three groups, TB only, TB and SL, and control. Before and after the training period, performance during the TB task (3 × 20 s) was recorded with a Vicon motion capturing system to assess the time in equilibrium.

Results: Analyses of variance revealed that neither the additional intra-session balance task in experiment 1 nor the inter-session task in experiment 2 had a significant effect on balance performance improvement in the first task (no significant group × time interaction effect for the training groups, $p = 0.83$ and $p = 0.82$, respectively, only main effects of time).

Conclusion: We could not find that additional intra- or intersession balance tasks interfere with the learning of a balance task, neither impairing it nor having a significant positive effect. This can also be interpreted as further evidence for the specificity of balance training effects, as different balance tasks do not seem to elicit interacting adaptations.

Keywords: contextual interference, rehabilitation, motor learning, varied practice, sensorimotor training, specificity, retrograde interference

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Davide Viggiano,
University of Molise, Italy
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****Correspondence:***
Louis-Solal Giboin
louis-solal.giboin@uni-konstanz.de

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INTRODUCTION

Falls have a major impact on the quality of life, especially in aged people and patients suffering from motor neuronal disorders (Sattin et al., 1990; Pavol et al., 2002; Gunn et al., 2013; Kim et al., 2013). Indeed, a fall for these populations is often a factor of hospitalization, immobilization, loss of autonomy, and increased mortality (Hannan et al., 2001; Jiang et al., 2005; Alvarez-Nebreda et al., 2008). One strategy to reduce fall occurrence consists in improving the balance of at risk populations (Gillespie et al., 2012). The success of this strategy is increased when the training is designed specifically to compensate pathological deficits or physical weaknesses occurring through the aging process (Horak et al., 2009; Gillespie et al., 2012). In addition, training one balance task seems to have quite task-specific effects (Giboin et al., 2015; Kummel et al., 2016; Donath et al., 2017). As one could encounter many types of balance perturbations in daily activities that could potentially lead to falls, a large spectrum of balance tasks should be included in the training. However, because of the reduced fitness abilities, but also the lack of motivation or time constraints of at risk populations, the number of tasks that can be trained, the overall volume and the time devoted to such training is very limited (Child et al., 2012; Khan et al., 2014; Mohler et al., 2014). Hence a very strong need to optimize the training sessions. Despite an extensive amount of scientific literature on motor learning and on the effects of balance training, less importance has been given to the ways of optimizing the motor learning of balance tasks. In particular, it is not clear whether the learning of two different balance tasks can affect the outcome, positively or negatively, of a short-term balance training.

Indeed, there is some evidence showing that variable practice, i.e., learning several tasks with a random or serial schedule during the training session, induces a lower acquisition but a better retention than constant training, i.e., learning only one task (Shea and Morgan, 1979; Shea and Kohl, 1990; Shea et al., 1990; Sekiya et al., 1996). This effect, called contextual interference (CI), is probably generated by the more difficult learning situation of the varied practice (Magill and Hall, 1990). Since CI effect may not be generalizable to every type of tasks (Magill and Hall, 1990; Brady, 2008), it seems of a great interest to test whether variable practice can enhance the learning of a balance task.

On the other hand, there is evidence from several fundamental motor learning studies that learning can be hindered and negative interference can occur when subsequently learning two or more novel motor tasks (Brashers-Krug et al., 1996). It has been proposed that when practicing a novel motor task, the motor memory of this task is unstable and can be disrupted by practicing another motor task (Shadmehr and Brashers-Krug, 1997), especially if the second task is quite similar (Lundbye-Jensen et al., 2011). The results of several visuomotor rotation and force-field adaptation studies suggest that this is not only the case when the second task is practiced shortly after the first one, but also when it is performed a day or even a week later (Caithness et al., 2004).

Thus, in the present study, practicing one balance task was compared to practicing the same balance task and an additional balance task, either in the same session or in a different session. In an applied setting, we wanted to assess whether adding the second balance task would facilitate learning as predicted by the CI framework, or on the contrary whether this would hinder learning as some fundamental motor learning and consolidation studies suggest.

MATERIALS AND METHODS

Participants

The experiments were approved by the local ethics committee and were in accordance with the latest revision of the Declaration of Helsinki. All participants gave written informed consent before starting the experiment. The 69 subjects were recruited among sport students and divided so 26 (15 males, 11 females, age 26 ± 7 years, height 176 ± 11 cm, body mass 74 ± 15 kg) of them participated in the experiment 1 and 40 (three drop-outs unrelated to the study, 32 male, 8 female, age 23 ± 3 years, height 180 ± 9 cm, body mass 74 ± 12 kg) of them participated in the experiment 2. All subjects had to be naïve to the test and training tasks.

Experiment 1: Additional Intra-session Task

The 26 subjects were divided in three groups: tilt board group (TB_1, $N = 9$), tilt board and slack line group ($TBSL_1$, $N = 9$), and control group (CON_1, $N = 8$). All subjects did a PRE and a POST practice measurement on the tilt board. The subjects were divided into the three groups after the PRE measurements so their initial balance ability on the tilt board was matched. The subjects from the TB_1 group practiced with only the tilt board for six sessions (three sessions per week for 2 weeks, at least one day of rest in between sessions), the $TBSL_1$ subjects had the same amount of tilt board practice as the TB_1 group, but in addition slack line practice during the same practice session for six sessions, and the subjects from CON_1 did not train, see also **Figure 1A**. The practice for TB_1 during each session consisted of 15 trials of 20 s on the tilt board separated by 10 s of rest. The practice of $TBSL_1$ consisted of 30 trials of 20 s, alternating between one trial on the tilt board and one trial on the slack line, also with 10 s of rest in between. Every five trials were separated by 1 min of rest.

Experiment 2: Inter-session Task

The 40 subjects that participated in the second experiment were divided into three groups: tilt board group (TB_2, $N = 12$), tilt board and slack line group ($TBSL_2$, $N = 14$), and control group (CON_2, $N = 14$). All subjects did a PRE and several POST measurements on the tilt board (after the third [POST1], the sixth [POST2] and the final practice session [POST3]). The subjects were divided into the three groups after the POST1 measurements so their initial balance performance on the tilt board during PRE and POST1 was matched. For all groups, the practice period lasted 3 weeks with 3 practice sessions per week (total of 9 practice sessions for TB_2 and $TBSL_2$, 6 sessions for CON_2). The subjects in TB_2 practiced only with the tilt board for the whole 3 weeks. The subjects in $TBSL_2$ practiced with the

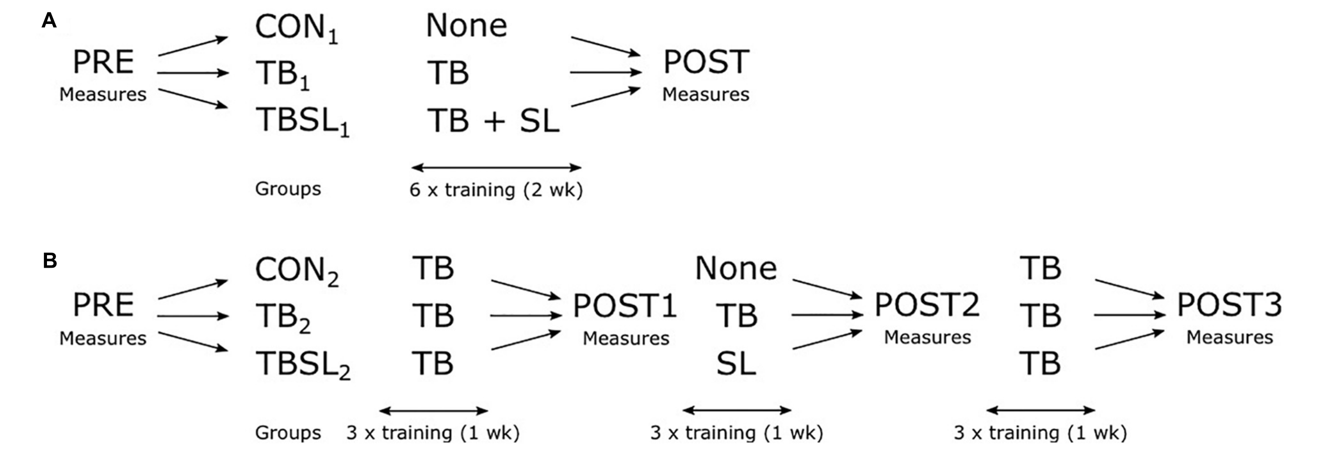

FIGURE 1 | Experimental design. **(A)** Experiment 1. After pre-practice measurements, subjects were divided into three groups (CON_1: no training, TB_1: tilt board training, $TBSL_1$: tilt board and slack line training). After six training sessions, post-practice measurements were done. **(B)** Experiment 2. After pre-practice measurements, subjects were divided into three groups. During week 1, all three groups trained with the tilt board. During week 2, CON_2 did not train, TB_2 trained with the tilt board, and $TBSL_2$ trained with the slack line. During week 3, all three groups trained with the tilt board. Measurements were done after week 1, 2, and 3.

TB during week 1 and week 3, but during week 2 they practiced with the slack line instead of the tilt board. The subjects in CON_2 also practiced with the tilt board during week 1 and 3, but did not train at all during week 2, see also **Figure 1B**. The practice with the tilt board consisted of 20 trials of 20 s separated by 10 s of rest. The practice with the slack line consisted of 20 trials of 20 s separated by 10 s of rest. Every five trials were separated by 1 min of rest.

Balance Tasks

Whatever the balance task, each subject had to balance on the preferred leg (always the same during the whole experiment), with hands on the hip.

The tilt board was custom-made and consisted of a wooden platform (25 cm × 25 cm × 1 cm) mounted on a semi-circular wooden structure with a height of 6.5 cm. The aim was to bring and maintain the platform of the tilt board into a horizontal position while standing on it with one leg. The tilt board was oriented so the axis of the semi-circular wooden structure was in parallel with the longitudinal axis of the foot. At the beginning of each trial (measurement or practice), the tilt board was always positioned with the same edge on the floor, with the preferred leg on the tilt board and the other leg firmly on the ground. Then, the subject had to lift this leg off the floor (**Figure 2A**) prior to bringing the platform of the tilt board into a horizontal position (**Figure 2B**). When the subject had to touch the ground with the free leg, i.e., to prevent a fall, this whole procedure was started again.

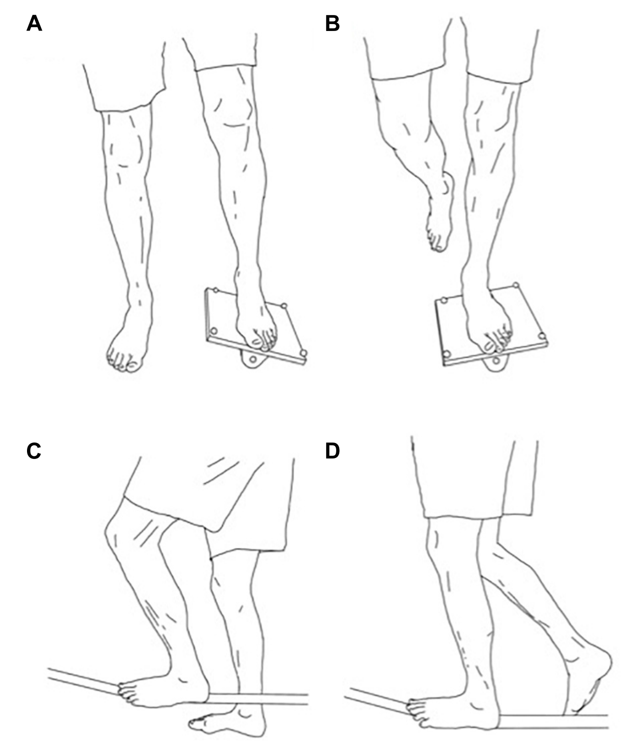

FIGURE 2 | Illustrations of the TB and SL tasks. **(A)** Starting position for the TB task. **(B)** Equilibrium position for the TB task. **(C)** Starting position for the SL task. **(D)** Equilibrium position for the SL task.

The slack line (Gibbon slacklines, Classic Line X13, width of 5 cm) was secured between two pillars separated by 5.6 m, at a height of 60 cm, with a mark in the middle of the line where the participants had to get on the slack line and try to balance them. Participants started each trial by standing on the side of the line, with the preferred leg on the line and the other leg standing firmly on the ground (**Figure 2C**). Then, the subject had to lift his leg off the ground and try to balance himself with only one foot on the slack line, while limiting the lateral oscillations of the slack line (**Figure 2D**).

During the practice and the measurements, investigators controlled and if necessary corrected the execution of the balance tasks.

PRE and POST Measurements

The PRE and POST measurements were done on the tilt board. They consisted of 5 trials of 20 s separated by 10 s of rest. The first 2 trials served as familiarization trials and only the last 3 were used to quantify the performance of each subject. POST measurements were done after at least one day of rest after the last practice session.

Data Collection and Analysis

Four reflective markers were placed on the corners of the tilt board, so that the position and angle of the board could be recorded and analyzed with a motion capture system at 200 Hz (Vicon Nexus, 12 T-series T40s cameras). The performance during a trial was evaluated by calculating the amount of time the tilt board was within a margin of $\pm 5°$ of the horizontal position [i.e., when the angle of the board with respect to the floor was between $-5°$ and $+5°$, see (Giboin et al., 2015)]. The performance was defined as the mean of the three recorded trials.

Statistics

Statistical tests were done with JASP (Version 0.8.3.1, University of Amsterdam). For both experiments, the changes in performance after the practice were assessed with a mixed design analysis of variance (ANOVA) with repeated measures, using time as repeated measure and group as between-subject factor.

RESULTS

Experiment 1: Additional Intra-session Task

As depicted in **Figure 3**, performance during the tilt board task increased after the six training sessions for TB_1 and $TBSL_1$. Indeed, when considering only the two training groups, the ANOVA showed no significant group*time effect ($F_{1,16} = 0.05$, $p = 0.83$), only a significant main effect of time ($F_{1,16} = 65.6$, $p < 0.001$). When adding the control group to the analyses to ensure that the time effect was not only due to test–retest improvements, there was a significant group*time effect ($F_{2,23} = 5.66$, $p = 0.01$).

Experiment 2: Inter-session Task

As depicted in **Figure 4**, the performance in the three groups increased similarly after the 3 weeks of training. When considering only the two training groups, the ANOVA showed no significant group*time effect ($F_{3,72} = 0.31$, $p = 0.82$), only a significant main effect of time ($F_{3,72} = 33.7$, $p < 0.001$). When adding the control group to the analyses, there was no significant group*time effect ($F_{6,111} = 0.62$; $p = 0.7$), only a significant main effect of time ($F_{3,111} = 49.76$; $p < 0.001$).

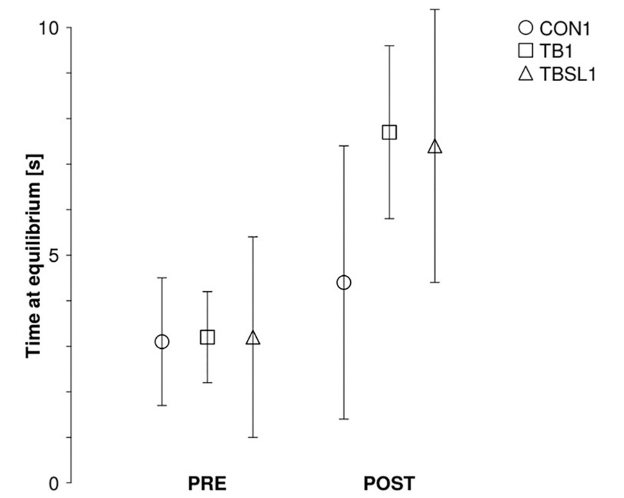

FIGURE 3 | Experiment 1, intra-session varied practice. Grouped data of the average performance in seconds on the TB from CON1 (circle, no training), TB1 (square, training with TB), and TBSL1 (triangle, training with TB, and SL) before (PRE) and after (POST) the six training sessions. Error bars represent standard deviations.

DISCUSSION

In this study, we found no evidence for a positive or negative interference on the learning of a novel balance task when adding a second balance tasks, neither when doing so in the same session nor in between sessions.

Additional Intra-session Balance Task

The addition of an intra-session balance task, i.e., variable practice, did not improve performance in the balance task to a greater extent than practicing just one novel balance task, i.e., constant practice, despite the double amount of total training volume. This lacking difference cannot be explained by a ceiling effect, since the measured performance remained lower than performance measured in experiment 2 after 3 weeks of training. However, the absence of effect from CI and variable practice is not unheard of when tested with complex whole-body tasks instead of for instance with simple pointing tasks (for review, see Brady, 2008). Several hypotheses may explain the absence of CI effect. First, because of the complexity of both tasks, the CI effect may have been too high and possibly induced an "overload" of the processing requirements, reducing the learning efficiency, especially at the beginning of the training but no more at a later stage, hence no difference with the constant practice group (Shea et al., 1990). Second, the balance task with the tilt-board (TB) may itself induce a high level of CI due to its complexity (Albaret and Thon, 1998). Indeed, as the board responds already to small bodyweight shifts, a large panel of different balance perturbation and balance recovery movement sequences can be seen throughout one training session. This variety of postural movements may by itself elicit a strong CI effect, which may already saturate the possible retention increase effect induced by high CI even with the addition of another task (Albaret and

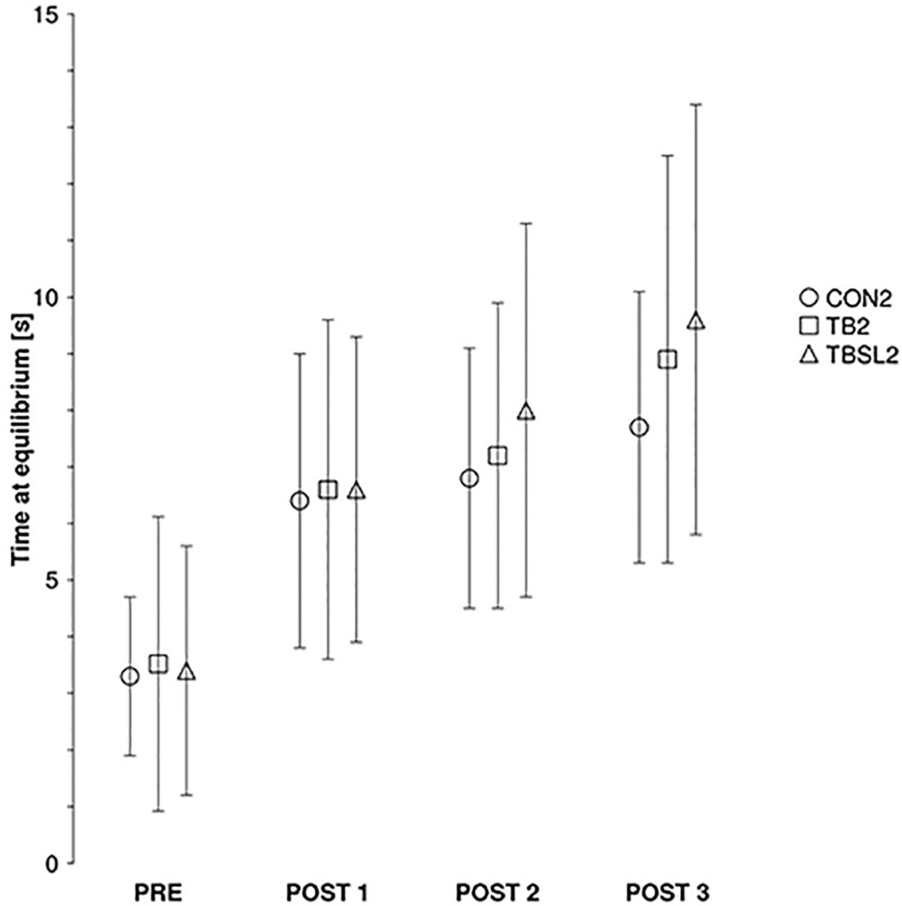

FIGURE 4 | Experiment 2, inter-session varied practice. Grouped data of the average performance in seconds on the TB from CON2 (circle, no training during week 2), TB2 (square, training with TB during the 3 weeks) and TBSL2 (triangle, training with SL during week 2) before (PRE), after 3 training sessions (POST1), 6 training sessions (POST2), and 9 training sessions (POST3). Error bars represent standard deviations.

Thon, 1998). Finally, it must be noted that even if no effect of retention was seen on the learning of a balance task after a short-term training, there was also no impairment observed. Indeed, it has been reported that intra-session practice of similar tasks could, when performed with a block design, induce retrograde interference. The resulting impairment in learning is possibly caused by memory consolidation disruption. The practice of two very similar tasks would in theory prevent synaptic change stabilization since both tasks share the same underlying neural networks (Lundbye-Jensen et al., 2011). In the present study, the lack of increase or impairment in the learning of the balance task despite the varied practice supports the concept that balance is more a sum of specific skills than a general ability (Giboin et al., 2015; Ringhof and Stein, 2018). Indeed, the lack of altered learning implies that both tasks were possibly too different to induce a CI effect, or alternatively, too different to share the same underlying neural networks and impair memory consolidation. In conclusion, even though no gain in learning efficiency could be found, the variable practice can still be recommended for short-term balance training if the goal is to learn several balance tasks.

Additional Inter-session Balance Task

The second experiment served a twofold purpose: investigate possible negative retrograde interference effects in a very applied setting, possibly caused by motor memory consolidation disruption (see above), and assess potential differences due to the timing of practice of the additional task: motor learning studies have shown that directly after learning, the motor memory of the task that was just practiced is still fragile, i.e., not consolidated yet (Brashers-Krug et al., 1996). If this phenomenon played an important role in the acquisition of a novel balance task, it should have been reflected in the results of the first experiment, where after each trial of the first task, a trial of the second–possibly interfering–task was performed. This was not the case, but retrograde interference has also been reported after days and even a week in visuomotor rotation and force field adaptation tasks (Caithness et al., 2004). To test the practical influence of this re-engagement of the motor memory that potentially makes it vulnerable to disruption again, we performed the second experiment. The results indicate that this effect does not seem to play an important role in the acquisition of novel balance tasks. If anything the group that practiced an additional novel balance task ($TBSL_2$) performed even slightly better in the first task, even though they did not train the first task during that time, in contrast to the other group (TB_2) that continued to practice the first task. This slight advantage of adding a second balance task might be explained by motivational issues, as practicing the same balance task for 3 weeks might be considered less exciting by some participants than when it is interspersed with 1 week of practicing a different novel task. In any case, the retrograde interference effect described for some motor learning studies using visuomotor rotation or force-field adaptation tasks was not observed in our study, indicating that this effect does either play a minor role in such an applied setting with complex whole-body tasks, and that other factors such as motivation play a more important role.

It is interesting to note that in the control group that did not train during the second week, the performance did not differ compared with the two training groups. However, robust recalls of a discrete motor task, even after longer retention test interval, have already been documented (Dail and Christina, 2004; Savion-Lemieux and Penhune, 2005). Moreover, we suggest that the slow learning phase was already reached after three training sessions. Then, at this stage of learning, one more week of training may not be sufficient enough to induce significant improvements, which could explain the lack of performance difference between the three groups. Thus, it must be noted that a possible performance difference could possibly appear with longer duration training.

CONCLUSION

The results of this study suggest that adding an additional distinct balance task within or in between practice sessions does not interfere with the motor learning of a novel balance task. The lack of interference, positive or negative, support the concept that balance training elicits mostly distinct task-specific adaptations that do not interfere with each other. From an applied perspective, sequentially practicing several balance tasks for rehabilitation or fall prevention can be recommended, as the additional tasks do not seem to hinder learning.

AUTHOR CONTRIBUTIONS

AK, MG, and L-SG conceptualized the study. AK and L-SG collected, analyzed, and interpreted the data. L-SG drafted the manuscript. AK and MG revised the draft.

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Matching Participants for Triceps Surae Muscle Strength and Tendon Stiffness Does Not Eliminate Age-Related Differences in Mechanical Power Output During Jumping

Matthias König[1], Svenja Hemmers[2], Gaspar Epro[1], Christopher McCrum[3,4], Thijs Maria Anne Ackermans[5], Ulrich Hartmann[2] and Kiros Karamanidis[1]*

[1] School of Applied Sciences, Sport and Exercise Science Research Centre, London South Bank University, London, United Kingdom, [2] Department of Mathematics and Technology, RheinAhrCampus Remagen, Koblenz University of Applied Sciences, Remagen, Germany, [3] NUTRIM School of Nutrition and Translational Research in Metabolism, Department of Nutrition and Movement Sciences, Maastricht University Medical Centre, Maastricht, Netherlands, [4] Institute of Movement and Sport Gerontology, German Sport University Cologne, Cologne, Germany, [5] School of Sport and Exercise Sciences, Liverpool John Moores University, Liverpool, United Kingdom

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***Correspondence:**
Matthias König
koenigm@lsbu.ac.uk

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Reductions in muscular power output and performance during multi-joint motor tasks with aging have often been associated with muscle weakness. This study aimed to examine if matching younger and middle-aged adults for triceps surae (TS) muscle strength and tendon stiffness eliminates age-related differences in muscular power production during drop jump. The maximal ankle plantar flexion moment and gastrocnemius medialis tendon stiffness of 29 middle-aged (40–67 years) and 26 younger (18–30 years) healthy physically active male adults were assessed during isometric voluntary ankle plantar flexion contractions using simultaneous dynamometry and ultrasonography. The elongation of the tendon during the loading phase was assessed by digitizing the myotendinous junction of the gastrocnemius medialis muscle. Eight younger (23 ± 3 years) and eight middle-aged (54 ± 7 years) adults from the larger subject pool were matched for TS muscle strength and tendon stiffness (plantar flexion moment young: 3.1 ± 0.4 Nm/kg; middle-aged: 3.2 ± 0.5 Nm/kg; tendon stiffness: 553 ± 97 vs. 572 ± 100 N/mm) and then performed series of drop jumps from different box heights (13, 23, 33, and 39 cm) onto a force plate (sampling frequency 1000 Hz). The matched young and middle-aged adults showed similar drop jump heights for all conditions (from lowest to highest box height: 18.0 ± 3.7 vs. 19.7 ± 4.8 cm; 22.6 ± 4.2 vs. 22.9 ± 4.9 cm; 24.8 ± 3.8 vs. 23.5 ± 4.9 cm; 25.2 ± 6.2 vs. 22.7 ± 5.0 cm). However, middle-aged adults showed longer ground contact times (on average 36%), lower vertical ground reaction forces (36%) and hence lower average mechanical power (from lowest to highest box height: 2266 ± 563 vs. 1498 ± 545 W; 3563 ± 774 vs. 2222 ± 320 W; 4360 ± 658 vs. 2475 ± 528 W; 5008 ± 919 vs. 3034 ± 435 W) independent of box height. Further, leg stiffness was lower (48%) in middle-aged compared to younger adults for all jumping conditions and we found significant correlations between average mechanical power and leg stiffness ($0.70 \leq r \leq 0.83$; $p < 0.01$). Thus, while jumping performance appears to be unaffected when leg extensor muscle strength and tendon stiffness are maintained, the reduced muscular power output during lower limb multi-joint tasks seen with aging may be due to age-related changes in motor task execution strategy rather than due to muscle weakness.

Keywords: leg stiffness, mechanical power, jumping, muscle strength, tendon stiffness, aging, motor control

INTRODUCTION

Lower limb muscular power production is crucial for an effective and safe locomotion in sport and during activities of daily living. For example, rapid muscle force generation is required during various daily lower limb multi-joint tasks and has been identified as a predictor for mobility (Rantanen and Avela, 1997; Suzuki et al., 2001; Bean et al., 2002; Cuoco et al., 2004) and falls risk (Skelton et al., 2002) in older adults.

Previous research suggests an age-related decline in muscular power production of the leg extensor muscles during single (e.g., maximal isokinetic ankle plantar flexion contractions; Thom et al., 2005, 2007) and multi-joint tasks such as walking (DeVita and Hortobagyi, 2000), running (Karamanidis et al., 2006; DeVita et al., 2016) and jumping (Ferretti et al., 1994; Wang, 2008) and that this decline has already begun by middle age (Kulmala et al., 2014). As joint mechanical power is the product of torque and angular velocity, one might suggest that this decrease in power generation capacity might be caused by age-related reductions in muscle strength (Asmussen and Heebøll-Nielsen, 1962; Viitasalo et al., 1985; Häkkinen and Häkkinen, 1991; Lindle et al., 1997) and muscle fiber maximal shortening velocity (Larsson et al., 1997; Korhonen et al., 2006). Furthermore, it is well established that the mechanical properties of the tendon affect the force potential of the muscle due to the force–length–velocity relationship (Hof et al., 2002; Roberts, 2016). For instance, higher patellar tendon stiffness has been shown to facilitate the rate of torque development during isometric knee extension contractions (Reeves et al., 2003; Bojsen-Møller et al., 2005). Therefore, the age-related decrease in leg extensor muscle-tendon unit (MTU) mechanical properties (i.e., isometric muscle strength and tendon stiffness; Karamanidis and Arampatzis, 2005, 2006; Onambele et al., 2006; Mademli and Arampatzis, 2008; Stenroth et al., 2012) may be a major contributor to the reduced muscular power production during lower limb multi-joint tasks seen with aging. Changes in lower limb muscular power production during running previously have been associated with an age-related deterioration in leg extensor isometric muscle strength and tendon stiffness (Karamanidis et al., 2006). However, it is yet not clear if the age-related differences in muscular power output during lower limb multi-joint tasks can solely be explained by changes in leg extensor muscle strength and tendon stiffness seen with aging.

A common approach for testing muscular power production of the leg extensor muscles during lower limb multi-joint tasks is maximal vertical jumping (Bojsen-Møller et al., 2005; Holsgaard Larsen et al., 2007; Caserotti et al., 2008; Edwén et al., 2014). Muscular power output during maximal vertical jumping (i.e., maximal counter-movement jumps and drop jumps; DJs) has previously been associated with the stiffness of the lower limb joints (referred to as ankle and knee joint stiffness or leg stiffness; Arampatzis et al., 2001a,b; Korff et al., 2009). When controlling for leg stiffness during maximal drop jumps (dropping from a box, and upon contact with the ground, executing a maximal vertical jump) by influencing ground contact times through verbal instructions, it has been shown that a maximization of mechanical power is achieved by optimal leg stiffness values (Arampatzis et al., 2001a,b). In an aging context, there is evidence that ankle joint stiffness during the braking phase of maximal DJs on a sledge apparatus is reduced with aging, causing a diminished jumping performance in older compared to younger adults (Hoffrén et al., 2007). However, as most studies comparing young and older adults included older adults with lower MTU mechanical properties, it remains unclear to what extent leg extensor isometric muscle strength and tendon stiffness or other neuromuscular factors (e.g., leg stiffness) contribute to the age-related changes in muscular power and performance output during lower limb multi-joint tasks. One might overcome this drawback by matching different age groups for one or more investigated parameters. For instance, analyzing younger and older adults with similar leg extensor muscle strength causes a reduction in the age-related differences in joint kinetics during gait, but these differences still remain significant, indicating that factors other than leg extensor muscle strength mediate the age-related changes in gait mechanics (Hortobágyi et al., 2016).

In the current study, we aimed to investigate if age-related differences in DJ performance (jumping height) and kinetics (maximal vertical ground reaction force, ground contact time, average mechanical power, and leg stiffness) would be eliminated when young and middle-aged adults are matched for triceps surae (TS) isometric muscle strength and tendon stiffness, in order to test the hypothesis that age-related differences in muscular power production during lower limb multi-joint tasks cannot solely be explained by alterations in leg-extensor muscle strength and tendon stiffness seen with aging. Therefore, the second aim of this study was to determine to what extent leg stiffness (referred to as the ratio of maximal vertical ground reaction force and the maximum vertical displacement of the body's center of mass during ground contact) is associated with the average mechanical power output during maximal vertical jumping in young and middle-aged adults with the hypothesis that leg stiffness may be a major factor of the age-related decline in muscular power production during lower limb multi-joint tasks.

As it has been shown that the mechanical properties of the TS MTU in particular are crucial for lower limb multi-joint tasks (e.g., walking, running, sprinting and jumping; Hof et al., 2002; Lichtwark and Wilson, 2007; Pandy and Andriacchi, 2010; Kulmala et al., 2014; Huang et al., 2015; Farris et al., 2016), the DJ task was chosen due to the specific requirements of TS muscle force generation. Further, by using different box heights we aimed to examine whether the outcome measures (and the potential age effects on these measures) vary with changes in task demand.

MATERIALS AND METHODS

Participants

Twenty-nine middle-aged (40–67 years) and 26 younger (18–30 years) physically active healthy male adults gave their written informed consent to participate in this study. Participants were included when two participants, one from the young and one from the middle-aged groups had similar TS MTU mechanical properties. Finally, eight middle-aged (range: 41–67 years) and eight younger (19–28 years) adults were matched for TS muscle strength and tendon stiffness. None of the participants had experience in regularly performing DJs. Exclusion criteria were any previous Achilles tendon ruptures, Achilles tendon injuries (e.g., tendinopathy) or pain within the last 12 months, or musculoskeletal impairments in the lower limbs (e.g., ankle joint pain), which could have influenced the outcomes of this investigation. The study was approved by the university's human ethics committee and was conducted according to the Declaration of Helsinki.

Analysis of Triceps Surae Muscle Strength and Tendon Stiffness

The experimental setup to assess TS MTU mechanical properties has been described previously in detail (Ackermans et al., 2016). Briefly, participants were seated on a custom-made strain gauge-type dynamometer (1000 Hz; TEMULAB, Protendon GmbH & Co. KG, Aachen, Germany) with their lower leg secured and the knee and ankle joints positioned at 90 degrees (foot and thigh perpendicular to the shank) and the foot placed on the dynamometer foot plate. The measurements began with a regimented warm-up of ten submaximal contractions guided by TEMULAB software and three maximal isometric contractions to precondition the tendon (Maganaris, 2003). Following this, the TS muscle strength and tendon force-elongation properties of the dominant leg were assessed during a maximal isometric voluntary ramp contraction (MVC) and three subsequent sustained contractions at 30, 50, and 80% of the determined maximal joint moment (Ackermans et al., 2016; McCrum et al., 2018b) by integrating dynamometry and ultrasonography (27 Hz; MyLab^{TM}One, Esaote, Genua, Italy). All sustained contractions were guided by visual feedback displayed on a computer screen showing the produced joint moment. In the current study, force values of ±5% of the target force were accepted (when the force was held for 3 s). If this was not achieved, the specific trial was repeated. During all maximal contractions, strong verbal encouragement was given to the participants in order to attain their actual MVC (McNair et al., 1996). By aligning the axis of rotation of the ankle and the force plate's center of rotation on the dynamometer, the ankle joint plantar flexion moment could be considered equal to the moment of the force plate (Ackermans et al., 2016). The gravitational forces were taken into account for all participants before each contraction. Achilles tendon force was calculated by dividing the resultant ankle joint plantar flexion moment by the tendon moment arm obtained from the literature (Maganaris et al., 1998). Please note that in the current model the magnitude of Achilles tendon force was estimated by calculating the resultant ankle joint moment via inverse dynamics without considering muscle activation and, therefore, we could not account for the moment contributions of all the other agonistic ankle plantar flexor muscles or antagonistic dorsiflexors. The gastrocnemius medialis (GM) tendon was examined using an ultrasound probe securely fixed on the GM within a rigid custom-made casing. Elongation of the GM tendon during the loading phase was determined by manually tracking the GM myotendinous junction in custom-made Matlab software (Matlab 2013b, MathWorks Inc., Natick, MA, United States). The effect of inevitable ankle joint angular rotation on the measured tendon elongation during contractions (Magnusson et al., 2001) was taken into account by multiplying the tendon moment arm by the ankle joint angular changes during contraction. In the present study, we used a potentiometer located under the heel, measuring any heel lift during loading and calculated the ankle joint angle changes via the inverse tangent of the ratio of the heel lift to the distance between the head of the fifth metatarsal bone and the potentiometer. This method was found to be in accordance with the results obtained using a motion capture system with an average difference in ankle joint angle changes of less than 1.1 degrees during maximal isometric plantar flexion contractions (Ackermans et al., 2016). Tendon stiffness was then subsequently determined as the ratio of the increase in the calculated tendon force and the increase in elongation from 30 to 80% of maximal tendon force. However, it is mandatory to address the fact that we did not account for the relative contribution of the GM to Achilles tendon force affecting our calculated GM tendon stiffness in absolute terms.

Analysis of Drop Jump Kinetics

On a second occasion, the TS muscle strength and stiffness-matched participants performed a series of three DJs from each different box height (13, 23, 33, and 39 cm; DJ13–DJ39, respectively) onto a force plate (90 cm × 60 cm, 1000 Hz; Kistler, Winterthur, Switzerland) in a randomized order. In order to avoid any effects through muscle fatigue, about 90 s of rest was provided between each trial. Since ground contact time influences leg stiffness during maximal vertical jumping (Arampatzis et al., 2001a,b), participants were instructed to jump as high as possible with as short contact time as possible. Prior to the measurements, all participants performed three familiarization sessions of the DJ task. Jumping height was determined using the flight time method (Asmussen and Bonde-Petersen, 1974) and ground contact time (defined as the time interval from the first instant when vertical ground reaction force reached a threshold value of 20 N to the first instant below 20 N) and maximal vertical ground reaction force during the support phase were determined. The average mechanical power was calculated by dividing the total work performed during the jump (summed work during the negative and the positive dynamic phase) by the ground contact time. Leg stiffness was assessed using the spring-mass model (Blickhan, 1989) and calculated as:

$$KLeg = \frac{F_{max}}{\Delta CoM}$$

where F_{max} is the maximal vertical ground reaction force and ΔCoM is the maximum vertical displacement of the body's center of mass during ground contact (Hobara et al., 2008). Vertical center of mass displacement was obtained by double integration of the vertical acceleration with respect to time:

$$CoM\,(t) = \iint \frac{F\,(t) - mg}{m} dt\,dt$$

where F represents the vertical ground reaction force, m is the body mass and g is the gravitational acceleration (Hobara et al., 2008). As lower limb stiffness has been shown to scale with body mass (Granata et al., 2002), leg stiffness was normalized to body mass and expressed as kN/m/kg.

Statistics

For all participants, the trial with the greatest jumping height for each jumping condition (box height) was taken into account for further analysis. A two-way repeated measures analysis of variance (ANOVA; factors: age group and box height) was performed in order to detect potential effects of age or box height on DJ height or kinetics with respect to average mechanical power, leg stiffness, maximal vertical ground reaction force and ground contact time. Detected significant main effects or interactions were further analyzed using Duncan's *post hoc* comparison. Anthropometric data, age, maximal isometric ankle joint plantar flexion moments and GM tendon stiffness were compared with t-tests for non-dependent samples. In order to determine the relationship between average mechanical power and leg stiffness during DJs from different box heights, Pearson's correlation coefficients were computed. All statistical analyses were performed using the software Statistica (Release 10.0, Statsoft; Tulsa, OK, United States). The level of statistical significance was set at $\alpha = 0.05$. All results provided in the text, tables and figures are presented as mean and standard deviation (SD).

RESULTS

The eight young and eight middle-aged adults were matched for TS isometric muscle strength and tendon stiffness (see **Figures 1**, **2**), which meant that there were no significant differences in maximal isometric ankle joint plantar flexion moments and GM tendon stiffness between young (3.1 ± 0.4 Nm/kg and 553 ± 97 N/mm, respectively) and middle-aged adults (3.2 ± 0.5 Nm/kg and 572 ± 100 N/mm). Paired samples t-tests revealed a significant group difference in age (23 ± 3 vs. 54 ± 7 years; $p < 0.001$), but no significant differences in body height (178.3 ± 6.5 cm vs. 179.9 ± 4.7 cm) and mass (79.3 ± 9.6 kg vs. 75.3 ± 4.9 kg).

Concerning the DJ performance, the implemented two-way ANOVA revealed a statistically significant age group × box height interaction for jumping height [$F(3,42) = 3.04$, $p < 0.05$]. Duncan's *post hoc* analysis revealed no differences in jumping height between TS muscle strength and stiffness-matched young and middle-aged adults independent of jumping condition (see **Figure 3**; Cohen's d over all box heights = $0.08 \leq d \leq 0.46$). Jumping height was significantly lower for the 13 cm compared to all other box heights in both age groups. However, a lower DJ performance for the 23 cm compared to the 39 cm condition was only found in younger adults.

Considering the DJ kinetics, there were significant age effects on ground contact time [$F(1,14) = 31.39$, $p < 0.001$; $2.04 \leq d \leq 3.30$] and maximal vertical ground reaction force during support phase [$F(1,14) = 26.35$, $p < 0.001$; $1.27 \leq d \leq 3.17$], with the middle-aged adults showing longer ground contact times and lower forces for all box heights (no interaction effect; see **Figure 4**). The two-way ANOVA revealed a statistically significant age × box height interaction for average mechanical power [$F(3,42) = 18.21$, $p < 0.001$]. Duncan's *post hoc* analysis revealed significant differences in average mechanical power between age groups independent of box height ($1.39 \leq d \leq 3.16$) and higher power values with increasing box height (see **Figure 4**). However, the middle-aged adults showed no significant difference in average mechanical power for the 23 and 33 cm conditions. Further, there was a statistically significant age group × box height interaction for leg stiffness [$F(3,42) = 4.76$, $p < 0.01$]. Duncan's *post hoc* analysis revealed lower leg stiffness values in the middle-aged compared to younger adults for all jumping conditions ($1.18 \leq d \leq 2.57$) and lower leg stiffness values with raising box height only in the younger adults (see **Figure 4**). However, no significant difference in leg stiffness was found between the 33 and 39 cm conditions.

Regarding a potential box height effect on DJ kinetics, there was a significant ($p < 0.001$) effect of jumping condition on maximal vertical ground reaction force during support phase [$F(3,42) = 17.23$], with higher vertical ground reaction forces with increments in box height. However, no significant difference in maximal vertical ground reaction force could be detected for the 33 and 39 cm conditions. Significant positive correlations were found between average mechanical power and leg stiffness over all analyzed participants ($n = 16$; $0.70 \leq r \leq 0.83$; $p < 0.01$; see **Figure 5**).

DISCUSSION

The first aim of this study was to investigate if age-related differences in DJ performance and muscular power generation would be eliminated when young and middle-aged adults are matched for TS isometric muscle strength and tendon stiffness in order to test the hypothesis that age-related differences in muscular power production during lower limb multi-joint tasks cannot solely be explained by alterations in leg-extensor muscle

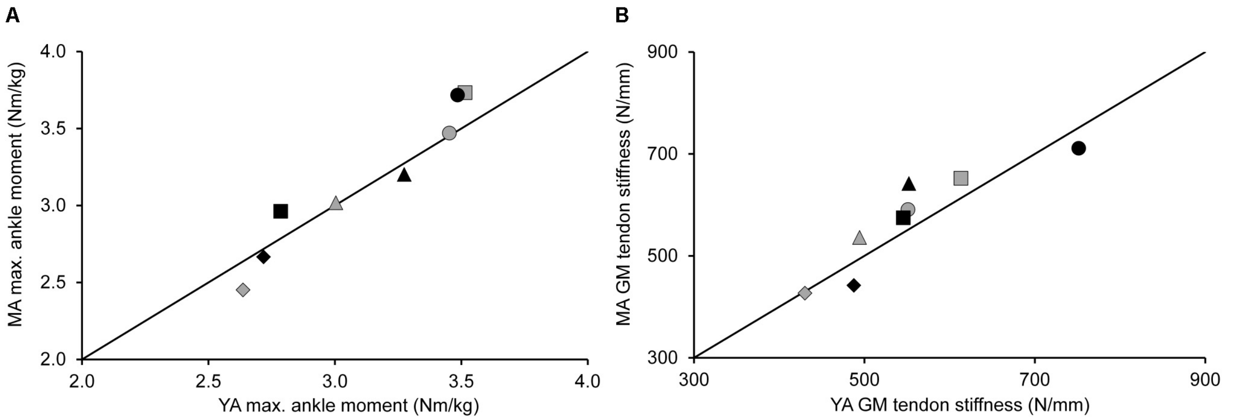

FIGURE 1 | Maximal ankle joint plantar flexion moment **(A)** and gastrocnemius medialis (GM) tendon stiffness **(B)** for the TS muscle strength and stiffness-matched young (YA; n = 8) and middle-aged (MA; n = 8) adults. Different symbols represent the participant pairs. The data points remain close to the identity line for both parameters, indicating that there were no statistically significant differences in maximal ankle joint plantar flexion moment and GM tendon stiffness between groups. Maximal ankle joint plantar flexion moment was normalized to body mass.

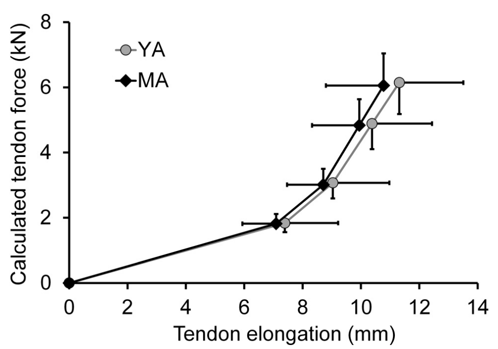

FIGURE 2 | Force-elongation relationship (mean and SD) of the gastrocnemius medialis tendon for the TS muscle strength and stiffness-matched young (YA; n = 8) and middle-aged (MA; n = 8) adults. Data points are displayed at 0, 30, 50, 80, and 100% of maximal calculated tendon force.

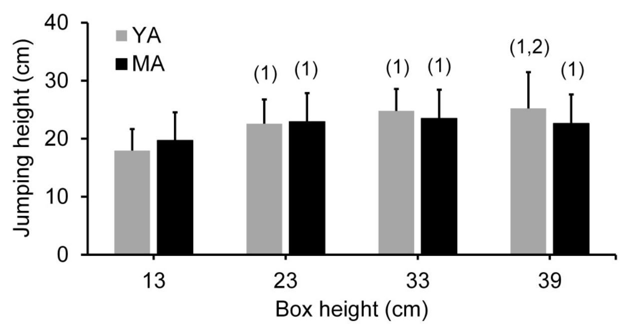

FIGURE 3 | Jumping performance during maximal drop jumps from four box heights in TS muscle strength and stiffness-matched young (YA; n = 8) and middle-aged (MA; n = 8) adults. There was no statistically significant age difference in drop jump height, independent of jumping condition. [1]Statistically significant (p < 0.05) difference to 13 cm; [2]Statistically significant (p < 0.05) difference to 23 cm.

strength and tendon stiffness seen with aging. In agreement with our hypothesis, middle-aged compared to younger adults generated lower average mechanical power during DJs despite being matched for TS isometric muscle strength and tendon stiffness. Our results are robust across four different jumping conditions i.e., task demands. Thus, they reflect previous findings seen in walking, showing that age-related differences in joint kinetics remained significant when matching younger and older adults for leg extensor muscle strength (Hortobágyi et al., 2016).

There might be potential factors other than maximum isometric muscle strength and tendon stiffness contributing to the age-related differences in leg extensor muscle power production during lower limb multi-joint tasks. Accordingly, the second aim of this study was to analyze whether a diminished muscular power production during maximal vertical jumping may be associated with potential differences in leg stiffness between age groups. In the current study, we found remarkably lower leg stiffness values in the middle-aged adults and significant positive correlations between leg stiffness and average mechanical power during support phase of maximal DJs from different box heights ($0.70 \le r \le 0.83$; $p < 0.01$), meaning that approximately 50 to 70% of the variability in average mechanical power during support phase can be related to the variance in leg stiffness within the pooled group of middle-aged and younger adults. These results support previous findings showing a significant relationship between muscular power output and the stiffness of the lower limb joints during maximal vertical jumping (Arampatzis et al., 2001a,b; Korff et al., 2009), indicating that leg stiffness is a major contributor to muscular power production during lower limb multi-joint tasks. Previously, aging has been associated with a less efficient utilization of tendon elasticity during maximal DJs on a sledge apparatus (Hoffrén et al., 2007). In the current study, TS MTU mechanical properties (maximal isometric muscle strength and tendon stiffness) were

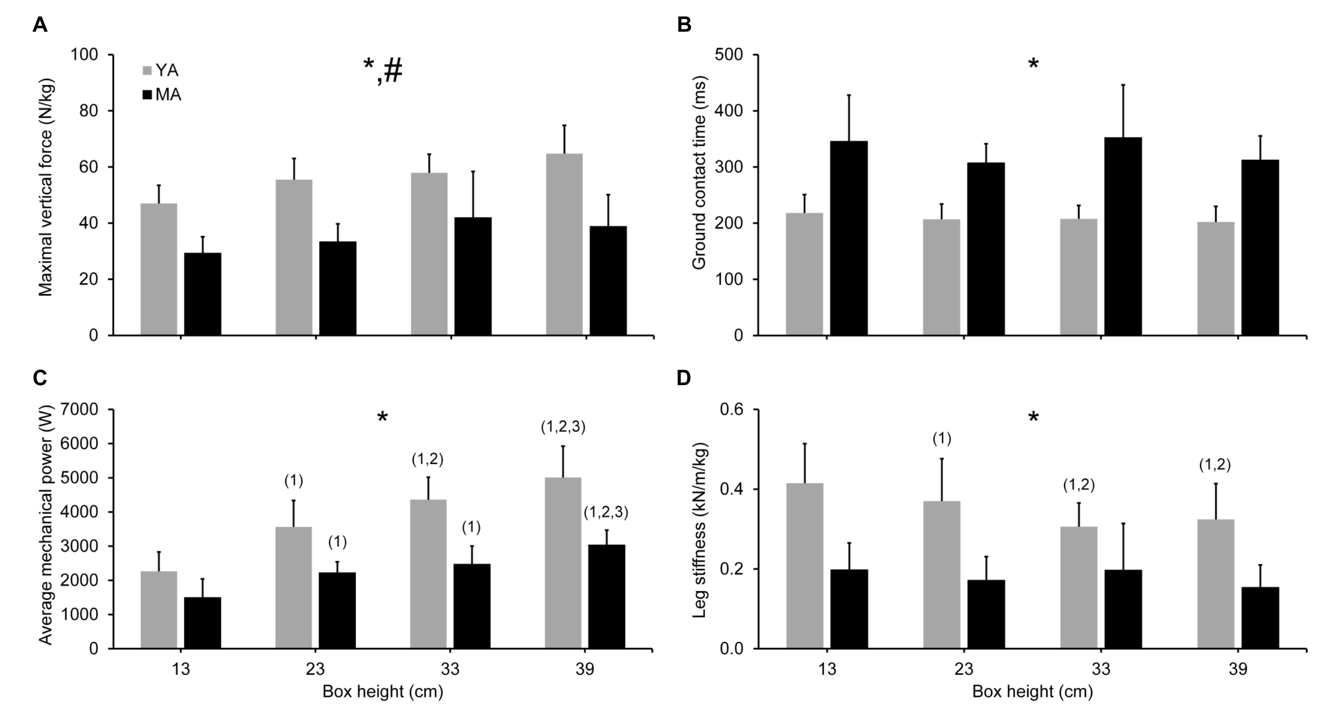

FIGURE 4 | Jumping kinetics during maximal drop jumps from four box heights in TS muscle strength and stiffness-matched young (YA; $n = 8$) and middle-aged (MA; $n = 8$) adults. There were statistically significant age differences in maximal vertical ground reaction force **(A)**, ground contact time **(B)**, average mechanical power **(C),** and leg stiffness **(D)**, independent of box height. Maximal vertical ground reaction force and leg stiffness were normalized to body mass. *Statistically significant ($p < 0.05$) difference between YA and MA; #statistically significant ($p < 0.001$) box height effect; [1]statistically significant ($p < 0.05$) difference to 13 cm; [2]statistically significant ($p < 0.05$) difference to 23 cm; [3]statistically significant ($p < 0.001$) difference to 33 cm.

matched between younger and middle-aged adults and hence, we found no significant age effect on GM tendon maximal energy storage capacity (i.e., integral under the GM tendon force-elongation-relationship during the loading phase of a MVC; young 215 ± 67 J; middle: 199 ± 80 J). Thus, one might speculate that the observed lower leg stiffness values in combination with the lower maximal vertical ground reaction forces in the middle-aged compared to younger adults may have led to a reduced amount of elastic strain energy stored in the GM tendon during the negative phase of the DJ potentially affecting power generation during the positive phase.

In the literature several factors have been associated with leg stiffness during lower limb multi-joint tasks. For example, leg stiffness is influenced by the stiffness of the passive elastic structures and the ability to appropriately activate the agonistic and antagonistic muscles (Hortobágyi and DeVita, 2000; Hoffrén et al., 2007) in order to stiffen the lower limb joints. During maximal DJs on a sledge apparatus, both a lower activation of the plantar flexor muscles and higher antagonistic coactivity during the braking phase have been related to a lower ankle joint stiffness experienced in older compared to younger adults (Hoffrén et al., 2007). Since TS isometric muscle strength and tendon stiffness were matched in the current study for middle-aged and younger adults, the lower leg stiffness in the middle-aged adults may be possibly explained by age-specific muscle activation patterns. However, with the current experimental set up (i.e., no electromyographic recordings of the lower limb muscles were taken) it cannot be directly answered whether possible age-related changes in muscle activation may be the primary drivers for the observed differences in leg stiffness during jumping between young and middle-aged adults or if other additional factors (e.g., differences in TS MTU parallel elastic structures) might also play a role.

Despite the above mentioned differences in muscular power production and leg stiffness between the two age groups, matching younger and middle-aged adults for TS isometric muscle strength and tendon stiffness resulted in no age-related differences in DJ height, independent of jumping condition (i.e., box height), indicating that leg-extensor isometric muscle strength and tendon stiffness play an important role for the performance in explosive motor tasks (e.g., DJ). The current finding, that a similar performance during maximal vertical jumping can be achieved by different motor task execution strategies i.e., levels of leg stiffness is in accordance to previous results (Arampatzis et al., 2001a,b). In the current study, middle-aged adults showed lower maximal vertical ground reaction forces and longer ground contact times compared to younger adults, suggesting that middle-aged adults maximize their jumping height by prolonging the time over which joint moments at the lower extremities are applied to the ground, due to a larger compliance of their lower limb joints during support phase. This is in line with earlier findings, showing lower maximal ground

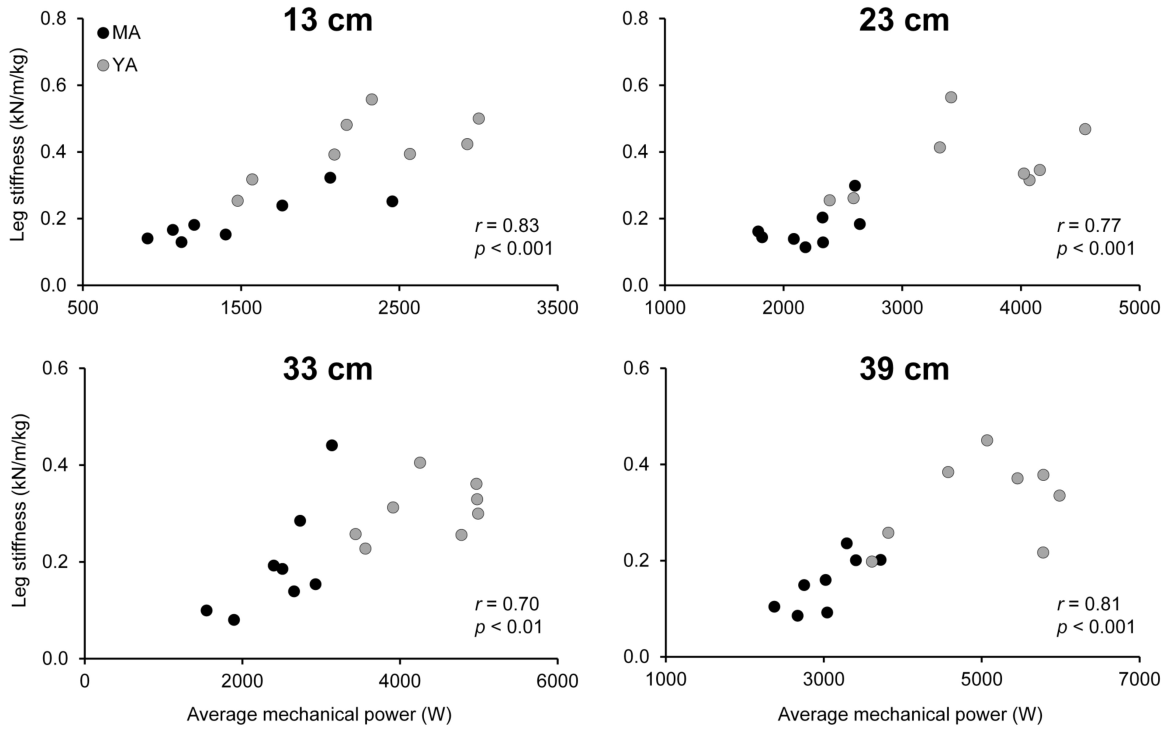

FIGURE 5 | Relationship between average mechanical power and leg stiffness during maximal drop jumps from four box heights in TS muscle strength and stiffness-matched young (YA; $n = 8$) and middle-aged (MA; $n = 8$) adults. Leg stiffness was normalized to body mass.

reaction forces but longer push-off phases in older compared to younger adults during hopping (Hoffrén et al., 2011) or DJs on a sledge apparatus (Hoffrén et al., 2007). Although not significant, there was a continuous increase in ground reaction force from 13 to 39 cm for the younger (with the exception of the 23 vs. 33 cm conditions) but not for the middle-aged adults (no differences between the 13 vs. 23, 23 vs. 39 and 33 vs. 39 cm conditions). These results suggest that middle-aged adults adopt a motor task execution strategy keeping maximal vertical ground reaction forces and hence, impact loads, on the musculoskeletal system within critical limits by less stiffening of their lower limb joints during the support phase compared to younger adults. This is supported by the reduced leg stiffness values with raising box height (i.e., task demand) even in younger adults. However, based on the current findings we cannot exclude that the middle-aged adults were not able to stiffen their lower limb joints to the same amount as younger adults, due to deficits in leg extensor muscle activation, for example (Stackhouse et al., 2001; Stevens et al., 2001; Morse et al., 2004; Clark et al., 2013).

A limitation of the current study may be that we did not use optical motion capture analysis but used a more convenient approach in order to assess mechanical power during maximal DJs. However, average mechanical power was calculated by the ratio of total work and ground contact time for all age groups and, therefore, we believe that this drawback potentially affects our results more in absolute terms than the validity of the corresponding comparative data. Regarding our ankle-knee joint configuration for GM tendon stiffness assessment, the contribution of the GM muscle to the net joint moment may be reduced compared to a fully extended knee joint angle since the GM operates on the ascending limb of the force-length relationship. However, in a previous study (Ackermans et al., 2016) we found no significant differences in tendon mechanical properties when using this configuration compared to a more dorsiflexed position (ankle joint at 85 degrees) that would increase the contribution of the gastrocnemii due to a rightward shift in the force-length relationship of the contractile elements. Further, our analysis of MTU mechanical properties does not account for potential age-related differences in knee extensor muscle strength and patellar tendon stiffness. One might argue that this may limit the validity of our findings, as next to the plantar flexors also the knee extensors contribute to leg stiffness and generated total muscular power during maximal DJs (Arampatzis et al., 2001a). However, since the age-related degeneration in muscle strength and tendon stiffness seems to be higher for the plantar flexors as for the knee extensors (see review McCrum et al., 2018a), we do not expect that this drawback significantly affects our main findings. Please note that in contrast to previously reported age ranges for middle-aged adults (typically 40–60 years; e.g., Süptitz et al., 2013; Saul et al., 2015) the current study included participants until the age of 67 years. However, the aim of the current study was basically to examine if matching younger and middle-aged adults for TS muscle strength and tendon stiffness eliminates age-related differences in muscular power production during DJ rather than age-related differences in neuromotor function *per se*. Despite the fact that seven of our middle-aged adults were between 41 to 59 years, we observed functionally relevant ($\geq$36%) differences for the main outcome parameters between the age groups, indicating that including a higher number of participants over the age of 59 years would have even strengthened the findings of the current study. Finally, a limitation of the current study is the relatively low number of participants in each group (n = 8 for the young; n = 8 for the middle-aged group) which reduces the potential for detecting statistical differences between age groups. Therefore we cannot exclude that a higher number of subjects might have led to significant differences in jumping height between young and middle-aged adults. However, this drawback has no effect on our observation that matching young and middle-aged adults for TS muscle strength and tendon stiffness does not eliminate age-related differences in mechanical power output during jumping.

CONCLUSION

In conclusion, independent of task demand, matching younger and middle-aged adults for TS isometric muscle strength and tendon stiffness eliminates age-related differences in jumping performance but not in mechanical power production during maximal DJs. Leg stiffness during the support phase was found to be significantly associated with muscular power production during maximal DJs, with lower leg stiffness values in middle- compared to younger-aged adults. Thus, while jumping performance appears to be unaffected when leg extensor muscle strength and tendon stiffness are maintained, the reduced muscular power output during lower limb multi-joint tasks seen with aging may be due to age-related changes in motor task execution strategy rather than due to muscle weakness.

DATA AVAILABILITY

The data supporting the conclusions of this manuscript will be made available by the authors, without undue reservation, to any qualified researcher.

ETHICS STATEMENT

This study was carried out in accordance with the recommendations of the German Sport university's human ethical commitee. The protocol was approved by the German Sport university's human ethical commitee. All subjects gave written informed consent in accordance with the Declaration of Helsinki.

AUTHOR CONTRIBUTIONS

KK, SH, and MK conceived the work. SH, TA, and MK acquired the data. MK drafted the manuscript. All authors contributed to analysis and interpretation of the work, prepared the figures, contributed to final approval of the version to be published, and agreed to be accountable for the work.

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frontiers in Physiology

SYSTEMATIC REVIEW
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Associations Between Types of Balance Performance in Healthy Individuals Across the Lifespan: A Systematic Review and Meta-Analysis

Rainer Kiss[1], *Simon Schedler*[2] and *Thomas Muehlbauer*[2]*

[1] *Department of Health and Social Affairs, FHM Bielefeld-University of Applied Sciences, Bielefeld, Germany,* [2] *Division of Movement and Training Sciences/Biomechanics of Sport, University of Duisburg-Essen, Essen, Germany*

Background: The objective of this systematic review and meta-analysis was to quantify and statistically compare correlations between types of balance performance in healthy individuals across the lifespan.

Methods: Literature search was performed in the electronic databases PubMed, Web of Science, and SPORTDiscus. Studies were included if they investigated healthy individuals aged ≥6 years and reported measures of static/dynamic steady-state, proactive, and/or reactive balance. The included studies were coded as follows: age group, gender, and balance type, test, parameter. Pearson's correlation coefficients were extracted, transformed (i.e., Fisher's z-transformed r_Z-value), aggregated (i.e., weighted mean r_Z-value), back-transformed to *r*-values, classified according to their magnitude, and statistically compared. The methodological quality of each study was assessed using the Appraisal tool for Cross-Sectional Studies.

Results: We detected twenty-six studies that examined associations between types of balance and exclusively found small-sized correlations, irrespective of the age group considered. More specifically, the weighted mean r_Z-values amounted to 0.61 (back-transformed *r*-value: 0.54) in old adults for the correlation of dynamic steady-state with proactive balance. For correlations between dynamic and static steady-state balance, the weighted mean r_Z-values amounted to 0.09 in children (*r*-value: 0.09) and to 0.32 in old adults (*r*-value: 0.31). Further, correlations of proactive with static steady-state balance revealed weighted mean r_Z-values of 0.24 (*r*-value: 0.24) in young adults and of 0.31 (*r*-value: 0.30) in old adults. Additionally, correlations between reactive and static steady-state balance yielded weighted mean r_Z-values of 0.21 (*r*-value: 0.21) in young adults and of 0.19 (*r*-value: 0.19) in old adults. Moreover, significantly different correlation coefficients ($z = 8.28$, $p < 0.001$) were only found for the association between dynamic and static steady-state balance in children ($r = 0.09$) compared to old adults ($r = 0.31$). Lastly, we detected trivial to considerable heterogeneity (i.e., $0\% \leq I^2 \leq 83\%$) between studies.

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Lars Donath,
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Emiliano Cè,
Università degli Studi di Milano, Italy
Katja Tomazin,
University of Ljubljana, Slovenia

***Correspondence:**
Thomas Muehlbauer
thomas.muehlbauer@uni-due.de

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Conclusions: Our systematic review and meta-analysis showed exclusively small-sized correlations between types of balance performance across the lifespan. This indicates that balance performance seems to be task-specific rather than a "general ability." Further, our results suggest that for assessment/training purposes a test battery/multiple exercises should be used that include static/dynamic steady-state, proactive, and reactive types of balance. Concerning the observed significant age differences, further research is needed to investigate whether they are truly existent or if they are caused by methodological inconsistencies.

Keywords: postural control, children, adolescents, adults, correlation

INTRODUCTION

In everyday life, adequate postural control is needed to safely manage activities of daily living (e.g., walking to school/college or climbing the stairs to one's office without sustaining a fall) and to regularly engage in sports-related activities (i.e., riding a bicycle or engaging in team sports). Thus, balance performance represents an important health- and activity-related component of everyday life that is also relevant across the human lifespan (Woollacott and Shumway-Cook, 1990; Granacher et al., 2011b). On the other hand, deficits in balance performance have been identified as important intrinsic factors increasing the risk of falling and sustaining an injury in children (Razmus et al., 2006), adolescents (Wang et al., 2006), young (Fousekis et al., 2011), and older adults (Rubenstein, 2006).

According to Shumway-Cook and Woollacott (2016) as well as, balance control involves static conditions in which the base of support (i.e., feet), and the ground remain stationary, as well as dynamic conditions in which both the base of support and the center of mass shift. Further, balance performance can further be subdivided into four types. These include, static steady-state balance (i.e., maintaining a steady position while sitting or standing), dynamic steady-state balance (i.e., maintaining a steady position while walking), proactive balance (i.e., anticipation of a predicted postural disturbance), and reactive balance (i.e., compensation of an unpredicted postural disturbance) (Shumway-Cook and Woollacott, 2016). This classification has widely been used in research on postural control in various populations (Bohannon, 2006; Springer et al., 2007), implying that these components represent different types of balance that are hardly associated and only show small-sized correlations among each other. On the other hand, balance performance has, particularly in textbooks (Fleishman, 1964; Schnabel et al., 2014; Meinel and Schnabel, 2018) used in college education, been introduced as a "general ability," suggesting that the various types of balance are highly interlinked. That is, a person with good static steady-state balance (e.g., less postural sway during one-legged stance) is supposed to also show superior performance in a dynamic steady-state balance task (e.g., fast gait speed during figure-eight walk). This implies large-sized correlations among the above-mentioned four types of balance because they are representatives of one balancing ability.

Additionally, the process of aging seems to have an effect on the association between types of balance in healthy individuals. In general, it has been reported that, depending on the parameter investigated, postural control shows a U-shaped trend for static (i.e., postural sway) or an inverted U-shaped course for dynamic (i.e., gait speed) steady-state balance performance across the lifespan (Granacher et al., 2011b). These age-related changes in balance performance are particularly based on the underlying neurophysiological structures responsible for postural control (Woollacott and Shumway-Cook, 1990). In children, the neuromuscular system is still developing due to maturation of the central nervous system (e.g., sensory integration) and has not reached its full functionality (Shumway-Cook and Woollacott, 1985; Woollacott and Shumway-Cook, 1994). In old adults, the neuromuscular system is in a state of functional decline and has lost its full capability due to, for example, a decline in the number of motor neurons and a diminished sensory feedback (Bouche et al., 1993; Terao et al., 1996; Maisonobe et al., 1997). These maturation-/aging-related limitations in postural control may contribute to age differences in the correlation between types of balance performance. Depending on the maturation-/aging-related limitation, one specific type of balance performance might be more affected than other components. Proof for this notation comes from studies that applied the Sensory Organization Test (SOT) to different age groups (Hirabayashi and Iwasaki, 1995; Peterson et al., 2006; Steindl et al., 2006). For example, Peterson et al. (2006) investigated the use of specific sensory information in maintaining postural stability in healthy children (6–12 years) and in adults (20–22 years). They found less postural stability in children compared to adults in test conditions using visual (i.e., normal vision, support sway-referenced) and vestibular (i.e., eyes closed, support surface sway-referenced) information. However, no age differences were detected in the condition using somatosensory information (i.e., eyes closed, fixed support). These results might indicate that age-related differences in the associations between balance dimension exist since adults performed equally-well in each test condition while children aged 6–12 years showed diverging performances.

Thus, the aim of this systematic literature review and meta-analysis was to quantify and statistically compare associations between types of balance performance in healthy individuals across the lifespan. The classification of balance performance in various types implies that balance is task-specific and thus small-sized correlations among types of balance are expected. Contrary, the use of balance performance in terms of a "general ability" suggests large-sized correlations among types of balance. Additionally, we assume age differences for the association between types of balance performance.

METHODS

Search of Literature

We performed a computerized systematic literature search in PubMed, Web of Science, and SPORTDiscus up to May 2018. The following Boolean search strategy was applied using the operators AND, OR, NOT: {[postural balance (MeSH) OR posture (MeSH)] AND (correlation study OR association OR relationship) NOT (patients OR disease)]}. With respect to the PubMed database, Medical Subject Headings (MeSH) were used as it was indicated before. The search was limited to: English language, human species, and to full text original articles. Further, we analyzed relevant review articles (Hrysomallis, 2007; Zemkova, 2014; Muehlbauer et al., 2015) in an effort to identify additional suitable studies for inclusion in the database.

Criteria for Selection

Studies were considered eligible to be included if they met the following criteria: (a) participants had to be healthy subjects, (b) participants were aged ≥6 years, and (c) outcomes from at least two types of balance had to be tested in the study. Studies were excluded if: (a) they investigated patients or people with diseases, (b) it was not possible to extract correlation coefficients from the results section or (c) authors did not reply to our inquiries sent by email. Based on the predefined inclusion and exclusion criteria, two independent reviewers (SS, TM) screened potentially relevant articles by analysing titles, abstracts, and full texts to determine their eligibility. If SS and TM did not reach a consensus concerning inclusion of a study, a third reviewer (RK) was contacted for clarification.

Study Coding

Each study was coded for the following variables: number of participants, sex, and chronological age. Further, we coded type, test, and parameter for the assessment of balance performance. With respect to the classification of postural control published by Shumway-Cook and Woollacott (2016), balance performance was separated into four types: static steady-state (i.e., maintenance of a steady position while standing), dynamic steady-state (i.e., maintenance of a steady position while walking), proactive (i.e., anticipating an expected postural disturbance), and reactive balance (i.e., compensating an unexpected postural disturbance). If several parameters were reported within one type of balance, the most representative measure was used for further analysis. In terms of dynamic steady-state balance, gait speed was used. With regards to static steady-state balance, center of pressure (CoP) displacement during one-legged stance was defined as the most relevant parameter. Concerning proactive balance, maximal reach distance in the Functional-Reach-Test was used. CoP displacements during perturbed one-legged stance was used as the most important outcome for reactive balance.

Quality Assessment and Statistical Analyses

The quality of all eligible studies was assessed using the Appraisal tool for Cross-Sectional Studies (Downes et al., 2016). This tool contains 20 questions that address the study design, the study quality, and the risk of bias. The questions were answered with either "yes," "no," or "do not know." There were seven questions (1, 4, 10, 11, 12, 16, and 18) related to the quality of reporting, 7 questions (2, 3, 5, 8, 17, 19, and 20) related to study design quality, and 6 questions (6, 7, 9, 13, 14, and 15) related to the possible introduction of biases in the study. Three questions (7, 13, and 14) asking for information on non-responders were not included in our analysis because the criterion was not applicable to the studies included in our review. Two independent reviewers (SS, TM) performed the quality assessments of the included studies. When any disagreement between the judges occurred, an additional rating was obtained from a third assessor (RK) to achieve a consensus.

Associations between types of balance were assessed using the Pearson product-moment correlation coefficient (r-value). r-values derived from different studies were pooled using "Fisher's z' transformation." In this regard, correlation coefficients were converted to the normally distributed variable z' (i.e., z-transformed r_z-value). The formula for this transformation is: $z' = 0.5[\ln(1+r) - \ln(1-r)]$ where ln is the natural logarithm (Kenny, 1987). In addition, the included studies were weighted according to the magnitude of the respective standard error (SE). The formula for the calculation of the SE is: $SE = 1/\sqrt{(N-3)}$ where N means the respective sample size (Kenny, 1987). Thereafter, we calculated the weighted mean r_z-values using Review Manager 5.3 software. For the classification and interpretation of correlation sizes, r_z-values were back-transformed to r-values. In accordance with the recommendation of Vincent (1995), values of $0 \leq r \leq 0.69$ indicate small, $0.70 \leq r \leq 0.89$ indicate medium, and $r \geq 0.90$ indicate large sizes of correlation. Lastly, we calculated the differences between the mean back-transformed r-values by age groups (Kenny, 1987; Preacher, 2002) using the following formula: $z = (z1 - z2)/\sqrt{(1/(n1-3) + 1/(n2-3))}$. Heterogeneity between studies was assessed using I^2 and Chi^2 statistics. Based on the recommendations of Deeks et al. (2008), values of $0\% \leq I^2 \leq 40\%$ indicate trivial, $30\% \leq I^2 \leq 60\%$ indicate moderate, $50\% \leq I^2 \leq 90\%$ indicate substantial, and $75\% \leq I^2 \leq 100\%$ shows considerable heterogeneity.

RESULTS

Study Characteristics

Figure 1 displays a flow chart that illustrates the different stages of the systematic literature search and the selection of studies over the course of the search. The initial search identified 3,024 articles that were potentially eligible for inclusion. After removal of duplicates and exclusion of ineligible articles, 21

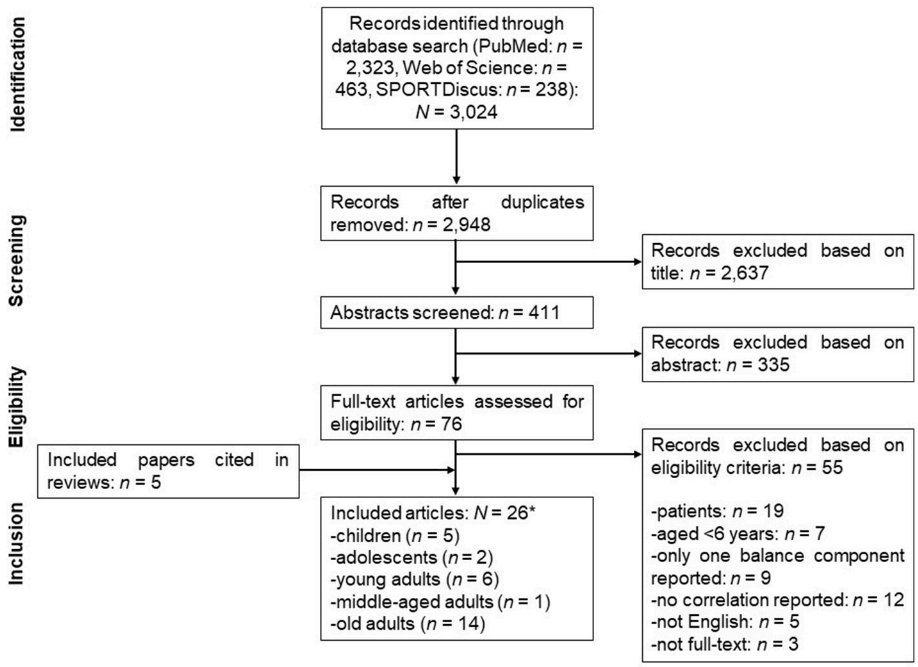

FIGURE 1 | Flow chart describing the systematic literature search.*Two studies investigated multiple age cohorts.

articles remained. We identified another 5 articles from the reference lists of already published review articles. Thus, 26 articles were included in the final analysis, whereas 2 of them (Shimada et al., 2003; Granacher et al., 2011a) investigated multiple age cohorts. **Table 1** illustrates the main characteristics of the included studies. Of the 26 articles, 4 studies investigated associations between types of balance in children ($n = 7{,}016$ subjects), 3 studies assessed adolescents ($n = 383$ subjects), 6 studies tested young adults ($n = 146$ subjects), 1 study used middle-aged adults ($n = 32$ subjects), and 14 studies examined old adults ($n = 1{,}756$ subjects). Irrespective of the age category, 4 studies reported correlations between dynamic steady-state and proactive balance, 2 studies between dynamic steady-state and reactive balance, 15 studies between dynamic and static steady-state balance, 2 studies between proactive and reactive balance, 9 studies between proactive and static steady-state balance, and 11 studies between reactive and static steady-state balance.

Quality of the Included Studies

Quality assessment revealed that the majority of studies included in our review met the criteria for (a) study design, (b) study quality, and (c) risk of bias above average. More specifically, twenty-five of the 26 included studies fulfilled $\geq$4 out of 7 criteria evaluating quality of study reports (**Table S1**, online supplement). Concerning quality of study design, $\geq$4 out of 7 criteria were fulfilled by 24 studies. Lastly, 20 studies fulfilled $\geq$2 out of 3 criteria with respect to risk of bias.

Correlations Between Dynamic Steady-State and Proactive Balance

Figure 2 illustrates the correlations of dynamic steady-state with proactive balance in old adults. The weighted mean r_z-value amounted to 0.61 and was accompanied with considerable heterogeneity ($I^2 = 83\%$, $Chi^2 = 11.72$, $df = 2$, $p = 0.003$). The back-transformed r-value of 0.54 indicated a small-sized correlation. Only one study (Muehlbauer et al., 2013a) reported a small correlation ($r_z = 0.26$, $r = 0.25$) between dynamic steady-state and proactive balance in children (**Table 1**). No study reported associations of dynamic steady-state with proactive balance in adolescents, young, and middle-aged adults.

Correlations Between Dynamic Steady-State and Reactive Balance

In children ($r_z = 0.22$, $r = 0.22$) (Muehlbauer et al., 2013a) and in old adults ($r_z = 0.03$, $r = 0.03$) (Muehlbauer et al., 2012a), only one study reported small-sized correlations between dynamic steady-state and reactive balance (**Table 1**). No study reported associations of dynamic steady-state with reactive balance in adolescents, young and middle-aged adults.

Correlations Between Dynamic and Static Steady-State Balance

Figure 3 displays the correlations of dynamic with static steady-state balance. Weighted mean r_z-values amounted to 0.09 in children ($I^2 = 33\%$, $Chi^2 = 2.99$, $df = 2$, $p = 0.22$) and to 0.32 in old adults ($I^2 = 80\%$, $Chi^2 = 44.69$, $df = 9$, $p < 0.001$) and were accompanied with moderate to considerable

TABLE 1 | Studies examining associations between types of balance by age group.

Reference	No. of subjects; sex; age, years (range or mean ± SD)	Balance type, test, parameter	z-transformed r_z-values, explained variance (r^2)
CHILDREN ($N = 4$)			
Drowatzky and Zuccato (1967)	50; F; 11–13	sSSB: sideward leap; bass stepping stone test; balance beam test dSSB: two-legged stork stance; diver's stand; stick test	sSSB-dSSB: 0.19, 4%
Granacher and Gollhofer, 2012	30; F (16), M (14); 6–7	sSSB: 20-s two-legged stance with eyes opened on a firmly fixed balance platform, CoP displacement length in ap-/ml-direction dSSB: 20-s two-legged stance with eyes opened on a free moving balance platform, CoP displacement length in ap-/ml-direction	sSSB-dSSB: 0.29, 8%
Humphriss et al., 2011	6,915; F (3,499), M (3,416); 10	sSSB: heel-to-toe stance on a beam, right/left foot forward, eyes open/closed, time; one-legged stance (right/left leg) with eyes open/closed, time dSSB: heel-to-toe beam walking, time	sSSB-dSSB: 0.03, 0%
Muehlbauer et al., 2013a	21; F (8), M (13); 7–10	sSSB: 30-s two-legged stance with eyes opened, CoP displacement length in ap-/ml-direction dSSB: 10-m walk, speed PB: TUG, time; FRT, distance RB: 10-s two-legged stance after perturbation with eyes opened, SO length in ap-/ml-direction	sSSB-PB: 0.41, 17% sSSB-RB: 0.31, 10% dSSB-PB: 0.26, 7% dSSB-RB: 0.22, 5%
ADOLESCENTS ($N = 3$)			
Granacher and Gollhofer, 2011	28; F (15), M (13); 16–17	sSSB: 30-s one-legged stance with eyes opened, CoP displacement length in ap-/ml-direction RB: 10-s one-legged stance after perturbation with eyes opened, SO length in ap-/ml-direction	sSSB-RB: 0.13, 2%
Ibrahim et al., 2011	330; F (165), M (165); 12–15	sSSB: 60-s one-legged stance with eyes opened/closed, time PB: 10-s jumps with feet together sideways, back and forth over a line	sSSB-PB: 0.33, 11%
Witkowski et al., 2014	25; M; 14–15	sSSB: Flamingo test, time dSSB: Marching test, points	sSSB-dSSB: 0.30, 9%
YOUNG ADULTS ($N = 6$)			
Granacher et al., 2011a	18; NR; 23 ± 3	sSSB: 30-s two-legged stance with eyes opened, total CoP displacement length dSSB: 10-m walk, coefficient of variation in stride velocity	sSSB-dSSB: 0.05, 0%
Hrysomallis et al., 2006	37; M; 23 ± 4	sSSB: 20-s one-legged stance with eyes opened, CoP displacement length in ml-direction PB: stepping balance task on an unstable surface, CoP displacement length in ml-direction	sSSB-PB: 0.35, 12%
Muehlbauer et al., 2013b	27; F (19), M (8); 23 ± 4	sSSB: 30-s one-legged stance with eyes opened, CoP displacement length in ap-/ml-direction RB: 10-s one-legged stance after perturbation with eyes opened, CoP displacement length in ap-/ml-direction	sSSB-RB: 0.20, 4%
Ringhof and Stein, 2018	24; F; 24 ± 1	sSSB: 30-s one-legged stance with eyes opened, total CoP displacement length PB: one-legged forward jump, time to stabilization RB: one-legged stance after perturbation with eyes opened, time to stabilization	sSSB-PB: 0.15, 2% sSSB-RB: 0.16, 3% PB-RB: 0.16, 3%
Sell, 2012	20; F (10), M (10); 23 ± 3	sSSB: 10-s one-legged stance with eyes opened/closed, SD of ground reaction force in ap-/ml-direction PB: forward/sideward hurdle jump, dynamic postural stability index	sSSB-PB: 0.13, 2%
Shimada et al., 2003	20; NR; 20–32	sSSB: SOT, score RB: perturbed walking on a treadmill, maximum anterior/posterior acceleration of the trunk	sSSB-RB: 0.27, 7%
MIDDLE-AGED ADULTS ($N = 1$)			
Muehlbauer et al., 2012b	32; F (9), M (23); 56 ± 4	sSSB: 30-s one-legged stance with eyes opened, CoP displacement length in ap-/ml-direction RB: 10-s one-legged stance after perturbation with eyes opened, CoP displacement length in ap-/ml-direction	sSSB-RB: 0.24, 6%
OLD ADULTS ($N = 14$)			
Callisaya et al., 2009	278; F (124), M (154); 60–86	sSSB: 30-s two-legged stance with eyes closed/opened, total CoP displacement length dSSB: 4.6-m walk, speed, cadence, step length/width	sSSB-dSSB: 0.18, 3%

(Continued)

TABLE 1 | Continued

Reference	No. of subjects; sex; age, years (range or mean ± SD)	Balance type, test, parameter	z-transformed r_z-values, explained variance (r^2)
Callisaya et al., 2010	410; NR; 72 ± 7	sSSB: two-legged stance with eyes closed/opened, postural sway dSSB: 4.6-m walk, speed, step length/width	sSSB-dSSB: 0.23, 5%
Carter et al., 2002	97; F; 69 ± 3	sSSB: SOT, equilibrium score dSSB: figure-eight walk, time	sSSB-dSSB: 0.68, 46%
Forte et al., 2014	57; F (33), M (24); 65–75	sSSB: 30-s Romberg test, total CoP displacement length, velocity; 30-s tandem stance test with eyes opened, total CoP displacement length, velocity dSSB: 10-m walk, gait speed	sSSB-dSSB: 0.03, 0%
Granacher et al., 2011a	18; NR; 74 ± 6	sSSB: 30-s two-legged stance with eyes opened, total CoP displacement length dSSB: 10-m walk, coefficient of variation in stride velocity	sSSB-dSSB: 0.63, 40%
Mackey and Robinovitch, 2005	25; F; 78 ± 7	sSSB: 15-s two-legged stance with eyes opened/closed, total CoP displacement length in ap-/ml-direction RB: balance recovery using a tether release protocol, maximum release angle	sSSB-RB: 0.29, 8%
Mayson et al., 2008	138; NR; 75 ± 7	sSSB: one-legged stance with eyes opened, time dSSB: DGI, score	sSSB-dSSB: 0.19, 4%
Melzer et al., 2009	43, F (27), M (16);78 ± 6	sSSB: 30-s two-legged stance with eyes opened, total CoP displacement area, length, velocity PB: LOS test, total CoP displacement length	sSSB-PB: 0.06, 0%
Miyazaki et al., 2013	124; M; 73 ± 7	sSSB: one-legged stance with eyes opened, time dSSB: 10-m walk, gait speed PB: TUG, time	sSSB-dSSB: 0.42, 18% sSSB-PB: 0.44, 19% dSSB-PB: 0.66, 44%
Muehlbauer et al., 2012a	24; F (13), M (11); 70 ± 5	sSSB: 30-s two-legged stance with eyes opened, CoP displacement length in ap-/ml-direction dSSB: 10-m walk, gait speed PB: TUG, time; FRT, distance RB: 10-s two-legged stance after perturbation with eyes opened, SO length in ap-/ml-direction	sSSB-dSSB: 0.28, 8% sSSB-PB: 0.17, 3% sSSB-RB: 0.04, 0% dSSB-PB: 0.10, 1% dSSB-RB: 0.03, 0%
Owings et al., 2000	79; F (50), M (29); 72 ± 5	sSSB: 20-s two-legged stance with eyes opened, CoP displacement length, speed in ap-/ml-direction PB: LOS test, total CoP displacement length RB: release from forward leaning, maximum recoverable angle; accelerated support surface; mechanically-induced trips	sSSB-PB: 0.18, 3% sSSB-RB: 0.08, 1% PB-RB: 0.14, 2%
Ringsberg et al., 1999	230; F; 75	sSSB: one-legged stance, time; 20-s two-legged stance with eyes opened/closed, total CoP displacement length dSSB: 30-m walk, time, cadence RB: 20-s two-legged stance with eyes opened on a moving platform, total CoP displacement length	sSSB-dSSB: 0.55, 30% sSSB-RB: 0.23, 5%
Shimada et al., 2003	20; NR; 65–79	sSSB: SOT, score RB: perturbed walking on a treadmill, maximum anterior/posterior acceleration of the trunk	sSSB-RB: 0.30, 9%
Shimada et al., 2011	213; F (130), M (83); 65–96	sSSB: 120-s one-legged stance with eyes opened, time dSSB: 6-m walk, time PB: TUG, time	sSSB-dSSB: 0.18, 3% sSSB-PB: 0.41, 17% dSSB-PB: 0.85, 72%

ap, anterior-posterior; CoP, center of pressure; DGI, dynamic gait index; dSSB, dynamic steady-state balance; F, female; FRT, Functional-Reach-Test; LOS, Limits-of-stability-Test; M, male; ml, medio-lateral; NR, not reported; PB, proactive balance; RB, reactive balance; sSSB, static steady-state balance; SD, standard deviation; SO, summed oscillations; SOT, sensory organization test; TUG, Timed-Up-and-Go-Test.

heterogeneity. Back-transformed r-values of 0.09 and 0.31 indicated small-sized correlations, respectively. In adolescents ($r_z = 0.30$, $r = 0.29$) (Witkowski et al., 2014) and in young adults ($r_z = 0.05$, $r = 0.05$) (Granacher et al., 2011a), only one study reported small-sized correlations between dynamic and static steady-state balance (**Table 1**). No study reported associations of dynamic with static steady-state balance in middle-aged adults.

Correlations Between Proactive and Reactive Balance

In young ($r_z = 0.16$, $r = 0.16$) (Ringhof and Stein, 2018) and in old adults ($r_z = 0.14$, $r = 0.14$) (Owings et al., 2000), only one study reported small-sized correlations between proactive and reactive balance (**Table 1**). No study reported associations of proactive with reactive balance in children, adolescents, and middle-aged adults.

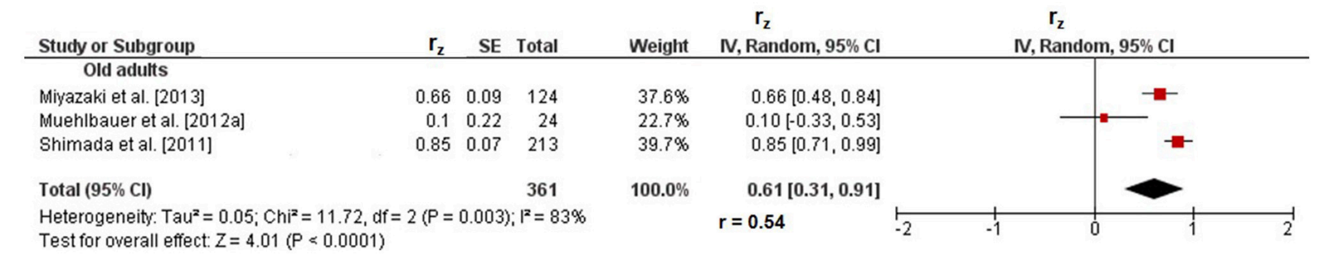

FIGURE 2 | Pearson's *r*-values (z-transformed) for correlations between dynamic steady-state and proactive balance in old adults. *CI* confidence interval, *df* degrees of freedom, *r* back-transformed Pearson's correlation coefficients, r_Z weighted z-transformed Pearson's correlation coefficients, *SE* standard error, *IV* inverse variance.

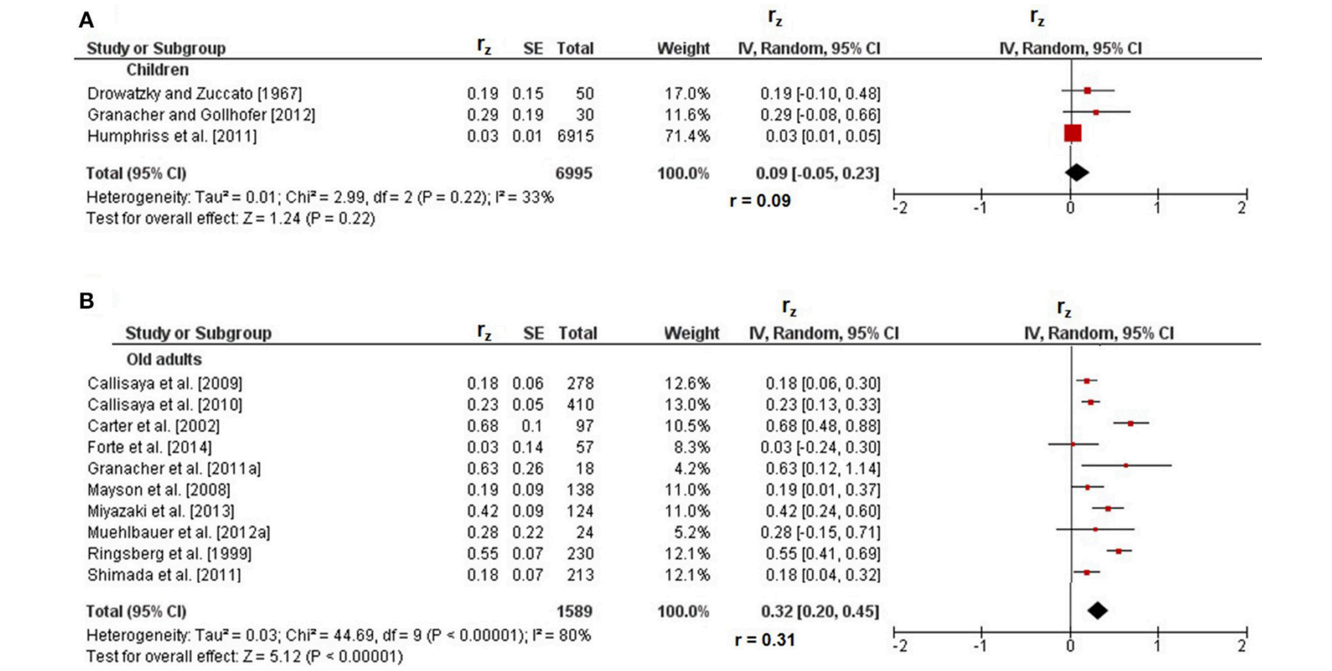

FIGURE 3 | Pearson's *r*-values (z-transformed) for correlations between dynamic and static steady-state balance in children **(A)** and old adults **(B)**. *CI,* confidence interval; *df,* degrees of freedom; *r,* back-transformed Pearson's correlation coefficients; r_Z, weighted z-transformed Pearson's correlation coefficients; *SE,* standard error; *IV,* inverse variance.

Correlations Between Proactive and Static Steady-State Balance

Figure 4 illustrates the correlations of proactive with static steady-state balance. Weighted mean r_z-values amounted to 0.24 in young adults ($I^2 = 0\%$, $Chi^2 = 0.80$, $df = 2$, $p = 0.67$) and to 0.31 in old adults ($I^2 = 59\%$, $Chi^2 = 7.40$, $df = 3$, $p = 0.06$) and were accompanied with trivial to substantial heterogeneity. The respective back-transformed *r*-values of 0.24 and 0.30 indicated small-sized correlations. In children ($r_z = 0.41$, $r = 0.39$) (Muehlbauer et al., 2013a) and in adolescents ($r_z = 0.33$, $r = 0.32$) (Ibrahim et al., 2013), only one study reported small-sized correlations between proactive and static steady-state balance (**Table 1**). No study reported associations of proactive with static steady-state balance in middle-aged adults.

Correlations Between Reactive and Static Steady-State Balance

Figure 5 displays the correlations of reactive with static steady-state balance. Weighted mean r_z-values amounted to 0.21 in young adults ($I^2 = 0\%$, $Chi^2 = 0.12$, $df = 2$, $p = 0.94$) and to 0.19 in old adults ($I^2 = 0\%$, $Chi^2 = 2.23$, $df = 4$, $p = 0.69$) and were accompanied with trivial heterogeneity. Back-transformed *r*-values of 0.21 and 0.19 indicated small-sized correlations. In children ($r_z = 0.31$, $r = 0.30$) (Muehlbauer et al., 2013a), adolescents ($r_z = 0.13$, $r = 0.13$) (Granacher and Gollhofer, 2011), and middle-aged adults ($r_z = 0.24$, $r = 0.24$) (Muehlbauer et al., 2012b), only one study reported small-sized correlations between reactive and static steady-state balance (**Table 1**).

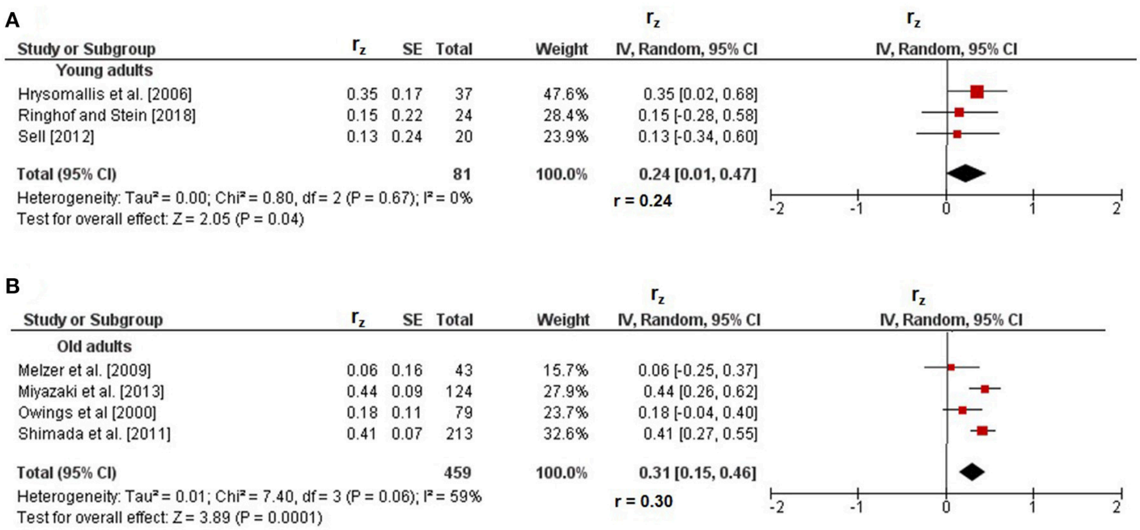

FIGURE 4 | Pearson's *r*-values (z-transformed) for correlations between proactive and static steady-state balance in young **(A)** and old adults **(B)**. *CI,* confidence interval; *df,* degrees of freedom; *r,* back-transformed Pearson's correlation coefficients; r_Z, weighted z-transformed Pearson's correlation coefficients; *SE*, standard error; *IV*, inverse variance.

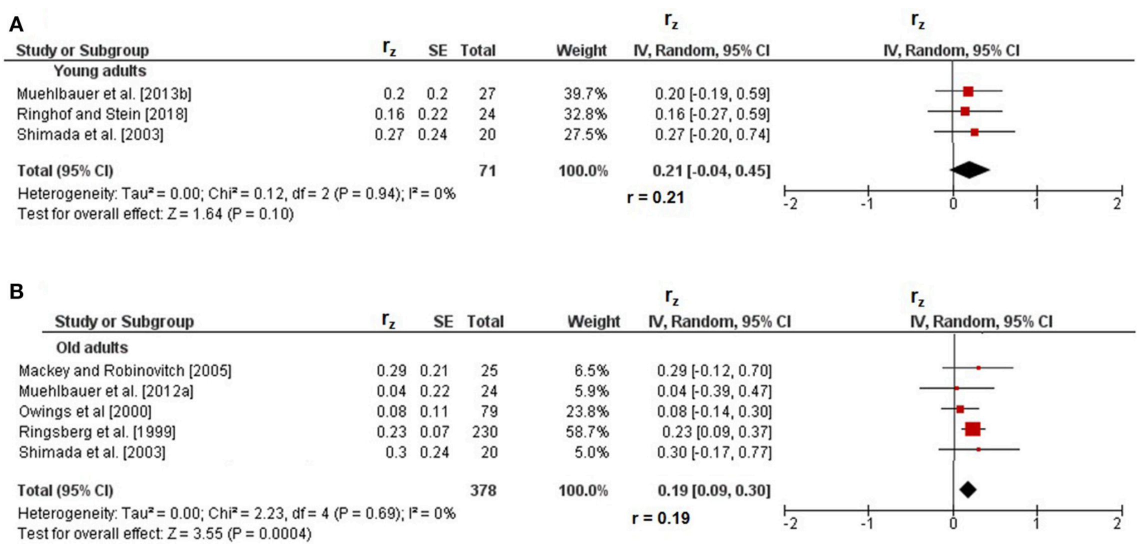

FIGURE 5 | Pearson's *r*-values (z-transformed) for correlations between reactive and static steady-state balance in young **(A)** and old adults **(B)**. *CI,* confidence interval; *df,* degrees of freedom; *r,* back-transformed Pearson's correlation coefficients; r_Z, weighted z-transformed Pearson's correlation coefficients; *SE*, standard error; *IV*, inverse variance.

Age Comparison of Correlations Between Types of Balance Performance

Statistically significant differences between age groups were obtained for the association of dynamic with static steady-state balance only. More precisely, the *r*-value in children ($r = 0.09$) was significantly smaller ($z = 8.28$, $p < 0.001$) than that in old adults ($r = 0.31$). Additional age comparisons of static steady-state balance with proactive ($z = 0.53$, $p = 0.60$) and reactive ($z = 0.16$, $p = 0.87$) balance did not reveal significant differences in young compared to old adults.

DISCUSSION

This systematic review and meta-analysis quantified and statistically compared associations between types of balance in healthy individuals across the lifespan. The main findings can be summarized as follows. First, we found exclusively small-sized correlations between types of balance in children, adolescents, young, middle-aged, and old adults. This finding was independent from the investigated type of balance (i.e., dynamic/static steady-state, proactive, and reactive balance).

Second, we detected significantly smaller correlations between dynamic and static steady-state balance in children compared to old adults. However, the analyses failed to detect further significant age differences for associations between other types of balance.

Associations Between Types of Balance Performance in Healthy Individuals Across the Lifespan

Our finding of exclusively small-sized correlations between types of balance contradicts the notion of balance as a "general ability," as indicated in textbooks (Fleishman, 1964; Schnabel et al., 2014; Meinel and Schnabel, 2018), and is in accordance with the presumption of Shumway-Cook and Woollacott (2016) who identified various types of balance performance (i.e., dynamic/static steady-state, proactive, and reactive balance). Based on the observation of small-sized correlations, it is suggested that types of balance performance are relatively independent and task-specific and thus, should be considered individually. For example, if a person shows a high amount of dynamic steady-state balance (e.g., fast gait speed in the 10-m walk test), an experimenter would not be able to predict how well that person would perform on a test of proactive balance (e.g., distance in the Functional-Reach-Test). Thus, if the goal is to assess balance performance, practitioners are not advised to only use one test, but rather utilize test batteries assessing different types of balance. Concerning the implication for training, the finding of low correlations indicates that programs including exercises for dynamic/static steady-state, proactive, and reactive types of balance should be applied if the goal is to enhance balance performance.

A possible reason for the observed small-sized correlations between types of balance could be differences in the specific task requirements. That is, during static steady-state balance tasks, such as one-legged standing, the base of support (i.e., foot) and the ground remain stationary as only the center of mass moves. However, during walking, as a representative of a dynamic steady-state balance task, the base of support and the center of mass shift, which provides different requirements to the involved neurophysiological structures than quiet standing. In this regard, Lau et al. (2014) investigated electrocortical activity using high-density electroencephalography during standing and walking on a treadmill in healthy young adults (age range: 20–31 years). They found that connections involving the sensorimotor cortex were significantly weaker during walking compared to standing. The authors interpreted this finding as a greater cortical involvement during standing than walking, because spinal neural networks play a larger role in the control of locomotion than stance. Further, the tested individuals might differentially experience balance task intensity and difficulty. For instance, normal walking or one-legged standing could be a low intensity balance task condition for young and middle-aged adults, but a high intensity task condition for children, adolescents, and/or old adults. Moreover, different mechanisms are involved for the control of proactive (e.g., distance in the Functional-Reach-Test) and reactive (e.g., postural sway during perturbed unipedal stance) balance (Riemann and Lephart, 2002). In the first case, feedforward control is necessary that involves the anticipation of a predicted postural disturbance during maximal forward leaning and the initiation of adequate muscle responses to prevent loss of balance. On the contrary, feedback control is characterized by the initiation of sufficient muscle responses after balance loss to compensate an unpredicted postural disturbance during one-legged standing and to avoid falling. In this respect, recent studies (Wälchli et al., 2017; Fujio et al., 2018) showed that the central nervous system differently prepares postural responses in expected compared to unexpected stance perturbations. For instance, Fujio et al. (2018) examined motor-evoked potential (MEP) induced by transcranial magnetic stimulation during expected (via acoustic signal) and unexpected (no signal) perturbations, while standing on a moveable platform in healthy young adults (mean age: 27 $\pm$ 2 years). As a result, the MEP for the tibialis anterior muscle was significantly enhanced under expected compared to the unexpected stance perturbation. Fujio and colleagues concluded that a prediction of an upcoming perturbation of standing balance modulates the excitability of corticospinal pathways. Additional physiological (i.e., fatigue) and psychological (i.e., attention, motivation) factors (Zech et al., 2012; Muehlbauer et al., 2013a) that are known to affect postural control might also have contributed to a larger or lesser amount while testing one compared to another type of balance and thus resulting in small-sized correlations. In summary, the observed small correlations between types of balance across lifespan are likely to reflect (i) differences in balance task complexity, difficulty, and/or intensity, (ii) discrepancies in the neurophysiological mechanisms involved in postural control, and (iii) the influence of additional physiological and psychological factors. Thus, the notion of balance as a "general ability" cannot be completely ruled out and further research is needed to examine whether these aspects masked the obtained correlations.

Age Comparison of Correlations Between Types of Balance Performance

Significant age differences were found for associations between dynamic and static steady-state balance in children compared to old adults. More specifically, the correlation coefficient was smaller in children compared to old adults. Based on this finding, one may argue that maturation/age have an effect on the association of selected types of balance. However, we could not detect further significant age differences in the relationship between other types of balance. Moreover, a closer look on the studies involved in the comparison that revealed significant age differences shows that fairly low correlation coefficients were reported in the study of Humphriss et al. (2011). For example, the association between dynamic (i.e., time for heel-to-toe beam walking) and static (i.e., time for the one-legged stance with eyes opened/closed) steady-state balance resulted in r-values of -0.0163 and -0.0531, respectively. When excluding the study by Humphriss et al. (2011) from our analysis, results indicated an increase of the back-transformed mean r-value from 0.09 to 0.23 and the formerly significant difference in associations between dynamic and static steady-state balance in children compared to old adults did not reach significance ($z = 0.74$, $p = 0.46$). Thus, methodological inconsistencies (i.e., no study directly compared several age cohorts using identical balance tests, parameters) between the involved studies could have also caused the significant age differences. As a consequence, further research is needed to investigate whether associations between types of balance are affected by maturation/age or methodological inconsistencies. To investigate whether the detected age differences between dynamic and static steady-state balance in children compared to old adults truly exist, it is recommended to conduct a series of single studies quantifying and statistically comparing correlations between various types of balance in children, adolescents, young, middle-aged, and old adults using identical tests and parameters.

CONCLUSIONS

The present systematic review and meta-analysis revealed exclusively small-sized correlations between types of balance performance in children, adolescents, young, middle-aged, and old adults. Findings indicate that balance performance seems to be task-specific rather than a "general ability." Thus, we advise practitioners to apply a test battery and not a single test for balance assessment. Further, multiple exercises including dynamic/static steady-state, proactive, and reactive types of balance should be used during balance training to target each balance dimension individually. In addition, we found significantly smaller correlation coefficients for the association of dynamic with static steady-state balance in children compared to old adults. This implies that maturation/age may have an effect on the association between selected types of balance. Yet, methodological inconsistencies (i.e., indirect age comparisons) between the involved studies could have also caused the significant age differences and thus further research is needed to investigate whether the observed age differences could be replicated.

AUTHOR CONTRIBUTIONS

RK worked on study design and manuscript preparation. SS assisted on data collection, data analysis, and worked on manuscript preparation. TM worked on study design, data collection, data analysis, and manuscript preparation.

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Do Older Adults Select Appropriate Motor Strategies in a Stepping-Down Paradigm?

Nick Kluft, Sjoerd M. Bruijn, Jaap H. van Dieën and Mirjam Pijnappels*

Department of Human Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam, Netherlands

Selecting motor strategies in daily life tasks requires a perception of the task requirements as well as of one's own physical abilities. Age-related cognitive and physical changes may affect these perceptions. This might entail that some older adults select inappropriate movement strategies when confronted with daily-life motor tasks, which could lead to balance loss or falls. We investigated whether older adults select motor strategies in accordance with their actual physical ability. Twenty-one older adults were subjected to a stepping down paradigm, in which full-body kinematics of selected and reactive behavior were recorded. Stepping down from a curb can be done with either (1) a relatively low effort but more balance threatening heel landing, or (2) a more controlled but more demanding toe landing. The probability of selecting a toe landing grows with an increase in curb height. We determined the curb height at which participants switched from heel to toe landing during expected stepping down over different heights as an indicator of their perceived ability. During an unexpected step down trial, participants encountered a step down of 0.1 m earlier than expected, because part of the walkway was removed and covered by a black cloth. We evaluated participants' actual physical ability from the reactive behavior, with performance defined as the reduction in kinetic energy between the peak value after landing and the onset of the next step. To unravel whether the selected motor strategies corresponded with actual physical ability, the ability to recover from the unexpected step down was correlated to the height at which the participants switched movement strategy. The switching height was not correlated to the ability to recover from an unexpected step down ($\rho = 0.034$, $p = 0.877$). This finding suggests that older adults do not select their movement strategy in stepping down based on their actual abilities, or have an imprecise perception of their actual abilities. Future research should evaluate whether inappropriate motor strategy selection in a stepping down paradigm can explain accidental falls in older adults.

Keywords: step descent, old age, degree of misjudgment, decision making, self-perception, falls, locomotion, perturbation

Edited by:
Oliver Faude,
Universität Basel, Switzerland

Reviewed by:
Eling D. de Bruin,
Karolinska Institutet (KI), Sweden
Masahiro Shinya,
Hiroshima University, Japan

****Correspondence:***
Mirjam Pijnappels
m.pijnappels@vu.nl

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INTRODUCTION

Moving through the environment requires integration of informational cues from the environment (Gibson, 1958). Combining these cues with the perception of one's physical abilities is essential for safe movement. However, 35 percent of the older adults experience a fall at least once a year (World Health Organization, 2007). Compared to young adults–with a fall incidence of 18 percent (Talbot et al., 2005)–older adults seem either vulnerable or reckless human beings, which puts into question older adults' ability to adapt their movement behavior to their actual physical abilities.

Appropriate perception of one's physical abilities would appear a necessity to avoid a mismatch between perceived and actual ability, but self-perception may be distorted in older adults, since even healthy aging is accompanied with a decline in cognitive capacities (Lustig et al., 2009; Segev-Jacubovski et al., 2011). Besides this cognitive decline, a decrease in physical abilities is observed as well (Vandervoort, 2002; Woollacott and Shumway-Cook, 2002), and continuous recalibration of perceived and actual abilities appears needed (Ellmers et al., 2018).

Two studies compared the actual ability to perform tasks to judgements of one's ability to perform these tasks (Butler et al., 2015; Kluft et al., 2016). When crossing narrow planks, almost one-third of the participating older adults showed risky behavior (Butler et al., 2015). Participants who chose a too narrow plank where more likely to fall in the upcoming year, showing the importance of such information for fall risk prediction models. In order to directly quantify the degree to which older adults misjudge their physical ability, we proposed a measure to evaluate the degree of misjudgment, and found that in most older adults perceived gait ability did not match their actual physical ability (Kluft et al., 2016). However, it is unclear whether this degree of misjudgment results in erroneous movement behavior.

A paradigm to investigate whether movement behavior is in alignment with movement ability, is motor strategy selection when stepping down a curb. It has been shown that there are two strategies of stepping down: a toe-landing strategy and a heel-landing strategy (Freedman and Kent, 1987; van Dieën et al., 2008; van Dieën and Pijnappels, 2009). At small curb heights, a heel landing is preferred, because toe landings are accompanied with higher effort (i.e., high ankle moment). When increasing the height of the curb, the probability of a toe landing increases, in older adults even more so than in their younger counterparts (van Dieën and Pijnappels, 2009). It has been suggested that a toe landing is preferred for higher curbs, as it allows for more controlled stepping down, since the kinetic energy generated during the step down stays within controllable limits (Buckley et al., 2008; van Dieën et al., 2008). Thus, stepping down a curb using the toe-landing strategy is thought to be safer than stepping down using a heel-landing strategy, at the cost of efficiency (i.e., higher joint moments and a loss of gait speed).

By relating the type of landing chosen to one's actual ability, we can determine whether the selected strategy is adequate. As strategy selection during anticipated stepping down entails a trade off between safety and efficiency, the actual ability measure should quantify the ability to be "safe" (minimizing balance threat) for fair comparison. This can be determined by the ability to regain balance after unexpected stepping down (van Dieën et al., 2007). Due to a sudden drop in walking surface, potential energy is quickly transformed to kinetic energy, and a large amount of kinetic energy should be dissipated to avoid balance loss (van Dieën et al., 2007). Another advantage of this paradigm is that the outcome is unlikely to be affected by one's perception of physical abilities, as there is no time for planning a motor strategy.

We aimed to investigate whether older adults select motor strategies in accordance with their actual physical ability, by comparing the selected strategy during an expected stepping down with the ability to recover from unexpected stepping down. The strategy selection during expected stepping down reflects participants' perceived ability, while kinetic energy reduction during the unexpected step reflects participants' actual ability. As we expect some individuals to select more and others to select less appropriate motor strategies, we hypothesized a moderate positive correlation between the ability to recover after an unexpected step down and the height at which subjects switched between heel landing and toe landing.

METHODS

Participants

Twenty-one healthy older adults participated in this study (for descriptives see **Table 1**); however, due to a technical error during data collection, data of one participant were excluded from further analysis. Participants were included if they reported no neurological or muscular impairments, were able to continuously walk for 10 minutes, had a mini-mental state examination (MMSE) of 25 or higher, and did not take medication, which could affect their gait stability. This study was carried out in accordance with the recommendations of "De Nederlandse Gedragscode Wetenschapsbeoefening," Association of Universities in the Netherlands. The protocol was approved by the "Vaste Commissie Wetenschap en Ethiek" (# VCWE 2016-129). All participants gave written informed consent in accordance with the Declaration of Helsinki.

Protocol

Participants were asked to walk over a 7 by 1.2 m platform, adopting the same speed as a set of light emitting diodes, which moved at a speed of 1.1 m per second alongside the platform at eye height of the participant (**Figure 1**). Participants were asked to step down at the edge of the platform while maintaining the indicated walking speed as much as possible. The height difference was adjustable and the participant was subjected to six different step heights (0.025, 0.05, 0.075, 0.10, 0.125, and 0.15 m). A 1 by 1 m custom-made force-plate was placed behind the height difference, so participants stepped down on top of this force plate (**Figure 1**). The participant first executed a 0.05 m step down; the landing strategies of six trials were registered and based on the resulting strategies, the platform was either lowered or raised to a new height. We continued varying the height until a height was reached where all six step downs were heel landings (lower bound) or toe landings (higher bound), and heights between the lower and upper bound were registered.

A multinomial logistic regression was fitted to the landing strategy data, and the height at which the chance that a toe landing was used equaled the chance that a heel landing was used (i.e., $P_{toe} = P_{heel} = 50\%$) was defined as the critical height (h_{crit}).

Subsequently, for the unexpected stepping down trials, participants were again instructed to step down at the middle of the platform, but instead stepped down earlier than they expected (**Figure 1**). The height difference was kept at 0.1 m and we informed the participants that they could experience unexpected stepping down during some of the next 15 trials. Three unexpected step downs were randomly assigned to one of these trials. The trial before the first unexpected step was considered as a normal walking condition. Behind the platform, a piece of black cloth spanned two bars, which were tensed by springs and kept in place by magnets, such that the cloth seemed to be walkable and part of the platform. When one of the foot markers crossed the actual height difference, it triggered the magnets to switch off, causing the cloth to quickly drop 0.1 m. This manipulation ensured a heel landing in the unexpected trials. In the other trials, the cloth covered a solid wooden platform of 0.1 m height. During all walking conditions, participants wore a safety harness attached by ropes to a railway mounted to the ceiling, to assure that the participant would not hit the floor if a fall occurred.

TABLE 1 | Participant descriptives.

Descriptives:			
Age	71	[7.25]	years
Females	7	(33%)	persons
Medication (≥ 4 different medicine)	3	[14%]	persons
Self-reported physical activity	1240	[309]	mins./week
Fallers (≥ 2 falls in the past year)	7	(33%)	persons
Falls in the past year	1	[1]	falls
MMSE	29	[2]	points
FES-I	18	[3.25]	points
Body weight	68.7	[12.3]	kg
Body height	1.69	[0.12]	m
Grip strength	284	[96.9]	N
Max. knee-extension torque	79	[4.6]	Nm

Prevalence with percentage of sample, or median values and interquartile range (IQR), the latter is indicated by the square brackets (i.e., median [IQR]).

Data Acquisition and Analyses

The positions of 12 infrared light emitting clusters of three markers were captured by three camera arrays (OptoTrak, Northern Digital Inc., Ontario, Canada), to measure full-body kinematics. A kinematic model with 12 linked segments was fitted to the kinematic data, resulting in kinematic trajectories without missing data (van den Bogert et al., 2013). The segments orientations of the most distal segments of which cluster markers were not visible during data collection for less than 30 consecutive samples, were interpolated using a spherical linear interpolation (Dam et al., 1998). The total kinetic energy (i.e., rotational and translational kinetic energy of all segments) and the mechanical work of the leading-leg joints were calculated for (I) expected and (II) unexpected stepping down and for (III) normal walking conditions (the last trial prior to the unexpected stepping down). The kinetic energy during the unexpected stepping down was time normalized from the mid stance before stepping down to the leading-leg landing, and from this event again to first trailing-leg step after landing. Gait events were determined using the kinematic data and afterwards visually checked to ensure correct timing of these events. The peak in the time-normalized kinetic energy was identified and the kinetic energy reduction after the occurrence of this peak and before the trailing foot landing was determined to represent the ability to recover from an unexpected step down (see **Figure 2** for illustration). Additionally, the body's angular momentum was calculated (reported in the **Supplementary Material**) as an alternative measure for safety in terms of balance control.

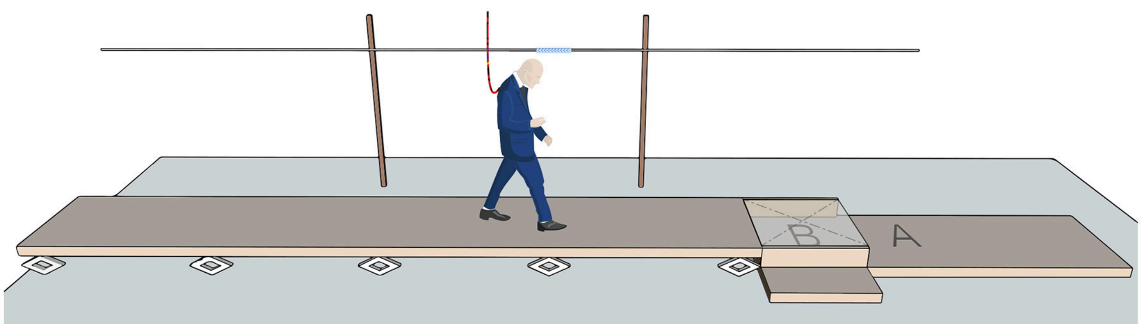

FIGURE 1 | Illustration of the experimental set up. Participants were asked to step down at the end of the platform (landing position indicated by A). The step height was adjustable with jacks from 0.025 to 0.15 m. During the unexpected stepping down condition participants stepped down at position B, while they believed that they would step down, as instructed, at position A, with a height difference of 0.1 m. Participants were secured by a safety harness fixed with climbing ropes to the ceiling. We instructed the participant to adopt the walking speed of 1.1 m/s, as indicated by a LED strip alongside the platform.

Paradigm Validity

The validity of our paradigm was evaluated by examining the kinetic energy and mechanical work based on three criteria. First, we expect no difference in kinetic energy between normal walking and unexpected stepping down before the expected heel contact (i.e., the position where the participant believed they would land). Second, similar to the findings of van Dieën et al. (2008), we expected a higher reduction in normalized kinetic energy and higher negative mechanical work done by the ankle during toe landings as compared to heel landings. Only the heel and toe landings that were observed within one step height in the expected condition were analyzed for this within-subject comparison. This comparison was based on a subset of the sample ($N = 18$), as two participants were very consistent in their strategy selection and did not switch strategies within any given step height. Finally, a higher kinetic energy was expected during unexpected stepping down compared to expected stepping down. Since unexpected stepping down always resulted in a heel landing, only heel-landing strategies during the expected stepping from a height of 0.1 m were selected for this comparison. It should be noted that not all participants performed a heel-landing at the 0.1 m step height, hence this comparison was made only based on a subset of the sample ($N = 11$).

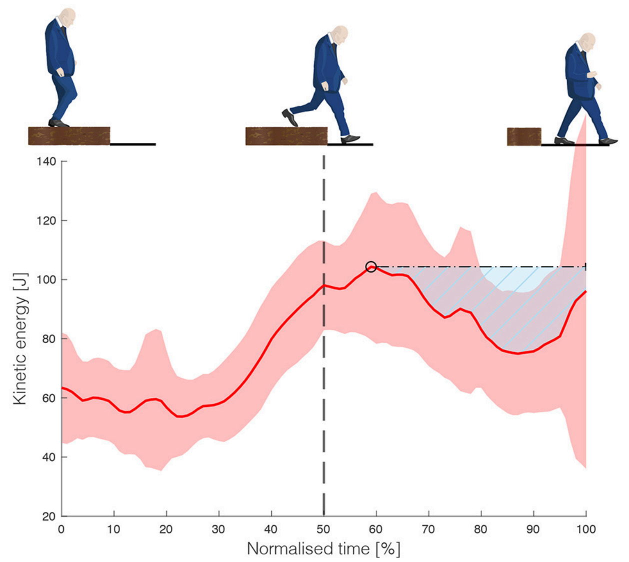

FIGURE 2 | Kinetic energy, time normalized from mid stance, via leading leg landing to trailing leg landing. The ability to recover from unexpected stepping down is defined by the area above the curve (marked by the masked area in the figure) between the peak in kinetic energy after landing and the trailing foot landing.

Statistical Analysis

Statistical parametric mapping (SPM, Friston et al., 2007; Pataky et al., 2013) was used to test the three assumptions for construct validity of the paradigm. SPM two-tailed paired t-tests were used to identify differences between conditions, and resulting in the weighted magnitude of the differences [referred to as SPM(t)] for the entire time series. To test the null hypothesis, the smoothness of the time series was estimated and a threshold was calculated using Random Field Theory (Adler and Taylor,

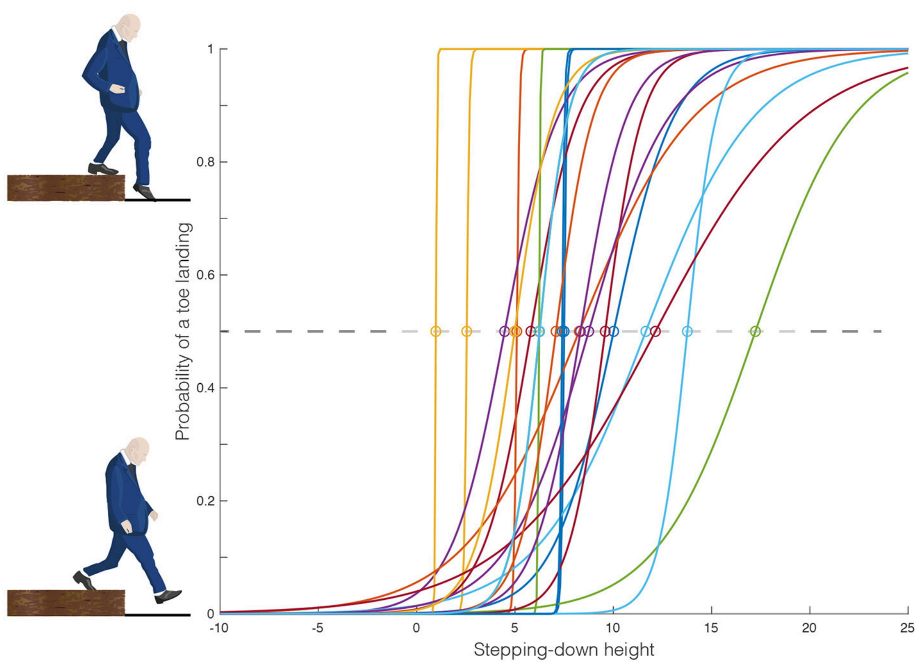

FIGURE 3 | Each participant's stepping-down behavior, determined as the probability of a toe landing by the step height using logistic regression. The individual lines depicts the logistic fits, and the critical switching height was defined as the height at which the probability of a toe landing equaled 0.5.

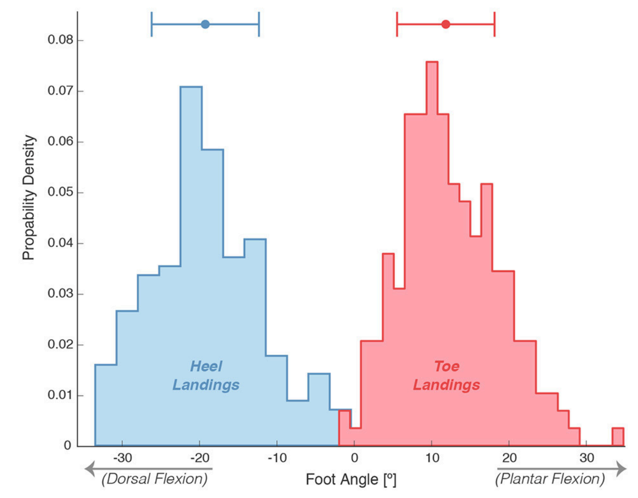

FIGURE 4 | Distributions of foot angles at first foot contact during stepping down. Positive angles are indicative for a plantar flexion orientation of the foot, where negative values represent a more dorsal flexion orientation.

2007) and α = 5%, implying that 5% of random but equally smooth curves would exceed this threshold. The null hypothesis is rejected at the instances the SPM(t) value exceeds the threshold. Further technical notes about this procedure can be found elsewhere (Friston et al., 1994). SPM analyses were performed using the spm1d package[1]. Next, the association between h_{crit} as the perceived stepping down ability and the actual ability to recover from an unexpected step down was evaluated using linear regression. Data analyses were performed using custom-made Matlab software (The Mathworks Inc., Natick, MA, RRID:SCR_001622).

RESULTS

Six expected condition trials out of a total of 120 trials needed to be excluded from further kinematic analyses due to unforeseen technical errors in the kinematics.

The variability in the participants' critical height, h_{crit}, as determined by logistic regression is presented in **Figure 3**. The logistic regression was fitted using binomial data, **Figure 4** displays the foot angle at foot contact during the step down. This data was shaped as a bimodal distribution, which confirms that stepping-down strategy selection is a binary process.

One participant performed toe landings in every trial. During normal walking, this participant performed heel landings, so we added an additional height of 0 cm and assumed that for this individual those were all heel landings, this led to an h_{crit} of 0.97 cm. The first assumption for validity assessment was met, as the test statistic [SPM(t)] did not exceed the threshold of 3.453 (see **Figure 5**), confirming that kinetic energy did not differ between unexpected stepping down and normal stepping prior to (expected) landing.

The second assumption was also met, as more kinetic energy was absorbed during toe landings compared to heel landings (see **Figure 6**). The SPM(t) exceeded the critical threshold of 3.767 ($p < 0.001$) shortly before landing at the lower platform (at 45% of the normalized time, or 0.08 s before landing), until shortly before the next step (at 93% of the normalized time, or at on average 0.70 s after leading leg landing). The mechanical work done by the ankle was larger after toe landing than after heel landing, as SPM(t) > .389 ($p < .001$) within the first 0.34 s after foot contact (**Figure 7**).

The third assumption that in unexpected stepping down less kinetic energy is absorbed than in expected stepping down, was also confirmed (**Figure 8**). In the first 0.2 s after foot contact, the test statistic SPM(t) exceeds the computed threshold of 4.321. Thus, the perturbation effectively increased the kinetic energy in the system after a sudden drop in walking surface, indicating an increase in balance threat. Finally, A non-significant association ($\rho = 0.034$, $p = 0.877$) was found between h_{crit} ($M = 7.82$ cm, $SD = 3.91$ cm) and the kinetic energy absorbed ($M = 1228.84$ J·%, $SD = 568.87$ J·%) during unexpected stepping down (**Figure 9**).

DISCUSSION

The objective of this experiment was to evaluate whether older adults select their movement strategies in line with their physical ability. We expected a moderate positive relation between the switching height and the ability to reduce kinetic energy in (unexpected) stepping down; however, a poor and non-significant association was observed. We offer two arguments that might explain this result.

Validity of the Actual Ability

The weak association could be attributed to the kinetic energy measure poorly reflecting one's actual ability. In this study, we assumed that the strategy selection was made based on a trade off between safety (i.e., control of kinetic energy) and efficiency (lower joint moments and maintenance of gait speed). The data showed that, compared to toe landings, the kinetic energy remained larger after heel landing until the onset of the next step of the trailing leg. This result can be seen to confirm our assumed trade off: a too high kinetic energy would be dangerous as it may be indicative of falling, but reducing kinetic energy too much after a step down (as during the toe-landing) may be inefficient, as a certain kinetic energy is required to walk at a given speed.

Alternatively to kinetic energy as an indicator of this threat, angular momentum could possibly be more directly linked to the balance threat imposed (c.f., Pijnappels et al., 2004). However, the gain in sagittal-plane angular momentum during unexpected stepping down appeared only limited (van Dieën et al., 2007, see the **Supplementary Material** for the evaluation of angular momenta during stepping down in the present data set).

Furthermore, the unexpected perturbation was triggered when a foot marker crossed the curb position. This involved that the timing of the trigger to drop the cloth was not relative to the gait cycle, and hence differed between participants. As the planning of stepping down occurs prior to toe-off of the leading leg–the planning is barely adjusted after toe-off (Timmis

[1] www.spm1d.org

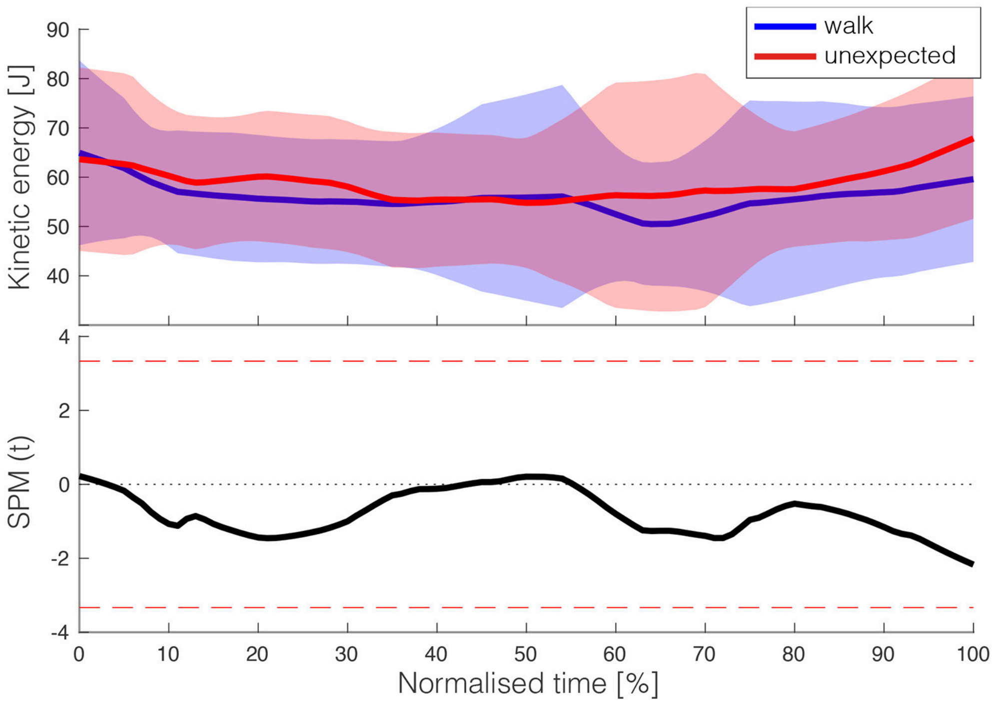

FIGURE 5 | (Top) depicts kinetic energy between mid stance and expected foot landing for normal walking and unexpected landing. For normal walking, the expected foot landing coincides with actual foot landing, while in the unexpected condition, it was the instant at which the vertical position of the leading foot crossed the platform level, where participants believed it would touch the ground at this moment in time, while actually this happened a fraction later. **(Bottom)** shows the SPM(t) results, indicating no significant differences between the two conditions.

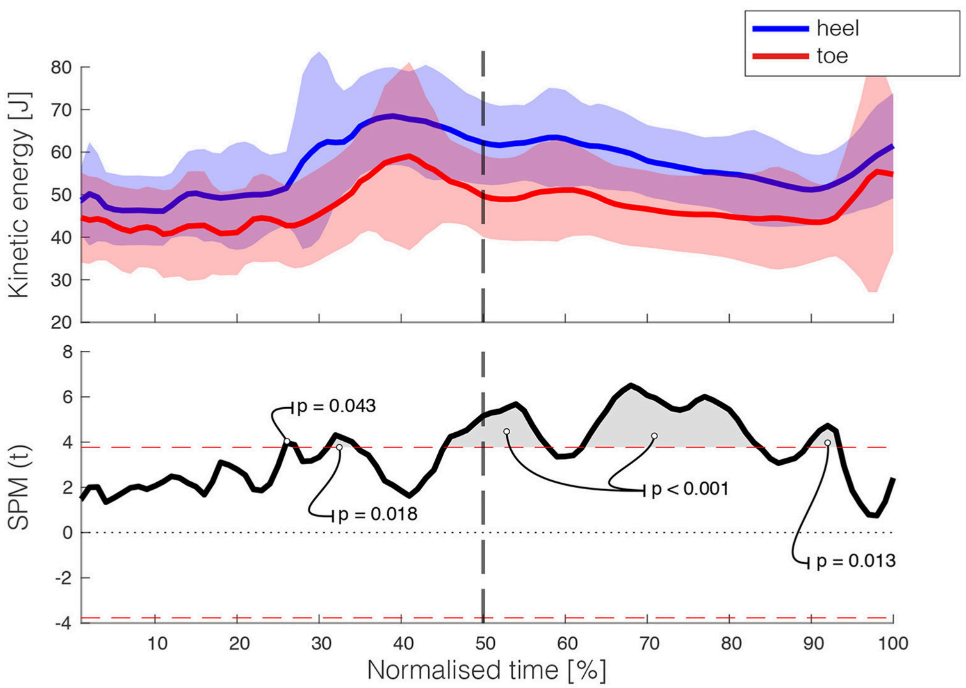

FIGURE 6 | (Top) depicts the magnitude of the kinetic energy vector of both the heel and toe landings (± SD is displayed in the gray area). The SPM(t) is shown in the **(Bottom)**, with the gray areas indicating where the difference between the two conditions was significant. The heel landing occurred at 50% of the normalized time (vertical dashed line).

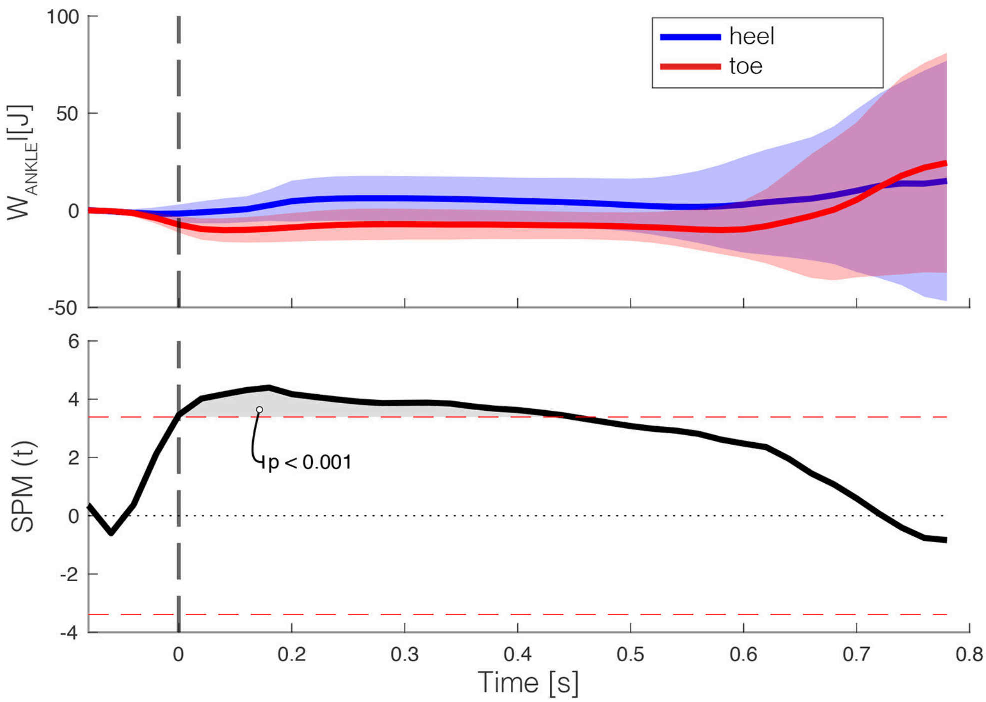

FIGURE 7 | Ankle joint mechanical work in the leading limb **(Top)** averaged over participants during stepping down using a toe-landing strategy (red) and heel-landing strategy (blue). Error bars display ± 1 standard deviation. The SPM(t) is shown in the **(Bottom)**, with the gray areas indicating where the difference between the two conditions was significant. The time series were aligned on foot landing (vertical dashed line).

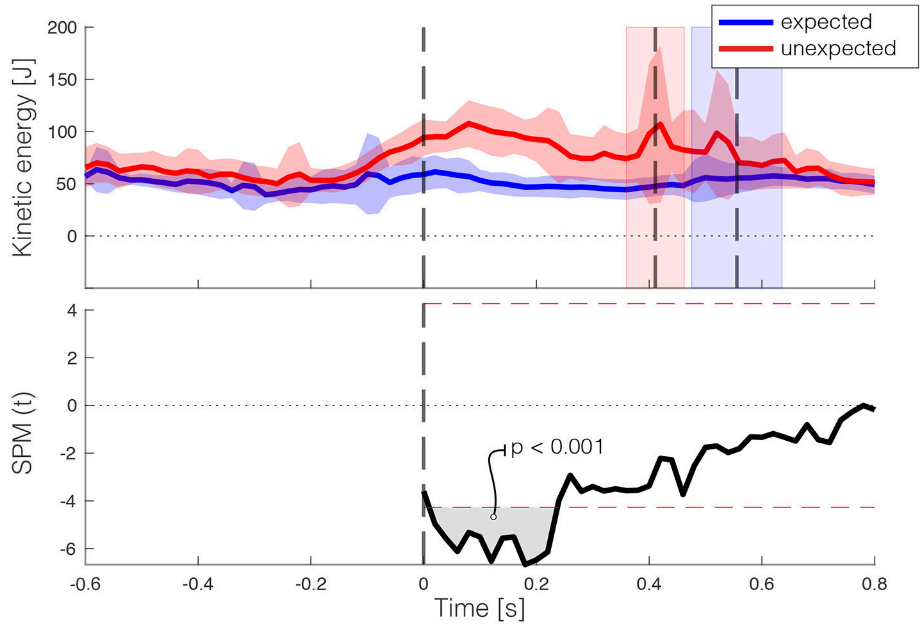

FIGURE 8 | Kinetic energy during step down under two conditions **(Top)**: unexpected stepping down (red) and expected stepping down (blue). The time series of the two conditions were aligned on first foot contact on the lower platform. Note that for fair comparison of expected stepping only heel-landing strategies were considered, as only heel-landing strategies can occur in the unexpected stepping down trial. The vertical line at time 0 displays the moment of leading-foot landing, the trailing-foot landings for both conditions are indicated by the second (unexpected; red) and third (expected; blue) vertical lines. Error bars display ± 1 standard deviation. The SPM(t) is shown in the **(Bottom)**, with the gray areas indicting where the difference between the two conditions was significant.

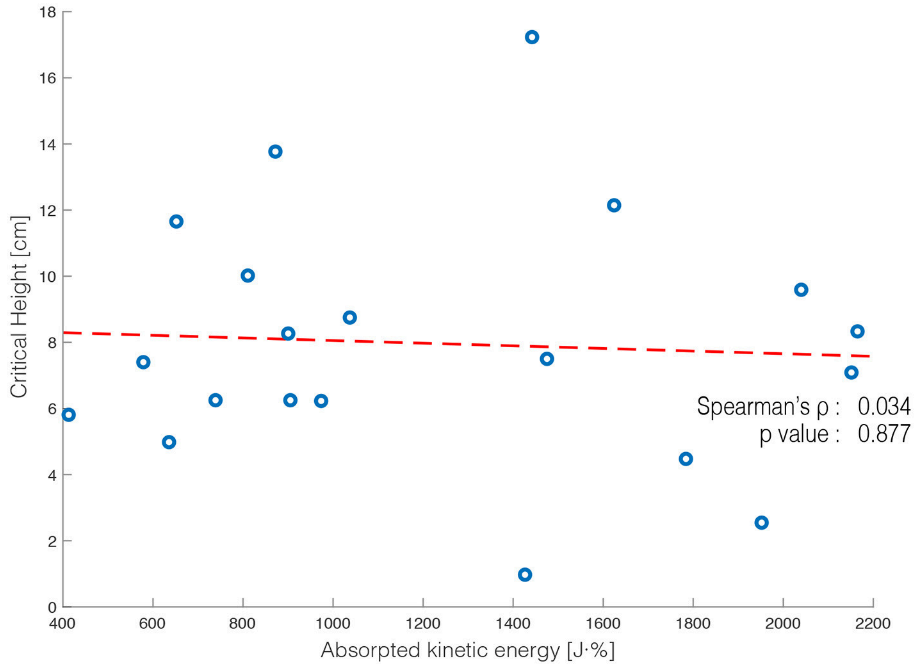

FIGURE 9 | The critical height (h_{crit}) and the kinetic energy absorbed after unexpected stepping down displayed for each participant (circles). A best fit line was fitted to the data and the corresponding Spearman's rho and *p*-value are shown.

et al., 2009)–it is unlikely that participants could have anticipated to the changed circumstances. In support, we did not observe adjustments in kinetic energy prior to the expected landing (**Figure 5**). Yet, analysis of the angular momentum revealed that before the instant of expected landing the momentum was altered (see **Supplementary Material**). In the unlikely situation that participants adjusted their behavior, our actual ability measure (i.e., the height of the peak in kinetic energy) could depend on the timing of the perturbation.

It is worth noting that individuals with poor ability to recover from strong perturbations, may be able to handle small perturbations very adequately (c.f., Bruijn et al., 2013). In this study, we determined the participants' ability only once and did not continue until they failed to recover from the perturbation. Given that none of the participants fell, it can be debated how threatening the unexpected step down was. As 12% of falls in older adults occur after erroneous foot placements (e.g., an unexpected step inside a small aperture in the pavement, Berg et al., 1997), we are confident that the manipulation provoked a balance threat that could have led to falling in daily life; however, future research should confirm this.

Imprecise Perception of Abilities

The second proposition that could explain the weak association between the strategy selected by participants and the actual ability to recover, is that older adults generally have an imprecise perception of their actual abilities in relation to the task at hand. This is in accordance with previous studies investigating the discrepancy between the perceived and actual step ability (Sakurai et al., 2013, 2016; Kluft et al., 2016, 2017), which found that approximately one-third of older adults either over-or underestimate their abilities, when explicitly asked for their perceived ability. However, in these studies, the perceived and actual ability measures were nevertheless significantly associated, in contrast with the present result.

Our sample consisted of a relative homogenous and fit subgroup of older adults (see **Table 1**)–enforced by the inclusion criteria–which might have hampered the association between the strategy selected by participants and the actual ability, due to a limited variability in actual abilities. Hence, these findings cannot be extrapolated to the complete population of older adults.

Another source of uncertainty is that aging is associated with a decrease in the ankle plantar flexion torque (Judge et al., 1996). In this study the calf strength was not recorded in the current study. Toe-landing strategies rely heavily on the ability to generate ankle plantar flexion torque; an age-related decrease in the strength of the calf musculature could therefore lead to a decrease of the occurrences of toe landings. On the contrary, overall older adults prefer to step down relatively smaller step heights using toe landings compared to young adults (van Dieën and Pijnappels, 2009).

As the aim of this study was to assess the relations between physical ability and behavioral choice, we did not ask for explicit rating of the participants' perception of their ability. Therefore, we cannot rule out the possibility that the discrepancy between strategy selection and physical ability may have been caused by other factors that could affect strategy selection such as psychological factors (e.g., fear, anxiety, and self-confidence), or physiological factors (e.g., reduced visual acuity Buckley et al., 2005). To develop a full picture of the formation of motor strategy selection, future research focussing on strategy selection and psychological factors is therefore recommended.

CONCLUSION

Overall, we did not find a significant association between strategy selection and actual ability. This suggests that the older adults in our study either did not select their movement strategy for stepping down in line with their actual abilities in terms of their ability to absorb kinetic energy after unexpected stepping down, or had an imprecise perception of their actual abilities. Future research should evaluate whether this motor strategy selection is affected by psychological factors, and whether accidental falls in older adults are results of selecting inadequate strategies.

AUTHOR CONTRIBUTIONS

NK, SB, JvD, and MP contributed to the conception and design of the study. NK build the set up and collected the experimental data. NK and SB developed the code for the data analysis. NK took the lead in writing the manuscript and designed the figures. All authors provided critical feedback and helped shape the research, analysis and manuscript. MP supervised the project. All authors contributed to manuscript revision, read and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2018.01419/full#supplementary-material

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Modular Control of Human Movement During Running: An Open Access Data Set

Alessandro Santuz[1,2], Antonis Ekizos[1,2], Lars Janshen[1], Falk Mersmann[1,2], Sebastian Bohm[1,2], Vasilios Baltzopoulos[3] and Adamantios Arampatzis[1,2]**

[1] Department of Training and Movement Sciences, Humboldt-Universität zu Berlin, Berlin, Germany, [2] Berlin School of Movement Science, Humboldt-Universität zu Berlin, Berlin, Germany, [3] Research Institute for Sport and Exercise Sciences, Liverpool John Moores University, Liverpool, United Kingdom

Edited by:
Lars Donath,
German Sport University Cologne,
Germany

Reviewed by:
Ramona Ritzmann,
Albert-Ludwigs-Universität Freiburg,
Germany
Gennaro Boccia,
Università degli Studi di Torino, Italy

***Correspondence:**
Alessandro Santuz
alessandro.santuz@hu-berlin.de
Adamantios Arampatzis
a.arampatzis@hu-berlin.de

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The human body is an outstandingly complex machine including around 1000 muscles and joints acting synergistically. Yet, the coordination of the enormous amount of degrees of freedom needed for movement is mastered by our one brain and spinal cord. The idea that some synergistic neural components of movement exist was already suggested at the beginning of the 20th century. Since then, it has been widely accepted that the central nervous system might simplify the production of movement by avoiding the control of each muscle individually. Instead, it might be controlling muscles in common patterns that have been called muscle synergies. Only with the advent of modern computational methods and hardware it has been possible to numerically extract synergies from electromyography (EMG) signals. However, typical experimental setups do not include a big number of individuals, with common sample sizes of 5 to 20 participants. With this study, we make publicly available a set of EMG activities recorded during treadmill running from the right lower limb of 135 healthy and young adults (78 males and 57 females). Moreover, we include in this open access data set the code used to extract synergies from EMG data using non-negative matrix factorization (NMF) and the relative outcomes. Muscle synergies, containing the time-invariant muscle weightings (motor modules) and the time-dependent activation coefficients (motor primitives), were extracted from 13 ipsilateral EMG activities using NMF. Four synergies were enough to describe as many gait cycle phases during running: weight acceptance, propulsion, early swing, and late swing. We foresee many possible applications of our data that we can summarize in three key points. First, it can be a prime source for broadening the representation of human motor control due to the big sample size. Second, it could serve as a benchmark for scientists from multiple disciplines such as musculoskeletal modeling, robotics, clinical neuroscience, sport science, etc. Third, the data set could be used both to train students or to support established scientists in the perfection of current muscle synergies extraction methods. All the data is available at Zenodo (doi: 10.5281/zenodo.1254380).

Keywords: muscle synergies, locomotion, running, motor control, EMG, data set

INTRODUCTION

The popularity of endurance running has greatly increased over the last few decades (Burfoot, 2007). Other than a purely social phenomenon, endurance running has been the center of attention in many research fields. Evolutionary anthropology, for instance, has been used to try to explain why humans show exceptional endurance running speeds compared to non-human primates (Bramble and Lieberman, 2004). In the neurosciences, running has long been ideal object for the investigation of movement due to its automatized, synergistic, general, cyclic, and phylogenetically old nature (Bernstein, 1967). However, a consensus as to how humans coordinate this repetitive and highly stereotyped movement is still far from being reached, mostly because direct experimental proofs are lacking.

Since the second half of last century it has been suggested that the central nervous system might simplify the production of movements by avoiding the complexity of activating each muscle individually (Bernstein, 1967; Lee, 1984; Bizzi et al., 1991, 2008). This feature might be implemented by linearly combining a small set of time-dependent commands, which have been called muscle synergies (Tresch et al., 1999). The idea that some synergistic neural components of movement exist was already suggested by Sherrington (1906) at the beginning of the 20th century. However, the so-called degrees of freedom problem and related principle of motor abundance was formally described only a few decades later (Bernstein, 1967). Following the thoughts of Bernstein, Lee (1984) published an essay in which the idea of "neuromotor synergies," defined as neurally based units of voluntary action, was explored and supported. Bizzi et al. (1991) were the first to experimentally show spinal synergies, which they represented as force fields. However, only in 1999 Lee's ideas could be numerically represented by showing the movement-specific recruitment of a small set of synergistic muscles in the spinal frog (Tresch et al., 1999). In the same year, Lee and Seung (1999) introduced the non-negative matrix factorization (NMF), a computational tool for extracting synergies from any kind of analyzed variable. To date, the NMF is one of the most common methods for reducing the high dimensional electromyographic (EMG) input into a small number of synergies (D'Avella, 2016).

Compared to the direct analysis of EMG signals, the muscle synergies concept has the clear advantage of being a high-throughput approach for analyzing muscle activities. In fact, it does not only provide the researcher with an automatic, low-dimensional, clustering of the activations during the gait cycle, but it also identifies the weighted contribution of each muscle for producing a certain movement. A number of studies were able to provide indirect evidence that synergies reside in the brainstem or spinal cord and follow a modular organization (Tresch et al., 2002; Hart and Giszter, 2004; Bizzi et al., 2008; Roh et al., 2011; Bizzi and Cheung, 2013). Synergies can be seen as low dimensional units that, via descending or afferent pathways, produce a complex electromyographic (EMG) pattern in muscles (Tresch et al., 2002; Bizzi and Cheung, 2013), creating a locomotor drive mediated by a certain amount of supraspinal control (Overduin et al., 2015). Synergies similar to those found in humans at a spinal (Ivanenko et al., 2006) or muscular level can be observed also in the motor cortex of the primate (Overduin et al., 2015) and cat (Yakovenko et al., 2011). This suggests a high degree of cooperation within the central nervous system's structure at all levels.

Amongst the various types of locomotion, running has been object of several studies involving muscle synergies (Cappellini et al., 2006; Lacquaniti et al., 2012; Hagio et al., 2015; Yokoyama et al., 2016; Nishida et al., 2017; Santuz et al., 2017a,b, 2018). It is well accepted that human running can be described, in young and healthy individuals, with a few muscle synergies (Cappellini et al., 2006; Santuz et al., 2017b, 2018). Specifically, when analyzing the EMG activities of lower limb muscles, usually 4 or 5 synergies are observed (Cappellini et al., 2006; Santuz et al., 2017b, 2018). However, the data sets considered by existing studies are usually quite small (commonly including between 5 and 20 participants) and not freely available (Cappellini et al., 2006; Hagio et al., 2015; Yokoyama et al., 2016; Nishida et al., 2017; Santuz et al., 2017a,b, 2018). Moreover, a consensus regarding factorization techniques, data conditioning, and interpretation is not unanimous (Devarajan and Cheung, 2014; Oliveira et al., 2014; Santuz et al., 2017a; Shuman et al., 2017; Kieliba et al., 2018; Soomro et al., 2018). Human data sets are more and more frequently being published and made available to everyone (Makihara et al., 2012; Van Den Bogert et al., 2013; Wang and Srinivasan, 2014; Auton et al., 2015; Moore et al., 2015; Santos and Duarte, 2016; Fukuchi et al., 2018). However, sample sizes can be highly variable. The advantages related to the increased volume and variety of data sources mainly lie in the broadened representation of human variability, improvement of analysis strategies, and shareability for both scientific and educational purposes.

With this study, we present an open access data set of EMG and synergy data for running in young and healthy humans. The presented data is available in three formats: (1) the raw EMG, unprocessed together with the touchdown and lift-off timings of the recorded limb; (2) the filtered and time-normalized EMG; and (3) the muscle synergies extracted via NMF. Several applications based on this data set can be foreseen. From obtaining a deeper, more extended representation of motor coordination during running, to increase the detail of musculoskeletal models and robotic controls, passing through the improvement of current factorization methods and didactic purposes.

MATERIALS AND METHODS

Experimental Protocol

For the development of the data set we recruited 135 volunteers (78 males and 57 females, height 175 ± 9 cm, body mass 69 ± 11 kg, age 30 ± 5 years, means ± standard deviation). The metadata file "participants_data.dat" includes the age and anthropometric data of the participants. All volunteers were regularly active and did not use orthotic insoles. None had any history of neuromuscular or musculoskeletal impairments, or any head or spine injury at the time of the measurements or in the previous 6 months. This study was reviewed and approved by the

Ethics Committee of the Humboldt-Universität zu Berlin. All the participants gave written informed consent for the experimental procedure, in accordance with the Declaration of Helsinki.

The data recordings were performed while the participants were running on a treadmill (mercury, H-p-cosmos Sports & Medical GmbH, Nussdorf, Germany) equipped with a pressure plate recording the plantar pressure distribution at 120 Hz (FDM-THM-S, zebris Medical GmbH, Isny im Allgäu, Germany). The muscle activity of 13 ipsilateral muscles was recorded with a frequency of 1000 Hz using a 16-channel wireless bipolar EMG system (myon m320, myon AG, Schwarzenberg, Switzerland). For the EMG recordings, we used wet-gel silver/silver chloride electrodes with foam backing material and snap connector (BlueSensor N-00-S/25, Ambu A/S, Ballerup, Denmark).

After a self-selected warm-up (Santuz et al., 2017b), the participants ran on the treadmill at an average speed of 2.65 ± 0.31 m/s (details on speed are provided in the metadata file "participants_data.dat") for the time necessary to record two trials of 60 s each. The reason for choosing this particular speed is that some participants ran at a pre-defined speed (2.2, 2.5, 2.8, or 3.0 m/s), while others were asked to find and run at their comfortable speed, depending on the experimental setup in which the data was collected (details on speed type are provided in the metadata file "participants_data.dat"). This was due to the fact that data was collected in different experimental setups. The procedure to find the comfortable speed was implemented using the method of limits (Treutwein, 1995), as follows. The speed was randomly increased with steps of 0.02 to 0.05 m/s at varying time intervals (around 5 to 10 s) until the participant was comfortable with a specific pace. The operation was then repeated starting from a faster speed and randomly decreasing it as previously done. If the comfortable speed value did not differ more than 10% from the previous, the average of the two values was taken as the preferred. Otherwise, the whole procedure was repeated as needed. In both the pre-defined and the preferred speed protocols, the warm-up procedure, including the selection of speed where applicable, typically lasted between 5 and 10 min.

Gait Cycle Parameters

The pressure plate's raw data was acquired using the proprietary software (WinFDM-T v2.5.1, zebris Medical GmbH, Isny im Allgäu, Germany) and then extracted in raw format for autonomous post-processing of the gait spatiotemporal parameters using a validated custom algorithm (Santuz et al., 2016) written in R version 3.5.1 (R Foundation for Statistical Computing, R Core Team, Vienna, Austria). As an indication of the foot strike pattern, the strike index was calculated using a validated algorithm based on the numerical analysis of foot pressure distribution (Santuz et al., 2016). As originally defined by Cavanagh and Lafortune (1980), we calculated the strike index as the distance from the heel to the center of pressure at impact relative to total foot length (thus the values range from 0 to 1). For the participants P0015 through P0032 the strike index values were not available.

EMG Data

The muscle activity of the following 13 ipsilateral (right side) muscles was recorded (see **Table 1** for details): *gluteus medius* (ME), *gluteus maximus* (MA), *tensor fasciæ latæ* (FL), *rectus femoris* (RF), *vastus medialis* (VM), *vastus lateralis* (VL), *semitendinosus* (ST), *biceps femoris* (long head, BF), *tibialis anterior* (TA), *peroneus longus* (PL), *gastrocnemius medialis* (GM), *gastrocnemius lateralis* (GL), and *soleus* (SO). We recorded two trials of 30 s for each participant. The EMG signals were high-pass filtered, then full-wave rectified and low-pass filtered (Santuz et al., 2017a) using a 4th order IIR Butterworth zero-phase filter with cut-off frequencies 50 Hz (high-pass) and 20 Hz (low-pass for the linear envelope) using R v3.5.1 (R Found. for Stat. Comp.). After filtering, any negative value was set to zero. Then, all the zero entries were set to the smallest non-zero value. The amplitude was normalized to the maximum activation recorded for each participant across both trials (Bizzi et al., 2008; Chvatal and Ting, 2013; Devarajan and Cheung, 2014; Santuz et al., 2018). Each gait cycle was then time-normalized to 200 points, assigning 100 points to the stance and 100 points to the swing phase (Santuz et al., 2017b, 2018). The reason for this choice was twofold. First, dividing the gait cycle into two macro-phases helps the reader understanding the temporal contribution of the different synergies, diversifying between stance and swing. Second, normalizing the duration of stance and swing to the same number of points for all participants (and for all the recorded gait cycles of each participant) is needed to make the interpretation of the results independent from the absolute duration of the gait events.

Muscle Synergies Extraction

Muscle synergies were extracted through a custom script (Santuz et al., 2017b, 2018) (R v3.5.1, R Found. for Stat. Comp.) using the

TABLE 1 | Muscles considered for the extraction of muscle synergies (ipsilateral, right side of the body).

Upper leg	Gluteus medius[a]
	Gluteus maximus[b]
	Tensor fasciæ latæ[c]
	Rectus femoris
	Vastus medialis
	Vastus lateralis
	Semitendinosus
	Biceps femoris (long)
Lower leg	Tibialis anterior
	Peroneus longus
	Gastrocnemius medialis
	Gastrocnemius lateralis
	Soleus[d]

Unless specified differently, the electrodes were positioned on the middle of muscle belly, along the main direction of the fibers. The specifications follow the SENIAM (Surface EMG for non-invasive assessment of muscles) recommendations. [a]Middle of line between iliac crest and greater trochanter. [b]Middle of line between sacral vertebræ and greater trochanter. [c]Line from anterior spina iliaca superior to lateral femoral condyle in the proximal 1/6. [d]At 2/3 of line between medial condyle of femur to medial malleolus.

classical Gaussian NMF algorithm (Lee and Seung, 1999; Santuz et al., 2017a) from the first 30 gait cycles of each acquisition. The $m = 13$ time-dependent muscle activity vectors were grouped in an $m \times n$ matrix V (n = 30 gait cycles * 200 points = 6000 points), factorized such that $V \approx V_R = WH$, where V_R represents the new reconstructed matrix, which approximates the original matrix V. The motor primitives (Dominici et al., 2011; Santuz et al., 2017a) matrix H contained the time-dependent coefficients of the factorization with dimensions $r \times n$, where r represents the minimum number of synergies necessary to reconstruct the original signals (V). The motor modules (Gizzi et al., 2011; Santuz et al., 2017a) matrix W with dimensions $m \times r$, contained the time-invariant muscle weightings, which describe the relative contribution of single muscles within a specific synergy (a weight was assigned to each muscle for every synergy). H and W described the synergies necessary to accomplish a movement. The update rules for H and W are presented in Equations (1) and (2).

$$\begin{cases} H_{i+1} = H_i \frac{W_i^T V}{W_i^T W_i H_i} & (1) \\ W_{i+1} = W_i \frac{V (H_{i+1})^T}{W_i H_{i+1} (H_{i+1})^T} & (2) \end{cases}$$

The quality of reconstruction was assessed by measuring the coefficient of determination R^2 between the original and the reconstructed data (V and V_R, respectively). The limit of convergence was reached when a change in the calculated coefficient of determination R^2 between V and V_R was smaller than the 0.01% in the last 20 iterations (Cheung et al., 2005; Santuz et al., 2017a), meaning that, with that amount of synergies, the signal could not be reconstructed any better. This operation was started by setting the number of synergies to 1. Then, it was repeated by increasing the number of synergies each time, until a maximum of 10 synergies. The number 10 was chosen to be lower than the number of muscles, since it would not make sense to extract a number of synergies equal to the number of measured EMG activities. The computation was repeated 10 times for each of the previous 10 steps, each time creating new randomized initial matrices H and W, in order to avoid local minima (D'Avella and Bizzi, 2005; Santuz et al., 2017a). The solution with the highest R^2 was then selected for each of the 10 synergies. To choose the minimum number of synergies required to represent the original signals, the curve of R^2 values versus synergies was fitted using a linear regression model, using all 10 synergies. The mean squared error (Cheung et al., 2005; Santuz et al., 2017a) between the curve and the linear interpolation was then calculated. Afterward, the first point in the R^2-vs.-synergies curve was removed and the error between this new curve and its new linear interpolation was calculated. The operation was repeated until only two points were left on the curve or until the mean squared error fell below 10^{-5}. This method searches for the most linear part of the R^2-vs.-synergies curve and it is equivalent to stating that the reconstruction quality is not improving much when the curve becomes linear. With this approach, the need for setting a threshold to the reconstruction quality is avoided, giving space to the possibility that quality might not improve at the same rate when the same NMF algorithm is applied to different data.

The aforementioned procedure allowed us to extract fundamental and synergies from the raw EMG data. A fundamental synergy can be defined as an activation pattern whose motor primitive shows a single peak of activation (Santuz et al., 2017a). When two or more fundamental synergies are blended into one, a combined synergy is identified (Santuz et al., 2017a,b, 2018).

RESULTS

Metadata

The file "participants_data.dat" is available at Zenodo (doi: 10.5281/zenodo.1254380) in ASCII and RData (R Found. for Stat. Comp.) format and contains:

- **Code**: the participant's code
- **Sex**: the participant's sex (M or F)
- **Speed**: the speed at which the recordings were conducted in [m/s]
- **Type**: gives information on whether the participant ran at their preferred (PR) or fixed (FX) speed
- **Age**: the participant's age in years
- **Height**: the participant's height in [cm]
- **Mass**: the participant's body mass in [kg]
- **SI**: the strike index, dimensionless quantity defined as reported in the methods section, reported as the average value of all the steps recorded in both trials; values referred to the right foot.

Gait Cycle Parameters

The files containing the gait cycle breakdown are available at Zenodo (doi: 10.5281/zenodo.1254380) in ASCII and RData (R Found. for Stat. Comp.) format. The files are structured as data frames with 30 rows (one for each gait cycle) and two columns. The first column contains the touchdown incremental times in seconds. The second column contains the duration of each stance phase in seconds. Each trial is saved both as a single ASCII file and as an element of a single R list. Trials are named like "CYCLE_TIMES_P0026_02," where the characters "CYCLE_TIMES" indicate that the trial contains the gait cycle breakdown times, the characters "P0026" indicate the participant number (in this example the 26th) and the last two characters indicate the number of trial for that participant (either "01" for the first trial or "02" for the second). The average contact times were of 288 ± 42 ms, with an average swing time of 452 ± 45 ms at 163 ± 10 steps/min. The strike index of the right foot was on average 0.152 ± 0.195, with a maximum of 0.699 and a minimum of 0.011. In total, 82.5% of the participants had a strike index lower than 0.333 (rearfoot strike pattern).

EMG Data

The files containing the raw, filtered and the normalized EMG data are available at Zenodo (doi: 10.5281/zenodo.1254380) in ASCII and RData (R Found. for Stat. Comp.) format. The raw EMG files are structured as data frames with 30000 rows (one for each recorded data point) and 14 columns. The first column contains the incremental time in seconds. The remaining thirteen columns contain the raw EMG data, named with muscle abbreviations that follow those reported in the "Materials and Methods" section of this paper. Each trial is saved both as a single ASCII file and as an element of a single R list. **Figure 1** represents a typical filtering process for an EMG signal. In **Figure 2** we report the EMG data acquired from one participant during one trial (cycles are superimposed and the average filtered signals are presented as well). Trials are named like "RAW_EMG_P0026_02," where the characters "RAW_EMG" indicate that the trial contains raw EMG data, the characters "P0026" indicate the participant number (in this example the 26th) and the last two characters indicate the number of trial for that participant (either "01" for the first trial or "02" for the second). The filtered and time-normalized EMG data is named, following the same rules, like "FILT_EMG_P0026_02."

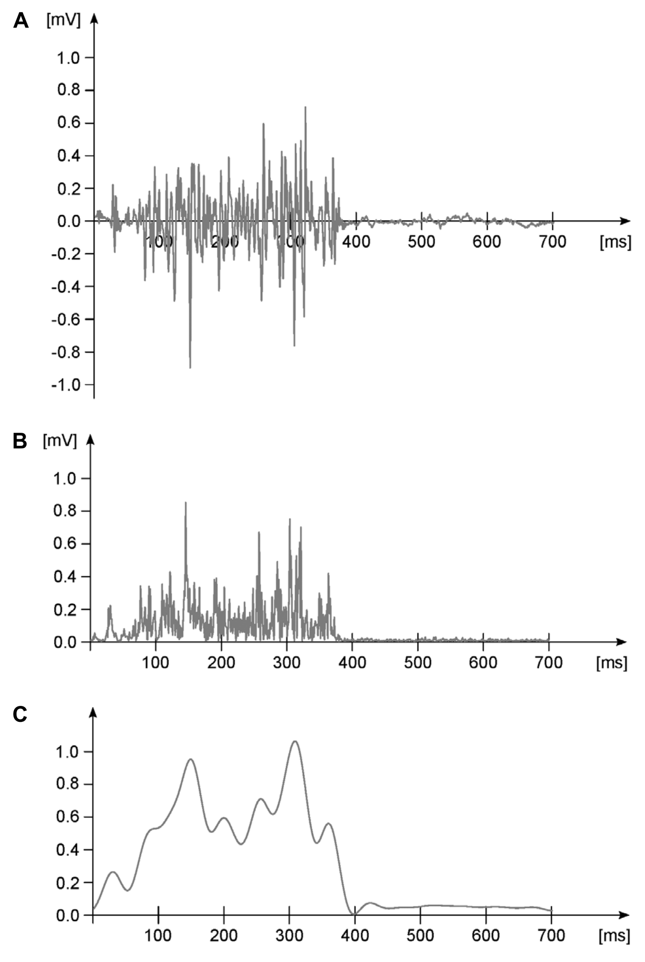

FIGURE 1 | Exemplary EMG activity of one muscle during one gait cycle. **(A)** Raw data. **(B)** Raw data after high-pass filtering (4th order IIR Butterworth zero-phase filter, cut-off frequency 50 Hz) and full-wave rectification. Panel **(C)** rectified and high-pass filtered data after low-pass filtering (4th order IIR Butterworth zero-phase filter, cut-off frequency 20 Hz) and normalization to the maximum (dimensionless *y*-axis units). This last step is done for creating the linear envelope of the signal.

Muscle Synergies

The files containing the muscle synergies extracted from the filtered and normalized EMG data are available at Zenodo (doi: 10.5281/zenodo.1254380) in ASCII and RData (R Found. for Stat. Comp.) format. The muscle synergies files are divided in motor primitives and motor modules and are presented as direct output of the factorization and not in any functional order.

Motor primitives are data frames with a number of rows equal to the number of synergies (which might differ from trial to trial) and 6000 columns. The rows contain the time-dependent coefficients (motor primitives), one row for each synergy (named, e.g., "Syn1, Syn2, Syn3", where "Syn" is the abbreviation for "synergy"). Each gait cycle contains 200 data points, 100 for the stance and 100 for the swing phase which, multiplied by the 30 recorded cycles, result in 6000 data points distributed in as many columns. Each set of motor primitives relative to one synergy is saved both as a single ASCII file and as an element of a single R list. Trials are named like "SYNS_H_P0026_02," where the characters "SYNS_H" indicate that the trial contains motor primitive data, the characters "P0026" indicate the participant number (in this example the 26th) and the last two characters indicate the number of trial for that participant (either "01" for the first trial or "02" for the second).

Motor modules are data frames with 13 rows and a number of columns equal to the number of synergies (which might differ from trial to trial). The rows, named with muscle abbreviations that follow those reported in the methods section of this paper, contain the time-independent coefficients (motor modules), one for each synergy and for each muscle. Each set of motor modules relative to one synergy is saved both as a single ASCII file and as an element of a single R list. Trials are named like "SYNS_W_P0026_02," where the characters "SYNS_W" indicate that the trial contains motor module data, the characters "P0026" indicate the participant number (in this example the 26th) and the last two characters indicate the number of trial for that participant (either "01" for the first trial or "02" for the second).

Figure 3 is an example of how muscle synergies can be graphically represented. The recorded muscle activations can be approximated by the linear combination of motor modules and motor primitives. Since they are time-invariant coefficients, motor modules are usually represented with bar graphs. On the contrary, motor primitives describe the evolution over time of the basic activation patterns and are therefore better represented with time-dependent curves. When multiplying and summing synergy-by-synergy the elements of the two matrices W (motor modules) and H (motor primitives), it is possible to reconstruct the original set of EMG data. For instance, it is possible to notice from **Figure 3** that the muscle PL, GM, GL, and SO are the major contributors to the second synergy, named "Propulsion." In fact, these ankle plantar flexors are important during the push-off in running, a phase that chronologically succeeds the weight acceptance (first synergy) and precedes the early swing (third synergy). The chronological order of synergies can be seen in the motor primitives, the fundamental activation patterns that

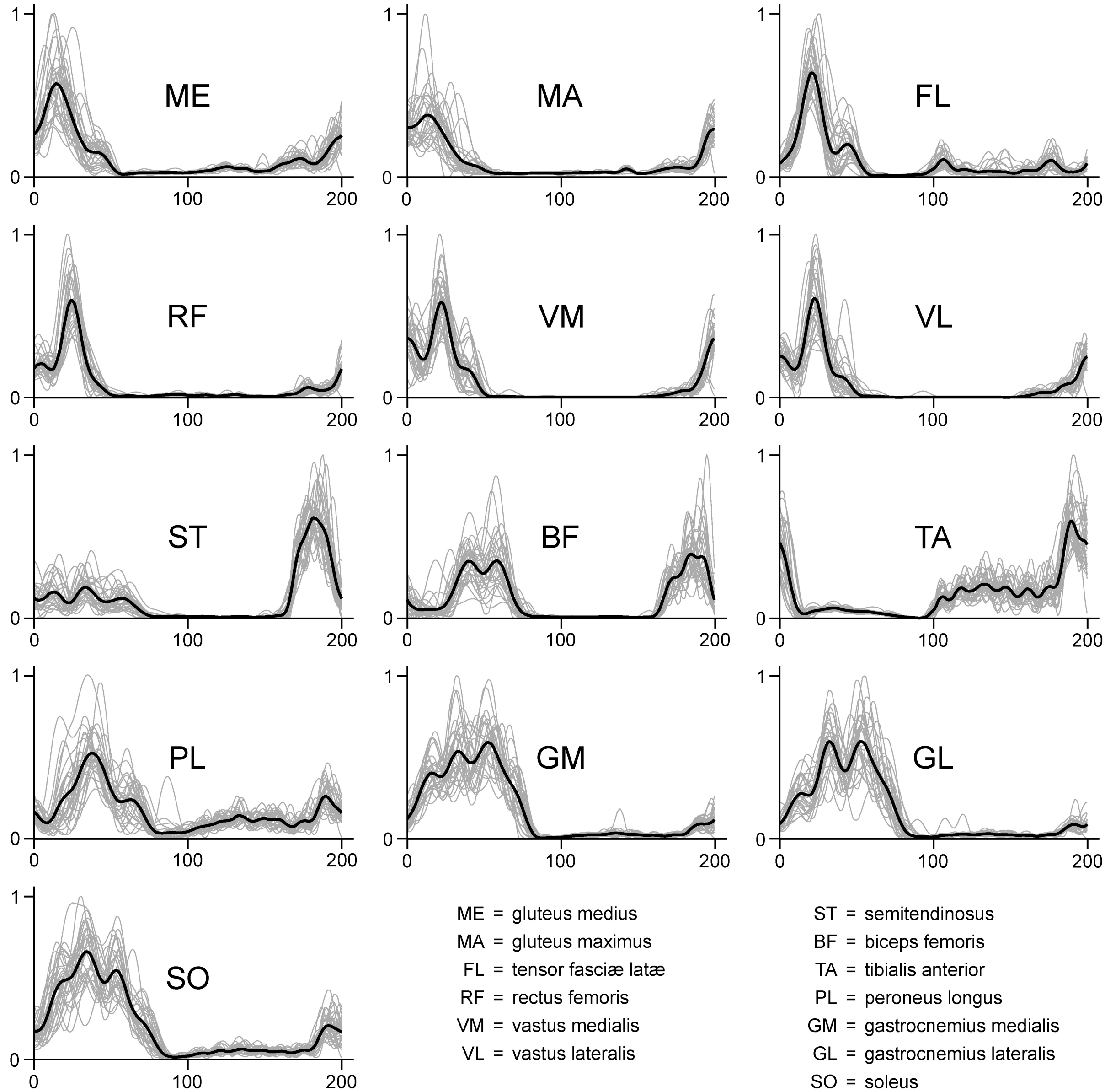

FIGURE 2 | Exemplary EMG activity of the 13 recorded muscles recorded for one participant during one trial (treadmill running). Signals were high-pass filtered (4th order IIR Butterworth zero-phase filter, cut-off frequency 50 Hz), full-wave rectified, low-pass filtered (4th order IIR Butterworth zero-phase filter, cut-off frequency 20 Hz) and normalized to the maximum (dimensionless *y*-axis units). Since each gait cycle was time-normalized to 200 points, in each graph the first 100 points on the *x*-axis represent the stance phase, while the second 100 points represent the swing.

describe the evolution over time of those commands which are common to differently functional groups of muscles (e.g., the plantar flexors in the second synergy).

The minimum number of synergies necessary to sufficiently describe the measured EMG activity during running was 4.7 ± 0.7. Excluding the combined synergies, four fundamental activation patterns could be identified (**Figure 3**). The four fundamental synergies were associated with temporally different phases of the gait cycle. The first synergy functionally referred to the body weight acceptance, with a major involvement of knee extensors and glutei. The second synergy described the propulsion phase, to which the plantar flexors mainly contributed. The third synergy identified the early swing, showing the involvement of foot dorsiflexors. The fourth and last synergy reflected the late swing and the landing preparation, highlighting the relevant influence of knee flexors and foot dorsiflexors.

Code

All the code used for the preprocessing of EMG data and the extraction of muscle synergies is available at Zenodo (doi: 10.5281/zenodo.1254380) in R (R Found. for Stat.

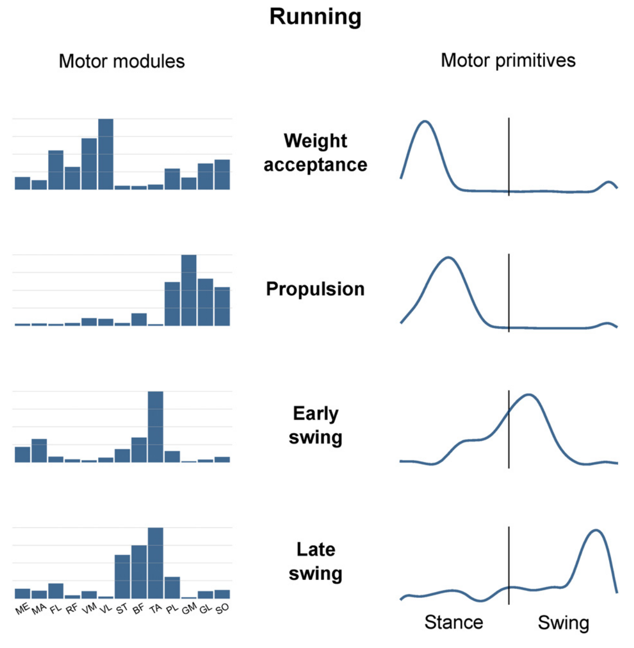

FIGURE 3 | Exemplary motor modules and motor primitives of the four fundamental synergies for human running (one trial). The motor modules are presented on a normalized y-axis base. For the motor primitives, the x-axis full scale represents the averaged gait cycle (with stance and swing normalized to the same amount of points and divided by a vertical line) and the *y*-axis the normalized amplitude. Muscle abbreviations: ME, gluteus medius; MA, gluteus maximus; FL, tensor fasciæ latæ; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; ST, semitendinosus; BF, biceps femoris; TA, tibialis anterior; PL, peroneus longus; GM, gastrocnemius medialis; GL, gastrocnemius lateralis; SO, soleus.

Comp.) format. Explanatory comments are profusely present throughout the scripts ("SYNS.R," which is the main script and "fun_synsNMFn.R," which contains the NMF function).

DISCUSSION

With this study, we make available a large data set of lower-limb EMG activity recorded during human running. Data was acquired from 13 ipsilateral lower limb muscles in 135 young and healthy individuals. An exemplary script, which can be used to pre-process and factorize the EMG data via NMF, is also part of the data set. A metadata file contains the relevant demographic and anthropometric data of the participants, together with important information regarding the experimental conditions and the general guidelines to interpret muscle synergy data.

The etymology of the word "synergy" is nested in the Greek language. Literally, synergy means "working together" (συνεργός). The idea that some synergistic neural components of movement exist was already suggested by Sherrington (1906) at the beginning of the 20th century. In his famous "The integrative action of the nervous system," Sherrington (1906) wrote "The stimulation [. . .] excites reflexly through the central organ an effect in the skeletal musculature which is co-ordinate and synergic." Yet, Sherrington (1906) took some distance from the concept of a functional organization of the motor spinal root, arguing that "the collection of fibers in a spinal motor root is not a functional collection in the sense that it is representative of any co-ordination." Bernstein (1967) published his "The co-ordination and regulation of movements," a book that became a milestone in the history of muscle synergies. For the first time, Bernstein (1967) formally described the so-called "degrees of freedom problem," stating that "the basic difficulties for co-ordination consist precisely in the extreme abundance of degrees of freedom, with which the [CNS] [. . .] is not at first in a position to deal." This concept of motor abundance is still one of the supporting pillars of modern motor control and laid the foundation of the muscle synergies idea. In the past two decades, the scientific publications embracing the concept of muscle synergies have been flourishing and exponentially increasing in number. Even if the consensus on factorization techniques, data conditioning and interpretation is not unanimous, it is well accepted that human locomotion can be described with a small number of synergies. When analyzing the EMG activities of lower limb's muscles (Santuz et al., 2017b, 2018), this number is usually equal to 4 or 5. A synergy might add when considering the upper body (Cappellini et al., 2006; Santuz et al., 2017a).

There are several examples of studies employing factorization of EMG activity to study human locomotion. For several reasons, the most widespread locomotion type that has been studied is walking (Ivanenko et al., 2004; Cappellini et al., 2006; Courtine et al., 2006; Clark et al., 2010; McGowan et al., 2010; Dominici et al., 2011; Allen and Neptune, 2012; Bolton and Misiaszek, 2012; Chvatal and Ting, 2012, 2013; Lacquaniti et al., 2012; Oliveira et al., 2012; Rodriguez et al., 2013; Barroso et al., 2014; Maclellan et al., 2014; Routson et al., 2014; Coscia et al., 2015; Gonzalez-Vargas et al., 2015; Hagio et al., 2015; Licence et al., 2015; Martino et al., 2015; Nazifi et al., 2015; Tang et al., 2015; Buurke et al., 2016; Gui and Zhang, 2016; Kim et al., 2016; Lencioni et al., 2016; Meyer et al., 2016; Pérez-Nombela et al., 2016; Yokoyama et al., 2016; Allen et al., 2017; Janshen et al., 2017; Santuz et al., 2017a; Shuman et al., 2017; Saito et al., 2018). Due to the easiness of examining this slow-speed type of locomotion, it is not a surprise that the majority of studies use walking as the main object of investigation. Also, it is clear that, contrarily to other locomotion types such as running, walking can be easily performed by patients, children and elderly and this feature notably extends the basin of potential participants. Nonetheless, running has been receiving increasing attention (Cappellini et al., 2006; Lacquaniti et al., 2012; Hagio et al., 2015; Yokoyama et al., 2016; Nishida et al., 2017; Santuz et al., 2017a,b, 2018) as well. This might be partially due to the growing popularity of distance running as a recreational sport activity over the last three decades (Burfoot, 2007). Another reason to choose running over walking (or to study both conditions within the same experimental setup) is that, due to the different absolute and relative length of the stance and swing phases, different control mechanisms are likely to be used by the CNS (Biewener and Daley, 2007; Santuz et al., 2018). Concerning this last matter, though, the field is still much open to new ideas, insights and exciting findings (Santuz et al., 2018). Unavoidably, the links between locomotion velocity and modular organization have been investigated as well (Ivanenko et al., 2004; Cappellini et al., 2006; Routson et al., 2014; Coscia et al., 2015; Gonzalez-Vargas et al., 2015; Hagio et al., 2015; Buurke et al., 2016; Gui and Zhang, 2016; Yokoyama et al., 2016). However, results are often contradictory and the reasons have not yet been clarified. Whether for computational or neurophysiological reasons, some studies found consistency in the recruitment of the same motor primitives and/or modules across varying velocities (Ivanenko et al., 2004; Cappellini et al., 2006; Routson et al., 2014; Buurke et al., 2016; Gui and Zhang, 2016), while others found walking-, running-, and/or velocity-specific sets of motor primitives and/or modules (Cappellini et al., 2006; Routson et al., 2014; Coscia et al., 2015; Gonzalez-Vargas et al., 2015; Yokoyama et al., 2016). The role of muscle synergies for locomotion in pathology has been a focus of a few groups in recent years (Latash and Anson, 2006; Clark et al., 2010; Giszter and Hart, 2013; Rodriguez et al., 2013; Routson et al., 2014; Coscia et al., 2015; Tang et al., 2015; Falaki et al., 2016; Lencioni et al., 2016; Meyer et al., 2016; Pérez-Nombela et al., 2016; Shuman et al., 2016, 2017; Wenger et al., 2016; Allen et al., 2017; Banks et al., 2017). Given the simplification in presenting the data due to the dimensionality reduction, it is appealing to think to a possible clinical application of the method. There have been comparisons between healthy and Parkinson's disease (Rodriguez et al., 2013; Falaki et al., 2016; Allen et al., 2017), multiple sclerosis patients (Lencioni et al., 2016), spinal cord injury (Giszter and Hart, 2013; Pérez-Nombela et al., 2016; Wenger et al., 2016), cerebral palsy (Li et al., 2013; Steele et al., 2015; Tang et al., 2015; Shuman et al., 2016, 2017), and post-stroke (Clark et al., 2010; Routson et al., 2014; Coscia et al., 2015; Meyer et al., 2016; Banks et al., 2017) patients. However, as for the studies on the influence of velocity on the modular organization of motion, also in pathology studies results are often difficult to interpret and require careful analysis. The study of the modular organization of locomotion in unsteady conditions has as well started to meet the interest of some research groups (Chvatal and Ting, 2012; Oliveira et al., 2012; Licence et al., 2015; Martino et al., 2015; Nazifi et al., 2015; Santuz et al., 2018), highlighting the importance of extending the controlled laboratory conditions to daily life.

The big, open access data set we present in this study, serves a threefold purpose. First, it increases the representative power of the data which is commonly obtainable with a standard experimental setup. Usually, due to experimental or design constraints, 5 to 20 individuals are recruited for each measurement campaign (Cappellini et al., 2006; Hagio et al., 2015; Yokoyama et al., 2016; Nishida et al., 2017; Santuz et al., 2017a,b, 2018). The choice is often dictated by the limited time available, difficulties in recruiting volunteers, budget limits, etc. With this publication, we make 135 (at the time of publication) young and healthy participants' data freely available and ready for numerical analysis. Our data can establish a baseline for those studies that aim to investigate, amongst others, different populations (such as elderly, children, patients, etc.) or conditions (walking, perturbed locomotion, etc.). Therefore, compared to a standard setup, the increased number of participants included in our study can be a prime source for broadening the representation of human motor control. While small samples might fail to capture the variety of population, the 135 proposed samples provide a preferential lane toward a more comprehensive description of the modular control of movement.

Second, the data could be used for many different scientific purposes in several research fields. For instance, both EMG and synergies data might be employed for the development of more advanced musculoskeletal models (Lai et al., 2014). Another possible application would be improving the control of active exoskeletons or robots for aiding or substituting human movement (Lai et al., 2014). The torques needed to generate a certain movement can be computed, but the complexity of motion equations dramatically increases with the number of degrees of freedom (D'Avella, 2016). Thus, synergies might be an effective way to store approximate yet sufficient information to build motor commands (D'Avella, 2016). This big data set might help scientists to transfer the knowledge coming from data acquired *in vivo* to *in silico* controls, providing a benchmark for what can be expected from artificial movement control.

Third, the data set could be used by other members of the scientific community interested in improving the existing or creating new muscle synergies extraction methods (Févotte et al., 2009; Devarajan and Cheung, 2014; Santuz et al., 2017a; Shuman et al., 2017; Kieliba et al., 2018; Soomro et al., 2018). This would greatly improve comparability across groups working in the field. For instance, several update rules have been and are continuously proposed for data factorization via NMF in a constant effort to improve their computational performance in terms of reconstruction capabilities and speed (Févotte et al., 2009; Devarajan and Cheung, 2014; Santuz et al., 2017a). However, to date, the classical Gaussian approach is the most used for EMG decomposition (Cappellini et al., 2006; Dominici et al., 2011; Santuz et al., 2017b). Also the choice of the minimum number of synergies necessary to sufficiently reconstruct the original signal is still matter of debate. Answering the question "how good is good enough?" has often led to an oversimplification of the issue, with many publications solving the problem by setting an arbitrary threshold on the R^2 values (Chvatal and Ting, 2013; Tang et al., 2015; Nishida et al., 2017). Moreover, some studies already investigated the influence of EMG preprocessing on muscle synergies (Santuz et al., 2017a; Shuman et al., 2017; Kieliba et al., 2018). Our data set provides a starting point for this kind of methodological studies. Last but not least, the educational potential of this data could be used to train students at all levels and from many different disciplines, from sport science, to medicine, from engineering, to mathematics and so forth.

It must be taken into account, however, that this data set has some limitations. First of all, it only includes data from young and healthy individuals. Thus, the data cannot be directly transferred to the study of children, adolescents, or elderly. Moreover, the muscles included in the recordings are limited to the lower limb. For extended considerations on the contribution of the upper body to the modular organization of running, more muscles should be included (Santuz et al., 2017a). Then, data was not recorded at the same speed for all participants, even if the average speed was close to the population's preferred (Santuz et al., 2016). Lastly, even though overground and treadmill running have been shown to share similar modular organization (Oliveira et al., 2016), this data set does only provide treadmill data.

AUTHOR CONTRIBUTIONS

AS, AE, LJ, FM, SB, VB, and AA contributed to conceptualization, writing the review, and editing. AS, LJ, and AA contributed to methodology. AS, AE, LJ, FM, and SB contributed to investigation. AS contributed to formal analysis and visualization. AS and AA wrote the original draft. VB and AA contributed to supervision.

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Changes of Maximum Leg Strength Indices During Adulthood a Cross-Sectional Study With Non-athletic Men Aged 19–91

Wolfgang Kemmler[1]*, *Simon von Stengel*[1], *Daniel Schoene*[1] *and Matthias Kohl*[2]

[1] *Institute of Medical Physics, University of Erlangen-Nürnberg, Erlangen, Germany,* [2] *Faculty of Medical and Life Sciences, University of Furtwangen, Schwenningen, Germany*

Age-related loss of muscle mass and function, also called sarcopenia, was recently added to the ICD-10 as an independent condition. However, declines in muscle mass and function are inevitable during the adulthood aging process. Concerning muscle strength as a crucial aspect of muscle function, maximum knee extension strength might be the most important physical parameter for independent living in the community. In this study, we aimed to determine the age-related decline in maximum isokinetic knee extension (MIES) and flexion strength (MIFS) in adult men. The primary study hypothesis was that there is a slight gradual decrease of MIES up to ≈age 60 years with a significant acceleration of decline after this "changepoint." We used a closed kinetic chain system (leg-press), which is seen as providing functionally more relevant results on maximum strength, to determine changes in maximum isokinetic hip/leg extensor (MIES) and flexor strength (MIFS) during adulthood in men. Apart from average annual changes, we aimed to identify whether the decline in maximum lower extremity strength is linear. MIES and MIFS data determined by an isokinetic leg-press of 362 non-athletic, healthy, and community-dwelling men 19–91 years old were included in the analysis. A changepoint analysis was conducted based on a multiple regression analysis adjusted for selected co-variables that might confound the proper relationship between age and maximum strength. In summary, maximum isokinetic leg-strength decline during adulthood averaged around 0.8–1.0% p.a.; however, the reduction was far from linear. MIES demonstrated a non-significant reduction of 5.2 N/p.a. (≈0.15% p.a.) up to the estimated breakpoint of 52.0 years and an accelerated loss of 44.0 N/p.a. (≈1.3% p.a.; $p < 0.001$). In parallel, the decline in MIFS (10.0 N/p.a.; ≈0.5% p.a.) prior to the breakpoint at age 59.0 years was significantly more pronounced. Nevertheless, we observed a further marked accelerated loss of MIFS (25.0 N/p.a.; ≈1.3% p.a.) in men ≥60 years. Apart from the "normative value" and closed kinetic chain aspect of this study, the practical application of our results suggests that sarcopenia prophylaxis in men should be started in the 5th decade in order to address the accelerated muscle decline of advanced age.

Keywords: maximum lower extremity strength, sarcopenia, dynamopenia, strength decline, leg-press

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Universität Duisburg-Essen, Germany

***Correspondence:**
Wolfgang Kemmler
wolfgang.kemmler@imp.uni-erlangen.de

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INTRODUCTION

Sarcopenia, characterized as a reduction of muscular mass and -function (Cruz-Jentoft et al., 2010; Fielding et al., 2011; Studenski et al., 2014) was included in the ICD-10 CM[1] code as a musculoskeletal disease in 2016 (M62.84). Although the relevance of muscle mass for healthy aging might be underestimated[2]; functional or more dedicated "dynamopenic"[3] (Greco et al., 2014) aspects are without doubt more important for older people's well-being and independent living.

In this context, studies have reported the particular crucial relevance of age-dependent declines in leg-extension/quadriceps strength on mobility limitations, disability, morbidity, and mortality in older people (Visser et al., 2005; Newman et al., 2006; Roshanravan et al., 2017). Unfortunately, the reduction in muscle strength in older age was reported to be much more pronounced (Goodpaster et al., 2006; Dey et al., 2009; Koster et al., 2011) than the decline in muscle mass. Further, maximum strength deterioration of the lower limbs was much higher compared with upper limbs (Viitasalo et al., 1985; Frontera et al., 2000; Landers et al., 2001; Dey et al., 2009; Amaral et al., 2014).

However, age-related declines in muscle mass, strength and function are inevitable developments in human adults. Nevertheless, what is the "normal age-appropriate" decline of muscle mass and function? Further, is this decline linear over the adult lifespan or are there changepoints of an accelerated loss of lower extremity muscle parameters?

Considering the essential effect of sex steroids GH, and IGF-I in muscle protein synthesis, the rapid menopausal reduction of both estrogens/testosterone (Veldhuis, 2008; Decaroli and Rochira, 2017) and GH/IGF-I (Sherlock and Toogood, 2007) suggests evidence for an accelerated decline of muscle mass and strength during women's early postmenopausal years (Maltais et al., 2009). However, (bioavailable) testosterone and corresponding declines in GH/IGF-I in men are much more linear (Harman et al., 2001; Veldhuis, 2008; Veldhuis et al., 2009; Decaroli and Rochira, 2017). Further, declines of muscle mass predominately affected by low serum concentration of anabolic agents did not consistently correlate with corresponding strength changes (Kim et al., 2018). Nevertheless the few data that focus on this issue (e.g., Larsson et al., 1979; Frontera et al., 1991; Lindle et al., 1997; Akima et al., 2001; Harbo et al., 2012) point to periods of accelerated strength decline of lower leg strength indices also in men.

However, there is no consensus as to when accelerated muscle strength decline starts. While most researchers (Larsson et al., 1979; Frontera et al., 1991; Akima et al., 2001; Harbo et al., 2012) located the accelerated decline in the mid and late 50ies, one author applied a more sophisticated statistical approach (Lindle et al., 1997) that led him to suggest a much earlier start of accelerated strength loss (40ies).

[1]ICD: International Classification of Diseases and Related Health Problems; CM: Clinical Modification.

[2]…..considering its relevance for resting metabolic rate and thermoregulation (Kenney and Buskirk, 1995; Kim et al., 2014).

[3]Greek origin: "dynamae" (δύναμη), c.f. dyna*mo*meter not dynameter.

One may argue that assessing the amount and time pattern of lower extremity strength declines in adults is a somewhat academic exercise. We do not accept that idea because in actual fact normative data of strength changes and potential accelerated strength decline changepoints form the basis for clinical decisions and corresponding therapeutic approaches. This might in particular be the springboard for more reliable and sophisticated strength testing which otherwise suffer from a lack of normative data.

The aim of the present contribution was thus to determine the decline in maximum isokinetic hip/leg extensor (MIES) and flexor strength (MIFS) as determined by an isokinetic leg press during adulthood. Based on the research question discussed above we focus on a male cohort 19–91 years old. Our primary hypothesis was that there is a linear decrease of MIES up to ≈age 60 years with a significant acceleration of decline after this "changepoint." Our secondary hypothesis was that there is a linear decrease of MIFS up to ≈age 60 years with a significant acceleration of decline after this "changepoint." Lastly, an experimental hypothesis was that adjusted for lean body mass (LBM) both MIES and MIFS loss their "significant changepoint."

METHODS

We used isokinetic leg-press data from several previous and ongoing cross-sectional and longitudinal[4] projects (e.g., Kemmler et al., 2010, 2014, 2016a, 2017) with men of different ages to evaluate the present research issue. All the studies were conducted between February 2008 and May 2018 by the Institute of Medical Physics (IMP), Friedrich-Alexander University (FAU) of Erlangen-Nürnberg. All the projects were approved by the university ethics committee of the FAU, Germany and fully complied with the Helsinki Declaration "Ethical Principles for Medical Research Involving Human Subjects." After detailed information, all the study participants gave their written informed consent. This also refer to the person in **Figure 1**, who provide written informed consent for his image to be published.

Outcomes

Primary study-endpoint

- Maximum dynamic strength (peak torque) of the leg/hip extensors (MIES).

Secondary study-endpoints

- Maximum dynamic strength (peak torque) of the leg/hip flexors (MIFS).

Participants

Altogether 362 community dwelling men 19–91 years old were included in the present analysis. Inclusion criteria were (1) male gender, (2) legal age, (3) community dwelling (cdw), i.e., independently living in the community. Subject were excluded when (1) taking medication (e.g., glucocorticoids >5 mg/d) or suffering from diseases with relevant impact on

[4]We used only the baseline data of these studies.

muscle metabolism, (2) conducting intense athletic performance (≥3 sessions/week/year), (3) diagnosed as having sarcopenia according to EWGSOP (Cruz-Jentoft et al., 2010), and (4) demonstrating low test compliance or unable to properly perform the isokinetic strength test (**Table 1**).

Assessments

Body-height was measured by a Harpender stadiometer (Holtain, Crosswell, UK); body mass and composition was determined via direct-segmental, multi-frequency Bio-Impedance-Analysis (DSM-BIA; InBody 230/770, Seoul, Korea) applying standardized protocols. BMI was calculated body mass/square body height; skeletal muscle mass index (SMI) was calculated according to Baumgartner et al. (1998) (i.e., appendicular skeletal muscle mass/square body height). Lean body mass was defined as fat-free body mass. Body fat as listed in **Table 1** refers to the fat rate for the total body. In order to standardize the BIA assessment, we consistently use the same BIA test protocol that includes minor physical activity for 8 h and 15 min of rest in a supine position immediately before the BIA assessment. Further, all participants were provided with written specifications about dos and don'ts including basic nutritional guidance 24 before testing. Baseline characteristics including physical activity and exercise were determined by questionnaires. For details, the reader is kindly referred to another publication (Kemmler et al., 2004).

FIGURE 1 | Leg press test (hip/leg extensor, hip/leg flexor strength) conducted on an isokinetic device (CON-TREX LP, Physiomed, Laipersdorf, Germany).

Maximum isokinetic strength of the leg and hip extensors/flexors was tested using an isokinetic leg press (CON-TREX LP, Physiomed, Laipersdorf, Germany) (**Figure 1**). Bilateral hip/leg extension and flexion was performed in a sitting, slightly supine position (15°), supported by hip and chest straps. Range of motion was selected between 30 to 90° of the knee angle, with the ankle flexed 90° and feet firmly fixed with straps positioned on a flexible sliding footplate. The standard default setting of 0.5 m/s was used. We consistently used our standard test specifications for all cohorts. Starting with a 5 min warm up on a cycling ergometer, one familiarization trial (5 repetitions) with the dedicated movement pattern ("push and pull"), and 3 min of rest, participants were then asked to conduct five repetitions with maximum voluntary effort. Participants conducted two trials intermitted by 2 min of rest. We consistently included the highest value for hip/leg extension and hip/leg flexion of the five repetitions and both trials in the data analysis. Hip/leg flexor (MIFS)/hip/leg extensor isokinetic strength (MIES)-Index (MIFS/MIES-Index) was calculated as MIFS divided by MIES. Reliability for the

TABLE 1 | Participant characteristics of 362 community dwelling men 19–91 years old structured in ranges of 15 years.

Variable	<35 yrs	35–49 yrs	50–64 yrs	65–79 yrs	80 yrs+
N	70	121	51	75	29
Age [years]	27.4 ± 5.2	43.6 ± 4.0	55.6 ± 4.7	71.4 ± 3.2	83.6 ± 3.0
Body height [cm]	182.5 ± 7.7	180.3 ± 7.0	180.8 ± 5.6	171.0 ± 5.6	170.0 ± 5.9
Body mass [kg]	82.4 ± 11.8	87.8 ± 13.8	88.4 ± 13.2	78.4 ± 10.1	74.5 ± 9.1
BMI [kg/m^2]	24.7 ± 3.7	27.0 ± 3.9	27.0 ± 3.6	26.8 ± 2.7	25.8 ± 2.5
Lean body mass [kg]	66.9 ± 8.9	67.5 ± 8.3	65.9 ± 8.3	58.8 ± 6.9	56.6 ± 4.7
SMI [kg/m^2]	8.68 ± 0.83	8.70 ± 0.90	8.52 ± 0.77	8.13 ± 0.70	7.96 ± 0.61
Body fat [%]	18.8 ± 4.1	23.1 ± 5.0	25.4 ± 4.9	25.0 ± 4.6	24.0 ± 4.8
Smokers [%/group]	13%	9%	10%	7%	7%
Activity Index[a]	3.3 ± 1.2	3.2 ± 1.3	3.4 ± 1.5	4.0 ± 1.3	3.9 ± 1.5
Exercise [min/week]	71 ± 48	51 ± 41	42 ± 37	35 ± 38	27 ± 36
MIES [N]	3,553 ± 584	3,373 ± 595	3,144 ± 621	2,141 ± 553	1,767 ± 425
MIES/LBM [N/kg]	53.2 ± 10.5	49.9 ± 9.0	47.7 ± 8.9	36.4 ± 8.7	31.2 ± 7.5
MIFS [N]	1,911 ± 372	1,593 ±325	1,506 ± 358	948 ± 270	729 ± 177
MIFS/LBM [N/kg]	28.6 ± 6.2	23.6 ± 5.0	22.9 ± 5.5	16.1 ± 4.2	12.9 ± 3.5
MIFS/MIES-Index	0.538 ± 0.080	0.472 ± 0.083	0.479 ± 0.102	0.443 ± 0.089	0.413 ± 0.70

BMI, Body Mass Index; SMI, Skeletal muscle mass index according to Baumgartner et al. (1998); i.e., appendicular skeletal muscle mass/body height2. MIES, maximum isokinetic hip/leg extensor strength; MIFS, maximum isokinetic hip/leg flexor strength.
[a]*Scale from (1) very low to (7) very high (Kemmler et al., 2004).*

maximum hip/leg extensor strength (Test-Retest-Reliability; Intra Class Correlation) was 0.88 (95%-CI: 0.82–0.93) for a male cohort 30–50 years old. The same test assessor responsible for the leg-press assessments conducted most of the tests.

Statistical Analysis

Based on a statistically (Shapiro-Wilk test) and graphically (Q-Q plots) checked normal distribution, the outcomes presented in **Table 1** were reported using mean values (MV) and standard deviation (SD). We abstained from a sophisticated statistical procedure for participant characterics structured according to age (**Table 1**). However, we statistically addressed the issue of overall strength changes during adulthood for MIES and MIFS comparing the oldest with the youngest subgroup using pairwise (independent) *t*-tests and Cohen's d effect sizes. Although not necessarily specified for subordinate study endpoints, *t*-tests were adjusted for multiple testing using the Bonferroni procedure. The statistical procedures listed above were performed using SPSS Statistics version 25.

Furthermore, we performed a multiple regression analysis with change point estimation using the statistical software R in combination with package segmented (Muggeo, 2008). Dependent variables within the multiple regression were MIES and MIFS while age, body height, body mass, ASMM[5], and exercise (**Table 1**) served as independent variables. The significance of the change points was assessed by the score-test proposed by Muggeo (Muggeo, 2016) and additionally verified by Davies' test (Davies, 2002). All tests were 2-tailed, significance was accepted at p <0.05.

RESULTS

Table 1 gives the characteristics of the 362 participants 19–91 years old included in the analysis. As evident from **Table 1**, the age groups were not equally distributed ($p < 0.001$); the lowest sample size was generated for men 80 years+ ($n = 29$). However, more crucially, there was a relatively small number of study participants ($n = 47$) between 51 and 69 years[6].

Further, **Table 1** gives strength indices determined for different age groups. In summary, maximum isokinetic strength of the hip/leg extension (MIES) and hip/leg flexion (MIFS) strength as determined by a closed kinetic system at least halved (MIES: −50 ± 21%, MIFS: −62 ±2 4%) during <35 to ≥80 years (adjusted $p \leq 0.001$; ES, d': 3.50 and 2.95, respectively).

Adjusting MIES for LBM (**Table 1**) or BMI (<35 years: 143.9 ± 30.3 vs. ≥80 years: 68.4 ± 17.6 N/[kg/m^2]) resulted in comparable steep declines (MIES/LBM: 41 ± 15%, ES d' = 2.42; MIES/BMI: 52 ± 22%; ES d' = 3.05).

Correspondingly, MIFS adjusted for BMI decreased from 77.4 ± 17.3 N/[kg/m^2] at age <35 years to 28.3 ± 7.8 N/[kg/m^2] at age ≥80 years; i.e., by −63 ± 25% (ES d'=3.66). In parallel MIFS/LBM (**Table 1**) decreased by 55±24% (ES d'= 3.12).

[5]Appendicular skeletal muscle mass i.e. muscle mass of the limbs. ASMM is the crucial muscle mass parameter within the sarcopenia criteria.

[6]i.e. in the range of our expected changes points.

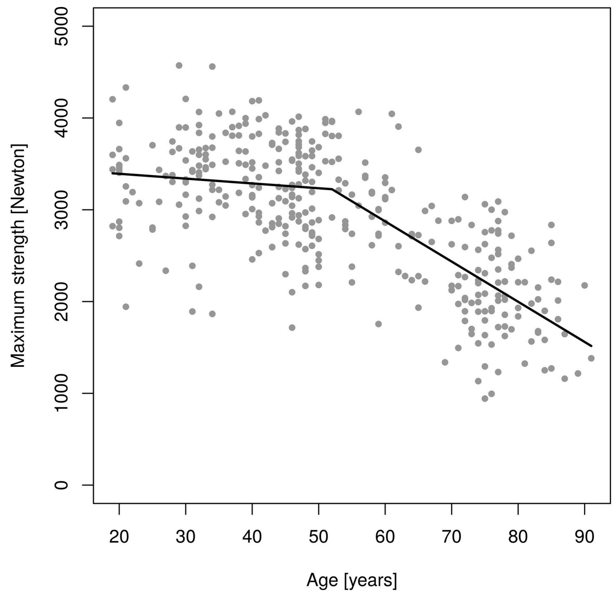

FIGURE 2 | Changes of maximum isokinetic hip/leg extensor strength (MIES) during adulthood as determined by an isokinetic leg-press.

Due to the more pronounced decline of MIFS vs. MIES the corresponding index also declined over the adults' age span (adjusted $p < 0.011$; ES, d' = 1.66).

Figures 2, **3** shows results of the "changepoint analysis" for MIES (**Figure 2**) and MIFS (**Figure 3**) after multiple regression analysis adjusted for the co-variables "age," "body-height," "body-mass," "ASMM"[7] and "exercise." The latter model explained 61% each of the variance of MIES (r^2: 0.611) and MIFS (r^2: 0.605). In summary, MIES demonstrates a non-significant reduction of 5.2 N/p.a. (95% CI: 3.2 to −13.7 N/p.a.; i.e., ≈0.15% p.a.) up to the estimated breakpoint of 52.0 years ($p < 0.001$) and an accelerated loss of ≈44 N/p.a. (95% CI: −34.5 to −53.1 N/p.a.; i.e., ≈1.3% p.a.) after this changepoint. Of importance, the more conservative Davies-Test (Davies, 2002) confirmed ($p < 0.001$) the significant changepoint[8] at age 52 years; a second significant changepoint was not determined. Thus, we have to revise our primary hypothesis that there is a linear decrease of MIES up to age ≈60 years with a significant acceleration of decline after this "changepoint."

In parallel to MIES, only one significant changepoint ($p = 0.001$) was determined for MIFS (**Figure 3**). However, in contrast to MIES the decline of MIFS prior to the breakpoint at age 59.0 years was already significant. Correspondingly, younger men lost 10.0 N/p.a. (−5.8 to −14.2 N/p.a.; i.e., ≈0.5% p.a.), while the decline in MIFS in men 60 years and older averaged 25.0 N/p.a. (−6.2 to −32.9 N/p.a.; i.e., ≈1.3% p.a.). The Davies-Test confirmed ($p < 0.004$) the significant changepoint at age 59 years. Thus, we confirmed our hypothesis that there is a linear decrease of MIES up to ≈age 60 years with a significant acceleration of

[7]Appendicular skeletal muscle mass.

[8]i.e., the point at which the significantly accelerated decline started.

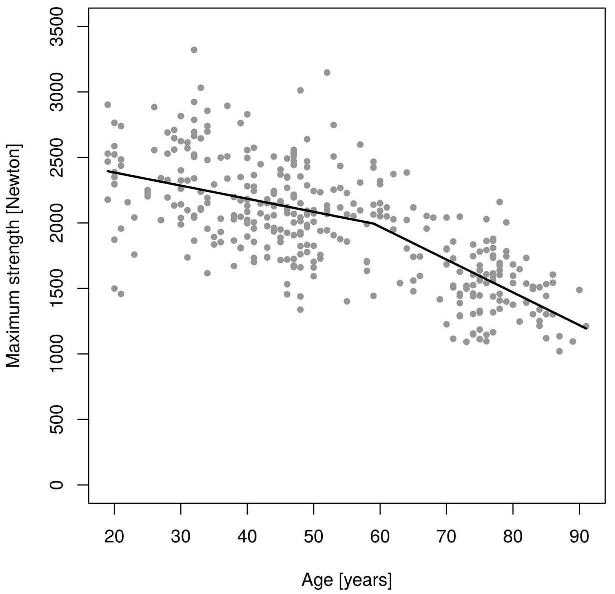

FIGURE 3 | Changes of maximum isokinetic hip/leg flexor strength (MIFS) during adulthood as determined by an isokinetic leg-press.

decline after this "changepoint." Correspondingly, we confirmed our secondary hypothesis that there is a linear decrease of MIES up to ≈age 60 years with a significant acceleration of decline after this "changepoint."

Applying the same statistical procedures for MIES and MIFS, we are unable to observe changepoints for a significant decline ($p > 0.100$) after adjusting for lean body mass (i.e., N/kg LBM) for either MIES or MIFS. Thus, we confirmed our experimental hypothesis I that adjusted for LBM both MIES and MIFS lost their "significant changepoint."

DISCUSSION

In this article, we particularly aimed to determine the amount and progress of age-dependent reductions of maximum lower limb strength in community-dwelling men. Our most striking motivation for focusing on hip/leg extension (and flexion) strength was the particular relevance of leg-extension strength on mobility limitations, disability, morbidity, and mortality in older people (Visser et al., 2005; Newman et al., 2006; Roshanravan et al., 2017). Of importance, the present study is the first to provide data on maximum hip/leg extensor and flexor strength over a wide range of male adulthood (19–91 years) using an isokinetic leg press. We consider the approach of applying a closed chain kinetic system important, since these tests might determine functional aspects of lower limb performance in more depth (Augustsson and Thomeé, 2000).

In summary, we determined an average yearly reduction of ≈0.8% p.a. for MIES and ≈1.0% p.a. for MIFS. However, we observed a significant acceleration of decline in MIES that started earlier (52 years) than expected and varied from the later significant decline in MIFS (59 years).

Although we have to admit that it might be inadequate to compare our results with open isokinetic chain testing (Ferraresi et al., 2013), we generally confirmed the course and volume of declines of knee extensor and flexor strength during male adulthood (e.g., Larsson et al., 1979; Frontera et al., 1991; Poulin et al., 1992; Porter et al., 1995; Lindle et al., 1997; Neder et al., 1999; Akima et al., 2001; Harbo et al., 2012; Cheng et al., 2014).

In detail, annual changes in concentric MIES/MIFS reported by studies that assessed men between the age of 20 and 70–80 (e.g., Larsson et al., 1979; Poulin et al., 1992; Porter et al., 1995; Lindle et al., 1997; Neder et al., 1999; Akima et al., 2001; Harbo et al., 2012) averaged between 0.5% p.a. (Lindle et al., 1997) and 1.2% p.a. (Akima et al., 2001). Closest to our results, Akima et al. (2001) reported age-related declines in maximum isokinetic strength of 1.2% p.a. for MIES and 1.1% p.a. for MIFS for their cohort of 100 Japanese men 20–84 years old.

Although it is difficult to compare cross-sectional with longitudinal study results, it might be worthwhile to briefly address age-related changes of lower extremity muscle strength in older people (e.g., Frontera et al., 2000; Goodpaster et al., 2006; Lauretani et al., 2008; Dey et al., 2009; Koster et al., 2011; Roshanravan et al., 2017). In summary, with few exceptions (i.e., Lauretani et al., 2008)[9] average maximum strength reductions range around 3.0–3.5% p.a. (Goodpaster et al., 2006; Koster et al., 2011) for concentric KES and from 3.3% p.a. (Hicks et al., 2012) up to 9% p.a. for isometric KES (Dey et al., 2009). Corresponding maximum strength declines at advanced age observed in the present study were lower (MIES: 2.0–2.5% p.a.). However, apart from the cross-sectional study design and differences in strength assessment (i.e., open / closed kinetic chain), the most striking difference was that we focus exclusively on community dwelling (CDW) men. This approach might constitute a relevant bias, since people with very low MIES might be unable to "live independently in the community." Thus, we might have excluded a cohort with (very) low maximum strength and consequently gained lower declines in MIES at older age. However, due to the lack of information, we are not convinced whether all the longitudinal studies account for this aspect.

Revisiting changepoints for accelerated strength loss of the lower extremities, we confirmed the results of Akima et al. (2001), Harbo et al. (2012) and Larsson et al. (1979). Although none of these researchers conducted a dedicated changepoint analysis, they found that the most pronounced decline of isokinetic and isometric knee extensor strength may well-occur in the mid to end 50ies. However, using regression analysis, Lindle et al. (1997) reported an earlier (fourth decade) start of accelerated decline in concentric or eccentric peak torque of the knee extensors in men and women.

Addressing the later changepoint (52 vs. 59 yrs.) for accelerated strength loss in MIES vs. MIFS, we are unable to provide a meaningful physiologic explanation for this feature. For want of corresponding data in the literature, we can only hazard

[9]However, the 6-year follow-up of the study (Lauretani et al., 2008) reported only an overall 0.8 kg (5.1%) decline in isometric KES.

as an explanation that the later decline of MIFS compared with MIES might be related to the more pronounced decline before and less pronounced decline after the MIFS changepoint.

We observed largely parallel changes of LBM and MIES/MIFS (**Table 1**). Thus, muscle strength declines can be attributed to a considerable extent to declines in muscle mass parameters (Akima et al., 2001), although this relationship might differ between races, at least in men (Araujo et al., 2010). However, relative changes of muscle mass parameters were much lesser pronounced (i.e., LBM: 15 ± 13%) compared with strength changes (MIES: −50 ± 21%, MIFS: −62 ± 24%) an aspect that might explain that adjusting MIES/MIFS for LBM resulted in a more linear decrease of strength declines and a loss of the significant changepoints for MIES and MIFS. However, the high relevance of muscle mass parameters for functional outcomes (here: "strength") in older individuals is also confirmed by the study of Akima et al. (2001) which concluded that muscle mass dimensions (here: CSA quadriceps femoris) are the "primary factor" involved in an aged individual's capacity to exert maximal force (isokinetic knee flexion and extension). Apart from this aspect, other factors not addressed in our project, e.g., nervous control, muscle recruitment, qualitative changes in contractile properties, etc. (review in Tieland et al., 2018) of course contribute relevantly to age-related changes of muscle strength.

In summary, using a closed kinetic chain testing procedure (i.e., isokinetic leg-press) we largely confirmed the average percentage loss of lower extremity muscle strength during adulthood. Further, and specifically addressed by our main hypothesis, we are in accord with most other studies that the changepoint of accelerated loss of hip/knee extensor and flexor strength is located between ages 50 and 60 years. So far, differences between open and closed kinetic chain-based testing with corresponding impact of functional performance were not remarkable.

However, our study features some particularities and limitations that may prevent a proper interpretation of our results and an adequate comparison with other studies in this field. (1) One may dispute the practical relevance of isokinetic testing in the context of sarcopenia/dynamopenia. While isokinetic movements may not represent a challenge in daily living, functional tests of the lower legs (i.e., squats, lunges) are more akin to daily activities. Reviewing the literature, correlations between isokinetic and functional testing vary from moderate (Augustsson and Thomeé, 2000) to high (e.g., Butcher et al., 2012; Gkrilias et al., 2018); the results do, however, depend on the specifications of the tests (e.g., °/s) and the cohorts tested. Further, most results were reported for isokinetic chairs, with their leg extension movement that differ considerably from the functionally more relevant closed kinetic chain exercise "leg-press." However, we finally opted to use an isokinetic test device for this research issue because of the higher degree of standardization that it offers. (2) Age groups were not equally populated and there is a relative lack of participants between 51 and 69 years old ($n = 47$), i.e., in the range of the expected changepoints of accelerated strength decline. (3) Although age-dependent BMI was inconspicuous and comparable to normative data for German men (Statistisches Bundesamt, 2012), there is a highly significant decline of body height between 50–64 and 65 years+. (4) Body fat rate of all the age groups is considerably lower (i.e., %-LBM was higher) than reported for German cohorts of comparable age and physical activity. Particularly in our youngest group, the body fat was quite low for a non-athletic population. However, based on our eligibility criteria (healthy, non-athletic, and cdw men, without medication known to affect muscle metabolism), in summary we consider our cohort as representative for the majority of the basic male population. Accordingly, we decided that the results of MIES, MIFS and MIFS/MIES-Index listed in **Table 1** can be widely considered as "normative data" for the corresponding age groups. (5) Most, but not all tests were consistently conducted by the same test assessor; further, we consistently applied our standardized test protocol across all studies. (6) Finally, we focus on cdw men, i.e., predominately men with a functional status that permits independent living in the community. Potentially, this might have generated a relevant bias, since institutionalized and correspondingly more functionally limited older men did not contribute to the result of our study. This estimation is confirmed by the rather high MIES and MIFS in our oldest cohort.

A limited amount of studies clearly focus on the issue of age-dependent strength declines. The present study adds further evidence that lower extremity strength decreases considerably during adult men's lives; however the more salient result was that this decline is far from linear. Summarizing the novel aspects and practical application of our study: (1) For the first time "normative values" of maximum isokinetic lower extremity strength have been provided for a closed isokinetic chain system. (2) Further, this is the first study to apply a sophisticated statistical approach to determine the precise changepoints for accelerated muscle loss for MIES and MIFS (3) Unlike most studies that focus on a narrow age range, the present study covers the full range of adulthood. (4) From a practical application point of view, our results provided further evidence that programs that focus on sarcopenia prophylaxis (…or at least its dynamopenic aspect) in men should be started in the 5th decade. Of importance, a related study that focused on trainability in different periods of life (Von Stengel and Kemmler, 2018)[10] reported a significant trainability of MIES and MIFS in all the age groups addressed here. This included our oldest cohort of men 80 years+, an observation confirmed by various longitudinal studies (review in Stewart et al., 2014).

In conclusion, due to the accelerated loss of muscle strength starting in the 50ies, health care programs that focus not only but specifically on the "dynamopenic" aspect of sarcopenia prevention should start early in men's lives. We speculate that due to the occupational and social situation of male subjects aged about 50 years characterized, inter alia, by a general lack of time[11], there might be a need for dedicated exercise programs for this cohort. Time-effective exercise

[10]The authors analyzed the longitudinal results on MIES and MIFS of studies that were also included in the present study (…however, baseline data only).

[11]…the reason predominately cited for not exercising (Rütten et al., 2009).

protocols [i.e., HIT-single set resistance exercise and/or whole-body electromyostimulation (Kemmler et al., 2016a,b)], ideally combined with adjuvant nutritional support (e.g. protein, BCAA), i.e., strategies recognized to increase muscle mass and strength (Tipton, 2008; Cermak et al., 2012) might be thus a feasible option for this cohort.

DATA AVAILABILITY

The anonymized data used to support the findings of this study are available from the corresponding author upon request.

AUTHOR CONTRIBUTIONS

SvS, DS, MK, and WK designed the study, completed data analysis and/or interpretation and drafted the manuscript. SvS, MK, DS, and WK contributed to study conception and design and revised the manuscript. WK accepts full responsibility for the integrity of the data sampling, analysis and interpretation.

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Force-Velocity Characteristics, Muscle Strength, and Flexibility in Female Recreational Marathon Runners

Pantelis Theodoros Nikolaidis[1,2], Thomas Rosemann[3] and Beat Knechtle[3,4]*

[1] Exercise Physiology Laboratory, Nikaia, Greece, [2] Laboratory of Exercise Testing, Hellenic Air Force Academy, Acharnes, Greece, [3] Institute of Primary Care, University of Zurich, Zurich, Switzerland, [4] Medbase St. Gallen Am Vadianplatz, St. Gallen, Switzerland

Physical fitness components that relate with performance in marathon running, e.g., aerobic capacity and body composition, have been studied extensively. On the other hand, data on components of the health-related physical fitness, such as flexibility and muscle strength, were missing in this sport. Therefore, the aim of the present study was to profile force-velocity (F-v) characteristics, muscle strength and flexibility in female recreational marathon runners and to examine their relationship with age, race time and anthropometric characteristics (body fat percentage, fat-free mass – FFM, and total thigh muscle cross-sectional area – CSA). Thirty three female marathon runners (age 40.0 ± 8.9 years, body fat percentage 19.5 ± 4.6% and personal record 4:34 ± 0:39 h:min), separated into three age groups (<35, 35–45 and >45 years) and three performance groups (race time <4:15 h:min, 4:15–4:45 h:min and >4:45 h:min), performed sit-and-reach test (SAR), isometric muscle strength tests, squat jump, countermovement jump and F-v test on a cycle ergometer. The main findings of the present study were that (i) participants had moderate scores of body composition and physical fitness considering norms of the general population, (ii) the <35 age group had better jumping ability than 35–45 and >45 age group, and the older age group had lower F_0, Pmax and rPmax than their younger counterparts, (iii) the slowest performance group scored the highest in SAR, and (iv) isometric strength, F_0 and Pmax correlated largely with body mass and FFM. Considering the lack of existing data on anaerobic power and neuromuscular fitness of female marathon runners, the findings reported in this study would be useful for strength and conditioning trainers to monitor the training of their athletes. Even if these parameters were not related to race time, they should be monitored regularly as they were either component of health-related physical fitness (muscle strength and flexibility) or could help runners (anaerobic power) under specific circumstances such as ascends during a race.

Keywords: aging, anthropometry, countermovement jump, cycle ergometer, handgrip strength, sit-and-reach test, squat jump

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***Correspondence:**
Beat Knechtle
beat.knechtle@hispeed.ch

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INTRODUCTION

During the last decades, an increased number of female runners participate in marathon races. For instance, the male-to-female runners ratio of finishers in the "New York City Marathon" decreased from 2.1 in 2006 to 1.4 in 2016 (Nikolaidis et al., 2018). Accordingly, an increased scientific interest has evolved for female marathon runners. With regards to their physiological characteristics, most studies focused on maximal oxygen uptake (VO_2max), anaerobic threshold and running economy (Helgerud et al., 1990; Daniels and Daniels, 1992; Pate and O'Neill, 2007). An explanation of the large number of studies focusing on these characteristics might be their correlations with endurance performance and marathon race time (Williams and Cavanagh, 1987; Midgley et al., 2007) and the role of aerobic capacity for health (Lee et al., 2010, 2011). On the other hand, anaerobic power or major components of health-related physical fitness, such as muscle strength and flexibility (Pate, 1983; Caspersen et al., 1985; Heyward and Gibson, 2014), have been rarely investigated in female marathon runners. Although some of the health-related physical fitness components (i.e., muscle strength and flexibility) might not be determinants of race time in marathon running, these components would allow humans performing daily physical activity with vigor (Cattuzzo et al., 2016). In addition, an optimal health-related physical fitness might play a key role for quality of life and successful aging (Lee and Tanaka, 1997).

Anaerobic power may be necessary in ascends or descends, or in situations such as to overcome rapidly an obstacle or an opponent during a marathon race, e.g., it has been shown that running in an augmented slope resulted in increased step frequency, ground reaction force and metabolic cost (Padulo et al., 2013). It has been shown that anaerobic power in elite marathon runners is relatively low compared with runners of shorter distances (Vuorimaa, 1996; Legaz-Arrese et al., 2011). For example, the results of a 20 s maximal anaerobic running test showed lower scores in elite marathon runners than sprinters and middle distance runners (Vuorimaa, 1996). Moreover, a comparative study of various running distances revealed that elite marathon runners had the lowest scores in the Wingate anaerobic test (WAnT) compared to their peers participating in shorter distances (Legaz-Arrese et al., 2011). So far, the force-velocity (F-v) test (Driss and Vandewalle, 2013), developed to characterize limits of the neuromuscular system to produce power (Cross et al., 2017), has been widely used in athletes such as judokas, boxers, taekwondo athletes (Busko, 2016), team handball (Nikolaidis et al., 2016), soccer (Nikolaïdis, 2012), tennis (Durand et al., 2010), and cyclists (Nikolaidis and Papadopoulos, 2011), but not in endurance athletes. Compared to the WAnT that evaluates anaerobic power, the F-v test provides additional information about the two constituents (i.e., force and velocity) of anaerobic power; thus, it can identify potential "weak" constituent that should be targeted for optimization (Driss and Vandewalle, 2013).

Few studies have even reported isometric muscle strength (e.g., handgrip) and jump ability (e.g., countermovement jump – CMJ) of female marathon runners in designs that used small number of female participants ($n \leq 6$) (Del Coso et al., 2013; Piacentini et al., 2013). In addition, the abovementioned studies did not distinguish scores between sexes; thus, we have no knowledge about performance of female runners in these tests. CMJ has been shown to differentiate among male runners of various distances (e.g., marathon versus middle-distance versus sprinters) (Vuorimaa, 1996); thus, it could provide useful insight of practical relevance for female marathon runners. In addition, handgrip muscle strength is related to health; for instance, it has been shown that higher level of this physical fitness component is associated with a reduced risk of all-cause mortality, especially in women (Garcia-Hermoso et al., 2018).

With regards to flexibility, an optimal level of this parameter is necessary for health; for example, the flexibility of the muscle, tendons and ligaments in the back might be associated with the range of motion and functional movement (Gordon and Bloxham, 2016). A study on international level male distance runners showed that the least flexible runners were also the most economical in terms of running economy (Jones, 2002) indicating a concern that chronic endurance training could induce a decreased flexibility. However, these findings remain to be confirmed in female marathon runners. Although the abovementioned studies improved our understanding of the demands of marathon running for anaerobic power, muscle strength and flexibility in female runners, no information exists concerning relatively large samples of female recreational marathon runners, especially with regards to the variation of these characteristics with age, performance, and anthropometry. Such information would be of great practical value for strength and conditioning coaches working with female marathon runners to evaluate the fitness level of their athletes and develop specific training programs. Therefore, the main aim of the present study was to profile F-v characteristics, isometric muscle strength, jump ability, and flexibility of female recreational marathon runners and examine their relationship with age, performance and anthropometry.

MATERIALS AND METHODS

Experimental Approach to the Problem

To examine the relationship of F-v characteristics, isometric muscle strength, jump ability and flexibility with age, performance and anthropometry, a cross-sectional study design was applied. Participants ($n = 33$) were divided into three age groups (<35, 35–45 and >45 years) to study the effect of age on these performance parameters. Also, differences among three even performance groups, based on the best race time (fast 3:53 ± 0:19 h:min, <4:15 h:min, $n = 11$; medium 4:30 ± 0:07 h:min, 4:15–4:45 h:min, $n = 11$; slow 5:14 ± 0:30 h:min, >4:45 h:min, $n = 11$), were examined. The best race time was considered instead of the most recent race time as the latter could be influenced by non-performance conditions, e.g., injury. The relationship of physical fitness (sit-and-reach test – SAR, squat jump – SJ, CMJ, isometric strength, and F-v test) with anthropometric characteristics (body mass, body mass index – BMI, body fat percentage – BF, total thigh muscle cross-sectional area – CSA, and fat-free mass – FFM) was examined using correlations.

Subjects

Thirty-three female recreational marathon runners (age 40.0 ± 8.9 years, height 162 ± 6 cm, body mass 57.7 ± 7.4 kg, BMI 21.8 ± 2.1 kg.m^{-2} and personal record 4:34 ± 0:39 h:min, completed marathons in the past 3.3 ± 3.6) mostly from the area of Athens volunteered to participate in this study, which had been advertised through popular websites for endurance runners. The participants reported sport experience 5.6 ± 4.6 years and were practicing endurance training for 4.1 ± 1.5 days weekly with each training session corresponding to 9.3 ± 2.8 km or 1.6 ± 0.5 h. During September and October 2017, the participants visited the laboratory where they were examined for anthropometric characteristics and performed a F-v test, isometric muscle strength, jump ability and flexibility. This study was approved by the local Institutional Review Board (Exercise Physiology Laboratory, Nikaia). The study was conducted in accordance with the Declaration of Helsinki. All participants gave written informed consent after having been provided detailed information about the risks and benefits of the research.

Procedures

Anthropometry

Height, body mass, and skinfolds were measured with participants in minimal clothing and barefoot. An electronic weighing scale (HD-351; Tanita, Arlington Heights, IL, United States) was employed for measurement of body mass (to the nearest 0.1 kg), a portable stadiometer (SECA Leicester, United Kingdom) for height (0.001 m), and a caliper (Harpenden, West Sussex, United Kingdom) for skinfolds (0.2 mm). BMI was calculated as the quotient of body mass (kg) to height squared (m^2), and BF was estimated from the sum of 10 skinfolds, i.e., cheek, wattle, chest I, triceps, subscapular, abdominal, chest II, suprailiac, thigh, and calf (Parizkova, 1978). FFM in kg was calculated as "body mass - (body mass * BF/100)." CSA was calculated as "(4.68 * midthigh circumference in cm) – (2.09 × anterior thigh skinfold in mm) – 80.99" (Housh et al., 1995).

Sit-and-Reach Test

The sit-and-reach test (SAR) was used to assess low back and hamstring flexibility (Mayorga-Vega et al., 2014). It was performed on a box providing 15 cm advantage, i.e., the participants scores 15 cm when they reach their toes. Two trials were performed with 1 min break between trials and the best score was recorded to the nearest 0.5 cm. Intra-class correlation coefficient (ICC) of single measures was 0.981 (95% confidence intervals, CI, 0.975; 0.986).

Isometric Muscle Strength Tests

To evaluate isometric muscle strength, the sum of four tests (right and left handgrip test, back test, back-and-leg test) in absolute and relative to body mass values was used. The handgrip test was performed with participants standing and having their elbow flexed at approximately 90°. They were instructed to squeeze the handle of the handgrip dynamometer (Takei, Tokyo, Japan) as hard as possible for 5 s. Two trials were performed for each hand, with a 1 min rest between trials. The best trial was recorded for each hand (Heyward, 2010). ICC was 0.945 (95% CI, 0.926; 0.959) in the both hands. A back strength dynamometer (Takei, Tokyo, Japan) was used for both back strength test and back-and-leg test (test-retest ICC 0.92) (Ten Hoor et al., 2016). The back strength test was performed with participants having their legs and backs straightened to allow the bar to level with the patella, while in the combined back-and-leg test, the chain length on the dynamometer was adjusted so that the participants squatted over the dynamometer with their knees flexed at approximately 30° (Heyward, 2010). All measurements were recorded to the nearest 0.1 kg.

Jumping Tests

The participants performed two trials for each jumping test (squat jump, SJ, and countermovement jump, CMJ) and the best result was recorded (Aragon-Vargas, 2000). There was 1 min break between trials and tests. Height of each jump was estimated using the Opto-jump (Microgate Engineering, Bolzano, Italy) and was expressed to the nearest 0.1 cm. ICC was 0.914 (95% CI, 0.885; 0.936) in SJ and 0.951 (95% CI, 0.934; 0.963) in CMJ.

Force-Velocity Test

The F-v test was used to assess Pmax, expressed as W and as W·kg^{-1} (rPmax), theoretical maximal velocity (v_0) in revolutions per minute (rpm) and force (F_0) in N, and v_0/F_0 was calculated in rpm.N^{-1}. The participants performed four sprints, each one lasting 7 s, against braking force (2, 3, 4, and 5 kg on a counterbalanced order) on a leg cycle ergometer (Ergomedics 874E, Monark, Sweden), interspersed by 5 min recovery periods. The seat height of the ergometer was adjusted to allow for a slight bend in the knee (approximately 175°) and in accordance with the participant's satisfaction. Each sprint began with a flying start, i.e., as soon as velocity reached 50 rpm (revolutions per minute), the weight basket was released and the braking force was applied. For each participant an individual linear regression was determined between peak velocity and braking force for each of the four sprints. F_0 and v_0 corresponded to the intercepts with F and v axes in the F-v graph. Pmax was calculated as $Pmax = 0.25 \cdot F_0 \cdot v_0$ (Vandewalle et al., 1985).

Statistical Analyses

Statistical analyses were performed using IBM SPSS v.20.0 (SPSS, Chicago, IL, United States) and GraphPad Prism v. 7.0 (GraphPad Software, San Diego, CA, United States). Normality was examined using Kolmogorov-Smirnov test and visual inspection of normal Q-Q plots. Data were expressed as mean and standard deviation (SD). One-way repeated measures analysis of variance (ANOVA) and a subsequent Bonferroni *post hoc* test (if there were differences among groups) were used to examine the differences among age and performance groups, separately. To interpret effect size (ES) for statistical differences in the ANOVA, partial eta square classified as small ($0.010 < \eta_p^2 \leq 0.059$), medium ($0.059 < \eta_p^2 \leq 0.138$), and large ($\eta_p^2 > 0.138$) was used (Cohen, 1988). The relationship of flexibility, isometric muscle strength, jumping ability and F-v characteristics with age, performance and anthropometry was examined using Pearson's product moment correlation coefficient (r). Magnitude of correlation coefficients was considered as trivial if $r < 0.10$, small if $0.10 \leq r < 0.30$, moderate if $0.30 \leq r < 0.50$, large if $0.50 \leq r < 0.70$, very large if $0.70 \leq r < 0.90$, nearly perfect if $r \geq 0.90$, and perfect if $r = 1.00$ (Batterham and Hopkins, 2006). The level of significance was set at $\alpha = 0.05$.

RESULTS

Profile

The anthropometric characteristics of participants, in total and by age group, can be seen in **Table 1**. In the F-v test, v_0 of all participants was 167 ± 15 rpm (ranging from 132 to 195 rpm), F_0 120 ± 20N (93–179 N), Pmax 507 ± 85W (340–704 W), rPmax 8.83 ± 1.17 W.kg^{-1} (6.0–11.1 W.kg^{-1}) and $v_0.F_0^{-1}$ 1.43 ± 0.28 rpm.N^{-1} (0.87–1.93 rpm.N^{-1}). With regards to neuromuscular fitness, SAR was 25.8 ± 8.3 cm (8.0–37.5 cm), SJ 17.7 ± 3.4 cm (10.3–25.2 cm) and CMJ 18.6 ± 3.7 cm (11.2–26.1 cm). In isometric muscle strength, right handgrip was 29.7 ± 4.5 kg (20.7–41.6 kg), left handgrip 29.7 ± 4.0 kg (23.4–39.7 kg), back 82.5 ± 16.2 kg (40.0–114.0 kg), back-and-leg 94.9 ± 19.0 kg (48.5–150.0 kg), sum 236.9 ± 40.1 kg (134.2–339.3 kg) and 4.12 ± 0.56 kg.kg^{-1} (2.90–5.23 kg.kg^{-1}).

TABLE 1 | Comparison of anthropometric characteristics among age groups.

	Total	Age groups		
		<35 years	35–45 years	>45 years
n	33	10	13	10
Age (years)*	40.0 ± 8.9	29.2 ± 4.5	41.0 ± 2.2	49.3 ± 5.3
Height (cm)	162 ± 6	161 ± 6	163 ± 6	163 ± 7
Body mass (kg)	57.7 ± 7.4	53.7 ± 6.9	58.5 ± 6.5	60.5 ± 7.8
BMI (kg.m^{-2})	21.8 ± 2.1	20.7 ± 1.6	22.0 ± 1.6	22.7 ± 2.8
BF (%)	19.5 ± 4.6	17.3 ± 5.0	19.5 ± 3.5	21.8 ± 4.9
FFM (kg)	46.3 ± 5.2	44.4 ± 6.0	47.0 ± 4.3	47.2 ± 5.3
CSA (cm^2)	118 ± 17	119 ± 14	120 ± 16	114 ± 21

*BMI, body mass index; BF, body fat percentage; FFM, fat-free mass; CSA, thigh muscle cross-sectional area. *Except age for which all age groups differed at $p < 0.001$, no difference was observed in any anthropometric characteristic.*

Age

A large main effect of age on SJ ($p = 0.002$, $\eta_p^2 = 0.361$) and CMJ ($p = 0.001$, $\eta_p^2 = 0.384$) was observed with higher values in the <35 age group than in the 35–45 and >45 age group, whereas no difference was shown in SAR ($p = 0.912$, $\eta_p^2 = 0.006$) (**Figure 1**). No difference was found in isometric muscle strength (**Figure 2**). With regards to F-v characteristics, a large main effect was observed on v_0 ($p = 0.034$, $\eta_p^2 = 0.208$), F_0 ($p = 0.004$, $\eta_p^2 = 0.323$) with higher score in 35–45 than <35 and >45 age group, Pmax ($p = 0.037$, $\eta_p^2 = 0.203$) with higher score in 35–45 than >45 age group, rPmax ($p < 0.001$, $\eta_p^2 = 0.468$) with higher score in <35 and 35–45 than >45 age group and v_0/F_0 ($p = 0.002$, $\eta_p^2 = 0.354$) with higher score in <35 than 35–45 age group (**Figure 3**).

Anthropometry and Physical Fitness by Performance Group

Anthropometric characteristics and physical fitness by performance level can be seen in **Table 2**. A large main effect of performance on flexibility was observed ($p = 0.039$, $\eta_p^2 = 0.201$) with higher score (+8.3 cm) in the slowest than in the average performance group.

Anthropometry

SJ and the sum of isometric strength tests (in relative values) correlated inversely moderately with BF (**Table 3**). All the other indices of isometric strength correlated largely with body mass, FFM and CSA. It should be highlighted that the magnitude of these correlations with FFM was larger than body mass. Vo and rPmax did not correlate with any anthropometric characteristic, whereas F0 and Pmax correlated largely with body mass and FFM.

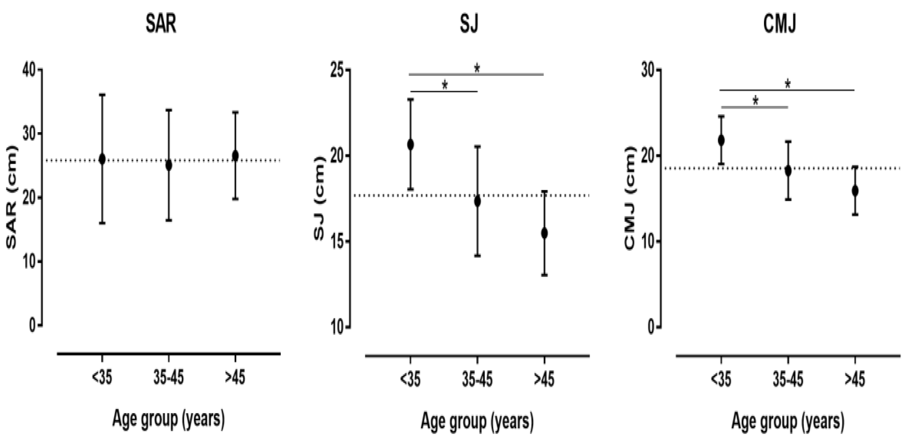

FIGURE 1 | Differences in sit-and-reach test (SAR), squat jump (SJ) and countermovement jump (CMJ) among age groups. * Difference at $p < 0.05$.

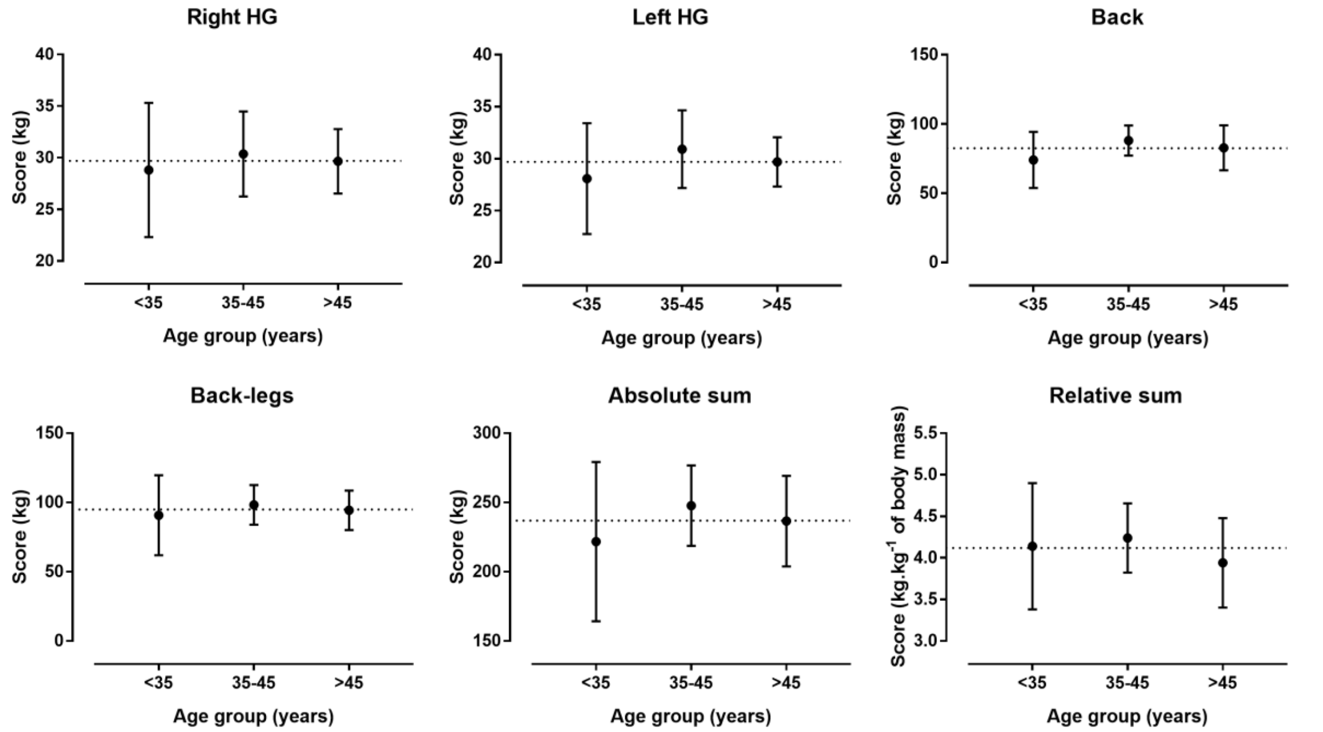

FIGURE 2 | Differences in isometric muscle strength among age groups. * Difference at $p < 0.05$.

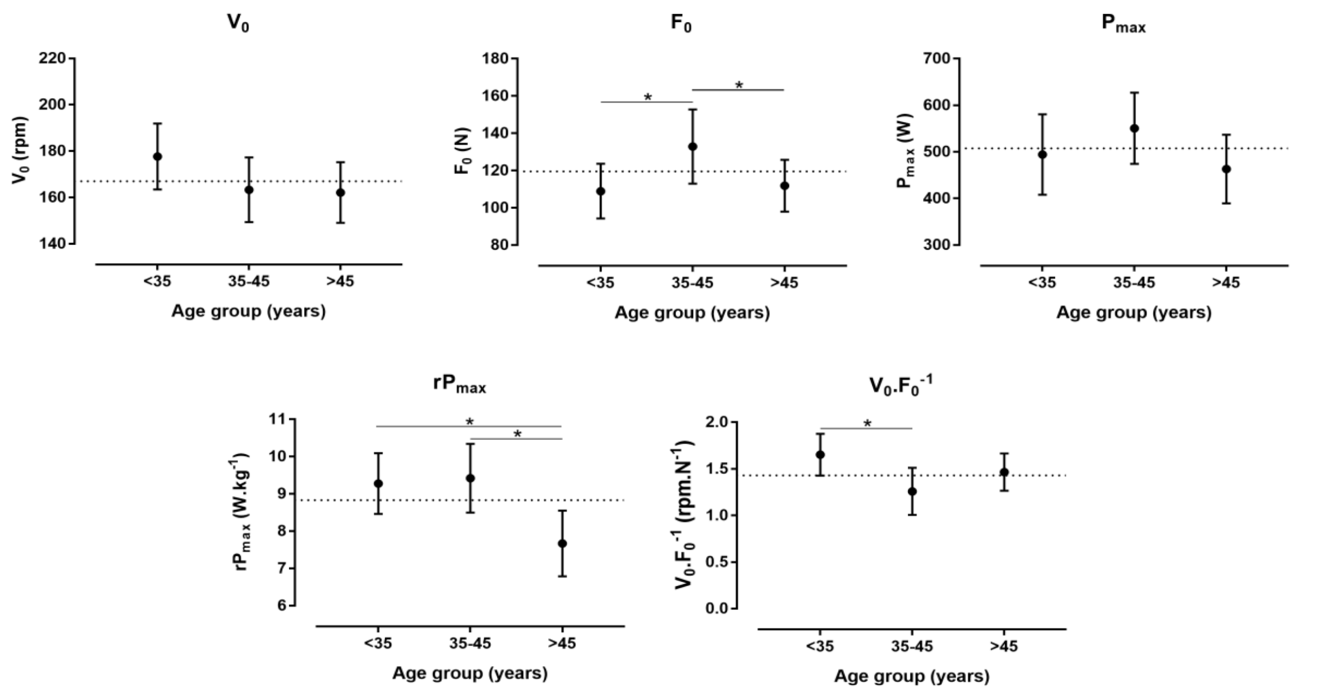

FIGURE 3 | Differences in theoretical maximal velocity (v_0), force (F_0), maximal power in absolute (Pmax) and relative values (rPmax), and $v_0.F_0^{-1}$ among age groups. * Difference at $p < 0.05$.

TABLE 2 | Comparison among performance groups (quartiles).

	Performance groups		
	Fast (n = 11)	Average (n = 11)	Slow (n = 11)
Age (years)	38.8 ± 10.7	41.0 ± 7.8	40.0 ± 8.6
Height (cm)	163 ± 7	162 ± 5	162 ± 8
Body mass (kg)	58.1 ± 7.4	57.6 ± 6.6	57.3 ± 8.6
BMI (kg.m^{-2})	21.7 ± 1.7	21.8 ± 1.9	21.9 ± 2.9
BF (%)	19.3 ± 2.6	20.6 ± 3.9	18.7 ± 6.6
FFM (kg)	47.0 ± 6.5	45.5 ± 4.1	46.3 ± 5.0
CSA (cm^2)	120 ± 18	117 ± 14	118 ± 20
SAR (cm)	24.0 ± 8.1	22.5 ± 7.6*	30.8 ± 7.2*
SJ (cm)	18.6 ± 3.5	17.6 ± 3.3	17.0 ± 3.5
CMJ (cm)	19.0 ± 3.9	18.7 ± 3.9	18.0 ± 3.8
Isometric muscle strength			
Right HG (kg)	30.8 ± 6.2	30.5 ± 4.0	28.0 ± 2.9
Left HG (kg)	30.8 ± 4.6	30.6 ± 4.0	28.0 ± 2.9
Back (kg)	89.8 ± 20.7	83.4 ± 10.0	75.1 ± 14.4
Back-legs (kg)	104.9 ± 26.4	94.4 ± 12.7	86.5 ± 12.6
Absolute sum (kg)	256.2 ± 54.4	238.8 ± 23.8	217.5 ± 30.8
Relative sum (kg.kg^{-1})	4.39 ± 0.69	4.16 ± 0.24	3.84 ± 0.58
Force-velocity test			
v_0 (rpm)	167 ± 20	170 ± 13	165 ± 12
F_0 (N)	118 ± 22	121 ± 24	120 ± 14
P_{max} (W)	503 ± 114	517 ± 81	502 ± 63
rP_{max} (W.kg^{-1})	8.59 ± 1.17	9.04 ± 1.34	8.84 ± 1.04
$v_0.F_0^{-1}$ (rpm.N^{-1})	1.45 ± 0.30	1.46 ± 0.30	1.39 ± 0.24

*BMI, body mass index; BF, body fat percentage; FFM, fat-free mass; CSA, thigh muscle cross-sectional area; SAR, sit-and-reach test; SJ, squat jump; CMJ, countermovement jump; HG, handgrip; v_0, theoretical maximal velocity; F_0, force; P_{max}, maximal power in absolute values; rP_{max}, maximal power in relative values. *Difference at p = 0.05.*

TABLE 3 | Correlations between anthropometric characteristics and physical fitness tests.

	Body mass	BMI	BF	CSA	FFM
SAR	0.149	0.067	−0.302	0.195	0.316
SJ	−0.112	−0.202	−0.418*	−0.027	0.078
CMJ	−0.123	−0.160	−0.301	−0.027	0.007
Isometric muscle strength					
Right HG	0.674‡	0.345	0.103	0.430*	0.710‡
Left HG	0.699‡	0.460†	0.140	0.500†	0.719‡
Back	0.487†	0.221	−0.016	0.460†	0.562†
Back-legs	0.514†	0.220	−0.056	0.484†	0.622‡
Absolute sum	0.587‡	0.278	−0.008	0.514†	0.674‡
Relative sum	−0.159	−0.347	−0.401*	0.158	0.034
Force-velocity test					
v_0	0.064	−0.023	−0.069	0.121	0.111
F_0	0.562†	0.435*	0.130	0.345	0.563†
P_{max}	0.604‡	0.416*	0.091	0.407*	0.633‡
rP_{max}	−0.207	−0.219	−0.309	0.007	−0.078
$v_0.F_0^{-1}$	−0.451†	−0.352*	−0.102	−0.261	−0.450*

*BMI, body mass index; BF, body fat percentage; FFM, fat-free mass; CSA, thigh muscle cross-sectional area; SAR, sit-and-reach test; SJ, squat jump; CMJ, countermovement jump; HG, handgrip; v_0, theoretical maximal velocity; F_0, force; P_{max}, maximal power in absolute values; rP_{max}, maximal power in relative values. *$p < 0.05$, †$p < 0.01$, ‡$p < 0.001$.*

DISCUSSION

The main findings of the present study were that (i) participants had average scores of body composition and physical fitness compared to the general population, (ii) the <35 age group had better jumping ability than 35–45 and >45 age group, and the older age group had lower F_0, Pmax and rPmax than their younger counterparts, (iii) the slowest performance group scored the highest in SAR, and (iv) isometric strength, F0 and Pmax correlated largely with body mass and FFM.

Profile

CMJ was in similar levels as sex- and age-matched general population (Edwen et al., 2014). Participants had lower CMJ and F-v characteristics than volleyball players (Nikolaidis, 2013) indicating that female marathon runners characterized by physiological range of muscle strength and power. This observation was in agreement with research in male runners, where relatively low scores of anaerobic power were shown in marathon runners compared to shorter distances' runners (Vuorimaa, 1996; Legaz-Arrese et al., 2011). For instance, CMJ in male marathon runners was ∼13 cm and 24 cm lower than middle-distance runners and sprinters, respectively (Vuorimaa, 1996). All measures of isometric muscle strength (right and left handgrip, back, leg, total, and relative) were classified as average compared to general population (Heyward and Gibson, 2014).

Age

The older age group had lower jumping ability, F_0, Pmax and rPmax than their younger counterparts, suggesting a decline of these physical fitness components with aging and indicating that probably the chronic endurance exercise does not prevent from loss in muscle strength and power. It should be highlighted that the participants had sport experience in endurance running for 5.6 years. These findings confirmed previous studies in male participants which showed a decline of anaerobic power with aging (Chamari et al., 1995; Bonnefoy et al., 1998). For instance, in a comparison between young (25 years) and master athletes (65 years) matched for weight, height and training, P_{max} was ∼43%, F_0 30% and v_0 15% lower in the older athletes (Chamari et al., 1995). A study of young (23 years) and elder male participants (71 years) showed a decline of rP_{max} by 8% per decade and of velocity by 4%, and a moderate inverse relationship between rP_{max} and age (r = -0.33) (Bonnefoy et al., 1998).

Performance

Among all anthropometric characteristics and physical fitness components examined in the present study, flexibility was the only one observed to differ among performance groups with the slowest one presenting the better score in SAR than the average group. This observation might be due to that running economy (which is related with performance) is inversely correlated with sit-and-reach test score (Jones, 2002; Trehearn and Buresh, 2009). An increased storage and return of elastic energy in stiffer musculotendinous structures might reduce the aerobic demand of submaximal running (Drew et al., 2011). In addition, no difference was shown among performance groups with regards to muscle strength and power. This finding was in agreement with a previous research showing that fast marathon runners were not characterized by high anaerobic power (Vuorimaa, 1996; Legaz-Arrese et al., 2011).

Anthropometry

Most indices of muscle strength and power correlated with both body mass, FFM and CSA, which might explain why female marathon runners are not characterized by high levels of muscle strength and power as their body dimensions are relatively small compared to other sports. An excess of FFM, even if this is "active mass," is a load that marathon runners need to carry with them; thus, their profile consists of small anthropometric characteristics and corresponding moderate levels of muscle strength and power. Although an increased Pmax might improved the cost of running (Giovanelli et al., 2017), its association with increased FFM would lead to slower race time.

Limitations, Strength, Practical Applications

A limitation of the present study was the specificity of protocols that assessed the various physical fitness components as caution would be needed to compare their findings with studies using other protocols. For instance, F-v test and WAnT reflect different aspects of anaerobic power and capacity and their findings should not be used interchangeably (Jaafar et al., 2016). Strength of this study was its novelty since it was the first to examine flexibility, isometric muscle strength, two jump tests (SJ and CMJ) and F-v characteristics of female marathon runners and the findings could be used as norms and references for future studies. Considering the gap in the existing literature about these physical fitness components in female marathon runners, the findings add new information. The present study confirmed on female runners the findings of previous studies on male runners showing that flexibility, muscle strength and power were not related to performance in marathon runners. On the other hand, flexibility and muscle strength are components of the health-related physical fitness, and in this context, they should be regularly monitored in addition to sport-related physical fitness components, such as aerobic capacity (maximal oxygen uptake, anaerobic threshold, and running economy). In view of the increased female participation in marathon races during the last decades (Lepers and Cattagni, 2012), the findings were of great practical value for strength and conditioning coaches in the context of training and testing of their runners. Surprisingly, although an optimal level of flexibility and muscle strength is important for health, these fitness components have been rarely studied in male marathon runners (Maud et al., 1981). Thus, the present study filled a gap in the existing literature as it was the first study – to the best of our knowledge – to provide data on F-v characteristics and the abovementioned health-related physical fitness components in female marathon runners.

CONCLUSION

Profiling physical fitness characteristics of marathon runners is of great practical importance for strength and conditioning coaches working in this sport. The assessment of physical fitness assists to evaluate the effectiveness of training. So far, a lack of reference data on female marathon runners' physical fitness, especially with regards to anaerobic power, muscle strength and flexibility, has been observed. Therefore, the data reported in this study would be useful for strength and conditioning trainers to monitor the training of female marathon runners. Strength and conditioning coaches may work with female marathon runners differing for age, performance level and anthropometric characteristics; thus, knowledge about the effect of age, performance and anthropometry on physical fitness would assist to accurately evaluate and prescribe training program. The findings highlighted the lower leg strength in the older age group; thus, strength and conditioning coaches should focus on the development of age-tailored training programs to enhance the jump ability of older female runners. On the other hand, flexibility should be monitored regularly targeting and be within a physiological range, whereas a high flexibility should be alarming as it associates with reduced performance in marathon. Considering the increased number of female finishers in marathon races during the last years, the findings have practical applications to a large number of recreational marathon runners. It should be also highlighted that the age and performance level of participants in the present study (∼40 years old and 4:34 h:min, respectively) was close to the average of finishers in large marathon races such as the "New York City marathon" (∼39 years old and 4:48 h:min, respectively).

AUTHOR CONTRIBUTIONS

PN performed the laboratory tests, analyzed the data and drafted the manuscript, TR and BK helped in drafting the manuscript.

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Acute Dehydration Impairs Endurance Without Modulating Neuromuscular Function

Oliver R. Barley, Dale W. Chapman, Anthony J. Blazevich and Chris R. Abbiss*

Centre for Exercise and Sports Science Research, School of Medical and Health Sciences, Edith Cowan University, Joondalup, WA, Australia

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Oliver Faude,
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United Kingdom

****Correspondence:***
Oliver R. Barley
o.barley@ecu.edu.au

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Introduction/Purpose: This study examined the influence of acute dehydration on neuromuscular function.

Methods: On separate days, combat sports athletes experienced in acute dehydration practices (n = 14) completed a 3 h passive heating intervention (40°C, 63% relative humidity) to induce dehydration (DHY) or a thermoneutral euhydration control (25°C, 50% relative humidity: CON). In the ensuing 3 h *ad libitum* fluid and food intake was allowed, after which participants performed fatiguing exercise consisting of repeated unilateral knee extensions at 85% of their maximal voluntary isometric contraction (MVIC) torque until task failure. Both before and after the fatiguing protocol participants performed six MVICs during which measures of central and peripheral neuromuscular function were made. Urine and whole blood samples to assess urine specific gravity, urine osmolality, haematocrit and serum osmolality were collected before, immediately and 3 h after intervention.

Results: Body mass was reduced by 3.2 ± 1.1% immediately after DHY ($P < 0.001$) but recovered by 3 h. Urine and whole blood markers indicated dehydration immediately after DHY, although blood markers were not different to CON at 3 h. Participants completed 28% fewer knee extensions at 85% MVIC ($P < 0.001$, $g = 0.775$) and reported a greater perception of fatigue ($P = 0.012$) 3 h after DHY than CON despite peak torque results being unaffected. No between-condition differences were observed in central or peripheral indicators of neuromuscular function at any timepoint.

Conclusion: Results indicate that acute dehydration of 3.2% body mass followed by 3 h of recovery impairs muscular strength-endurance and increases fatigue perception without changes in markers of central or peripheral function. These findings suggest that altered fatigue perception underpins muscular performance decrements in recovery from acute dehydration.

Keywords: combat sports, dehydration, hypohydration, weight cutting, weight loss

INTRODUCTION

Total body water is essential to the physiological function of the human body. As such, the balance of total body water has been the topic of much previous research, with inducing a deficit in total body water (dehydration) and the effects on exercise performance being a large focus (Horswill and Janas, 2011; Cheuvront and Kenefick, 2014). Dehydration of 2–3% of body mass (BM) has been found to impair both aerobic exercise performance, especially when convective cooling is minimal (Barr, 1999; Cheuvront et al., 2010), and anaerobic performance (Kraft et al., 2012). The research is less clear when examining ultra-endurance performance with studies finding increased dehydration to be associated with improved performance (Zouhal et al., 2010; Knechtle et al., 2012). The method of how dehydration is induced may also be of some importance, as it can be induced using exercise (actively) or by using environmental stress at rest (passively) (Cheuvront and Kenefick, 2014). The negative effects of both active and passive dehydration have also been found to persist even following rehydration, with researchers reporting impairments in repeat-effort capacity (Barley et al., 2017b), sports-specific skills (Baker et al., 2007), mood (Hall and Lane, 2001) and cognitive function (Choma et al., 1998) for up to 5 h, and in some cases 24 h, despite *ad libitum* rehydration. Such findings have substantial implications for athletes in sports with weight classifications (e.g., combat sports), as athletes often attempt to rapidly lose and regain body mass prior to being weighed for competition, with athletes reporting losing 3–5% of their body mass rapidly before being weighed for competition (Barley et al., 2017a).

Dehydration impairs exercise performance through multiple mechanisms including an increased cardiovascular strain (González-Alonso et al., 1997), a reduced muscle blood flow compromising oxygen delivery and aerobic metabolism (Cheuvront et al., 2010), impaired thermoregulatory function (Casa, 1999; Cheuvront et al., 2010), and increased carbohydrate metabolism (Casa, 1999). Dehydration may also compromise neuromuscular function as a result of alterations in electrolyte concentrations (particularly sodium and potassium) within the interstitial and intracellular spaces (Sjøgaard, 1985; Casa, 1999). Electrolyte balance is important for the maintenance of membrane electrochemical potential and actin-myosin function which, if significantly altered, may reduce function of neurones and muscle fibers (Sjøgaard, 1985). Consequently, dehydration may impair physical performance as a result of fatigue occurring both proximal (central) and distal (peripheral) to the neuromuscular junction (Minshull and James, 2012). Thermal exposure has also been found to negatively influence neuromuscular function both and rest and following exercise which may be of importance when thermal stress is used to induce dehydration (Ross et al., 2011; Goodall et al., 2015). Research examining the effects of dehydration on physical performance and neuromuscular function is mixed, with studies typically observing a negative (Bigard et al., 2001; Ftaiti et al., 2001; Minshull and James, 2012; Bowtell et al., 2013) or no effect (Evetovich et al., 2002). However, the effects of acute dehydration on performance and neuromuscular function following a periods of recovery and/or rehydration is even less clear due to a paucity of research (Bigard et al., 2001; Rodrigues et al., 2014). As a result, it is currently unclear whether the reduction in exercise capacity observed in the hours following dehydration (Hall and Lane, 2001; Barley et al., 2017b) results from prolonged alterations in neuromuscular function (i.e., central and peripheral fatigue) or other mechanisms such as altered carbohydrate metabolism or mental fatigue (Sjøgaard, 1985; Montain et al., 1998; Casa, 1999; Bigard et al., 2001; Minshull and James, 2012). Indeed, dehydration has also been found to negatively influence mood and cognitive function (Choma et al., 1998; Hall and Lane, 2001) and increase perceived exertion, which could be associated with increased mental fatigue (Marcora et al., 2009; Barley et al., 2017b). While it is possible that changes in mood and cognition are a consequence of changes in neural function, we are not aware of any studies that have examined this in detail. As such, the aim of the present study was to examine the influence of 3% acute dehydration on muscular force production and endurance, neuromuscular markers of central and peripheral fatigue, mood and cognition in combat sport athletes with experience in acute dehydration strategies. While previous research has investigated the influence of dehydration on neuromuscular function and muscular endurance, we are unaware of research that has conducted an in-depth evaluation of the potential central and peripheral changes in neuromuscular function following rehydration. We hypothesized that acute dehydration would impair muscular endurance alongside central and peripheral neuromuscular function.

MATERIALS AND METHODS

Participants

Fourteen highly trained male combat sports athletes (age 25 ± 4 years, height 1.8 ± 0.05 m, body mass 80 ± 11 kg) with no history of leg injuries and at least 2 years of competitive combat sports experience volunteered for the study. Subjects were recruited via advertisements or word of mouth. All participants were required to have experience using acute dehydration strategies to make weight for competitions. We conducted an *a priori* power analysis using previous research investigating the reliability of neuromuscular assessments of the quadriceps muscles (Place et al., 2007) and estimated that 14 participants would be required to identify statistical differences of a 0.25 effect size or greater with 95% power and an α level of 0.05. This study was carried out in accordance with the recommendations of the Australian National Statement on Ethical Conduct in Human Research with written informed consent from all subjects. All subjects gave written informed consent in accordance with the Declaration of Helsinki. The protocol was approved by the Edith Cowan University Human Research Ethics Committee.

Study Design

Participants completed one familiarization session and two experimental sessions. In the familiarization session, assessments of mood, cognition and neuromuscular function were practiced until the participants expressed confidence and provided repeatable results. The experimental sessions were performed at the same time of day and separated by at least 7 days, with the session order randomized and counterbalanced. During experimental sessions participants performed either a dehydration (DHY) or control (CON) protocol, which were both followed by 3 h of *ad libitum* food and fluid intake. DHY involved 3 h of passive heat exposure to induce dehydration which is a common method of weight loss in combat sports (Barley et al., 2017a) while CON involved 3 h of exposure to thermoneutral conditions as detailed below. Mood was assessed before, immediately and 3 h after both DHY and CON. Cognition and neuromuscular function were assessed only before and at 3 h after DHY and CON. Cognition and neuromuscular function were not assessed immediately following DHY and CON due to the likelihood of the testing influencing performance in the tests at 3 h. Each participant was asked to record their nutritional intake for the 24 h prior to and during the first experimental session and then replicate intake during the second experimental session.

DHY involved 3 h of passive heat exposure in an environmental chamber at 39.9 ± 0.3°C, and 63 ± 2% relative humidity with the aim of reducing body mass by 3–4%. Participants wore a plastic suit (plastic sweat suit, Wrap Yourself Slim, Australia) and were not permitted to consume fluids during the protocol. CON involved 3 h of thermoneutral exposure (23.5 ± 0.7°C, and 50 ± 12%) with the aim of maintaining body mass and euhydration. Participants were permitted to drink fluids throughout the protocol. Core temperature was continuously monitored throughout the experimental trials using a gastrointestinal pill ingested 4 h prior to testing (CorTemp ingestible core body temperature sensor, HQinc, United States). Heart rate was continuously recorded using a Polar heart rate monitor (Model S810i, Polar Electro Oy, Kempele, Finland). Following both DHY and CON participants were free to consume food and fluid *ad libitum*, with the participants encouraged to consume food and fluid as they would in preparation for a competition.

Immediately before, after and 3 h after DHY or CON, nude body mass (Mettler 1D1 multirange, FSE, Australia), brachial blood pressure (Automatic blood pressure monitor, OMRON, Singapore), core temperature, tympanic temperature (ThermoScan, Braun, Germany) and heart rate were recorded. At these same time points, finger prick whole blood samples were collected into capillary tubes (Capilette MPW-212, MPW med. Instruments, South Australia) and spun at 3600 rpm to determine haematocrit (Centrifuge MPW-212, MPW med. Instruments, South Australia), whilst urine samples were collected and assessed for osmolality (Advanced 3250 single-sample osmometer, Advanced instruments, United States) and urine specific gravity (Atago hand refractometer, model UNC-NE, Atago, Japan). Venous blood samples were collected into an 8.5 ml serum separating tube vacutainer and spun at 12000 rpm for 15 min at 4°C in a centrifuge (Multifuge 3 S-R, Kendro, United States) to obtain serum. A 200 μl sample of serum was assessed for osmolality (Advanced 3250 single-sample osmometer, Advanced instruments, United States) while the remaining serum was aliquoted equally and immediately frozen at −80°C to be later analyzed for brain-derived neurotropic factor (BDNF) to evaluate any potential physiological influences on cognition (Roh et al., 2017) and tumor necrosis factor alpha (TNFα) to evaluate any potential tissue damage resulting from heat stress (Collins and Grounds, 2001) using enzyme-linked immunosorbent assay kits (Quantikine HS ELISA, R&D Systems, Minneapolis, MN, United States).

Profile of Mood States Short Form

Mood was assessed using a Profile of Mood States Short Form (POMS-SF). The POMS-SF assesses anger, confusion, depression, fatigue, tension and vigor. Anger items included "Angry" and "Annoyed"; confusion items included "Bewildered" and "Forgetful"; depression included terms such as "Unhappy" and "Hopeless"; fatigue items included "Worn out" and "Fatigued"; tension items included "On Edge" and "Nervous"; and vigor included terms such as "Active" and "Lively." Items were rated on a 5-point scale ranging from "Not at all" [0] to "Extremely" [5]. Responses to the POMS-SF have been found to be comparable to the original POMS (Curran et al., 1995).

Cognitive Assessment

Cognitive function was assessed using the CogState computerized test battery (CogState, CogState Ltd, Melbourne, VIC, Australia). Cogstate has been found to have a high test–retest reliability and to be sensitive to mild changes in cognitive state (Collie et al., 2003). The test takes approximately 15 min and utilizes playing cards as the stimulus. The test assesses simple reaction time, choice reaction time, attention/visual learning and memory, and attention and working memory. Response time and accuracy were reported for each task. In order to minimize potential learning effects, the participants were familiarized with the protocol during the familiarization session.

Neuromuscular Assessments

Neuromuscular function was assessed using a maximal voluntary isometric knee extension contraction (MVIC) protocol before DHY or CON as well as both before and after a fatiguing knee extensor exercise protocol at 3 h after DHY and CON. The MVIC protocol required the performance of six maximal voluntary isometric contractions. Three were performed using brief contractions (<2 s) with the verbal instruction "as fast as possible" whilst another three required longer contractions (3–5 s) utilizing the verbal instruction "as hard and as fast as possible"; a 30-s rest was provided between efforts (Maffiuletti et al., 2016). An electrical stimulus was applied to the quadriceps during the three longer contractions to assess muscle voluntary activation, as described below. The participants were seated in an isokinetic dynamometer (Biodex System 3 Pro, Biodex Medical System, Shirley, NY, United States) with their dominant leg attached and the knee fixed to 60° (0°= Full extension). Before tests (as a warm-up) the participants performed eight brief voluntary knee extensor contractions beginning at 30% of perceived MVIC and progressively increasing until reaching 100% of perceived MVIC for the final contraction. A 2-min rest was given before the testing commenced. The contractile rate of force development (RFD) was assessed using the torque generated in the first 75 ms of each MVIC ($T_{75\ ms}$). This time epoch was chosen as it is reported that RFD is strongly influenced by neural factors in this time period (Maffiuletti et al., 2016). Additionally, the reliability of shorter time periods (i.e., 30–50 ms) may be poor when conducted on commercially available dynamometers (Maffiuletti et al., 2016).

During the fatiguing exercise protocol participants performed repeated 5-s knee extensor contractions with 5-s of recovery at 85% of their baseline MVIC. Contractions continued until participants failed to reach the required torque in two consecutive contractions. Visual feedback was constantly provided to participants on a large monitor using LabChart software (Version 7.1.3, ADInstruments, Sydney, NSW, Australia). The best MVIC and $T_{75\ ms}$ values obtained in the six MVICs before exercise and the first contraction following the exercise protocol (i.e., at fatigue) were used for data analysis. Torque data were recorded using LabChart software on a laptop computer using a 16-bit analog-to-digital converter (PowerLab 16/35, ADInstruments, Sydney, NSW, Australia) sampling at 4000 Hz.

Surface Electromyography (EMG)

During all contractions surface EMG data were obtained from vastus lateralis (VL) and vastus medialis (VM). EMG electrodes (720 Neuroline, Ambu, NSW, Australia) were placed on the quadriceps muscles as per SENIAM guidelines (Hermens et al., 1999) with a bipolar electrode configuration with a 1-cm inter-electrode distance. Prior to the application of EMG electrodes, the skin was shaved, abraded and cleaned with alcohol to reduce inter-electrode resistance below 5 kΩ. Following this, placement locations were marked with permanent marker ink and the electrodes were applied before being secured using sports cloth tape. EMG electrodes were replaced if they had an inter-electrode resistance above 5 kΩ after DHY as previous research has shown this to provide reliable results (Abbiss et al., 2006). All EMG data were amplified (× 1000) and filtered using a 20–500 Hz band-pass filter before a symmetric root-mean-square filter was applied with a 500-ms averaging window (EMG_{RMS}). Prior to analysis EMG_{RMS} data were normalized to the maximal M-wave amplitude (M_{max}) to control for potential peripheral changes (Place et al., 2007). EMG data were recorded synchronously with torque data at a 2000 Hz analog-to-digital conversion rate using LabChart software on a laptop computer (PowerLab 16/35, ADInstruments, Sydney, NSW, Australia). The maximal EMG_{RMS} during the six MVICs before fatigue and the first MVIC immediately after fatigue were used for analysis.

Electrical Stimulation Procedures

Before and 3 h after both DHY and CON the maximal M-wave amplitude (M_{max}) and excitation-contraction (E-C) coupling efficiency were assessed. E-C coupling efficiency was also assessed immediately following the fatiguing exercise protocol. Femoral nerve stimulation was used to deliver electrical stimuli for M_{max} assessment, whilst tetanic muscle stimulation was used to assess E-C coupling efficiency. The femoral nerve was located manually and then stimulated using single 0.2-ms square-wave pulses with a constant-current stimulator (DS7A, Digitimer Ltd., Welwyn Garden City, United Kingdom). A compex electrode pen was used to locate the nerve and then an electrode (WhiteSensor 4570M, Ambu, NSW, Australia) was placed on the skin for subsequent stimulations. To find M_{max}, resting femoral nerve stimulations were imposed every 10 s from a subthreshold intensity, where no evoked response was observed, until a peak M-wave amplitude was observed. The stimulus intensity used to elicit the M_{max} was then increased by 40% for subsequent testing to ensure a supramaximal stimulus intensity to account for possible depression in motor responses during fatigue (Trajano et al., 2013). To assess E-C coupling efficiency, electrical square-wave stimuli (0.5-ms pulse width) were delivered to the knee extensor muscle belly through four self-adhesive electrodes (5 × 9 cm, Durastick II, Chattanooga group, Hixson, TN, United States) using a constant-current stimulator (DS7A, Digitimer Ltd., Welwyn Garden City, United Kingdom). For all tetanic stimulations, the stimulation intensity necessary to reach 50% of MVIC with a 0.5-s 80 Hz tetanic stimulation was used (Martin et al., 2004). Three evoked contractions of the same duration were delivered with 15 s between each contraction using the following trains: (1) 20 Hz train of 11 pulses (0.05-s interpulse interval); (2) variable-frequency train (VFT) (i.e., 2 pulses at 0.01-s plus, 10 pulses at 0.05-s interpulse interval); (3) 80 Hz train of 36 pulses (0.0125-s interpulse interval) (Trajano et al., 2013). Voluntary activation (VA%), peak twitch torque ($T_{tw,p}$), time to peak twitch ($t_{tw,p}$), peak twitch half relaxation time ($t_{1/2}$), the ratio of torques evoked by 20 and 80 Hz stimulations (20:80), and the ratio of torques 20 Hz and VFT trains (20:VFT) were assessed. The three tetanic trains have been used by previous researchers to provide information relating to muscular calcium concentration, sensitivity and the rate of binding to troponin (Binder-Macleod and Lee, 1996; Martin et al., 2004; Binder-Macleod and Kesar, 2005). The maximal VA% and $T_{tw,p}$ and minimum $t_{tw,p}$ and $t_{1/2}$ during the MVICs and stimulations before fatigue and the first MVIC and stimulation immediately following volitional fatigue were used in data analysis. $T_{75\ ms}$, VA%, VL and VM EMG/M were used as markers of central neuromuscular function (Trajano et al., 2013; Maffiuletti et al., 2016) while 20:80, 20:VFT, $T_{tw,p}$, $t_{tw,p}$, and $t_{1/2}$ were used as markers of peripheral neuromuscular function (Behm and St-Pierre, 1997; Behm et al., 2002; Martin et al., 2004). Voluntary activation (VA%) was determined using the correction equation described by Strojnik and Komi (1998):

$$VA\% = 100 - Tw_{\text{MVIC}} \times (MVIC_{\text{stim}}/MVIC_{\text{peak}})/Tw_{\text{pot}} \times 100$$

where VA% is the corrected voluntary activation, Tw_{MVIC} is the additional twitch torque at MVIC, $MVIC_{stim}$ is the torque level at stimulation, $MVIC_{peak}$ is the peak torque during MVIC, and Tw_{pot} is the potentiated twitch torque at rest.

Statistical Analysis

Shapiro–Wilk's tests were used to verify the assumption that the data were normally distributed. Torque, EMG, temperature, blood and urine data were analyzed separately using two-way repeated measures ANOVAs. Data from the POMS-SF and CogState cognitive tests were not normally distributed and were assessed using Friedman's two-way ANOVAs. Where differences were observed in either normally or non-normally distributed data, *post hoc* tests with the Holm–Bonferroni sequential correction adjustment was used to determine the location of the differences (Holm, 1979). Statistical analysis was performed using SPSS version 24.0 (SPAA Inc., Chicago IL, United States) with statistical significance assumed at $P < 0.05$. Torque and EMG data are reported as mean, standard deviation

(SD), confidence intervals (95%) and Hedges' g effect sizes. An effect size of 0.2 was considered small, 0.5 moderate and >0.8 large. All other data are reported as mean (± SD) with P-values.

RESULTS

Body mass was not significantly (P = 0.947) different between conditions before intervention (DHY = 80.0 ± 1.9 and CON = 80.3 ± 1.4 kg), however, it was lower immediately after DHY (P < 0.001) when compared with CON (77.5 ± 10.4 and 79.7 ± 10.9 kg, respectively). Serum osmolality was significantly (P = 0.003) greater immediately following DHY (293.1 ± 4.8 mOsm) than CON (285.1 ± 4.4 mOsm), but no statistical differences were observed at any other time point. Immediately after, haematocrit was greater in DHY than CON (45 ± 3 and 43 ± 2, respectively, P = 0.034) but was not different 3 h after. Urine osmolality was greater in DHY (740 ± 249 and 728 ± 326 mOsm) than CON (338 ± 159 and 379 ± 192 mOsm) immediately (P < 0.001) and at 3 h post-intervention (P < 0.001). Urine specific gravity was greater in DHY (1.027 ± 0.008 and 1.022 ± 0.01 SG) than CON (1.008 ± 0.004 and 1.01 ± 0.005 SG) immediately (P < 0.001) and at 3 h post-intervention (P < 0.001) (**Table 1**). Resting heart rate was higher immediately following DHY than CON (117 ± 21 and 56 ± 9, respectively, P < 0.001). Both tympanic and core temperature were higher immediately following DHY than CON (P < 0.001 for both markers) but were not different at any other timepoint. No significant differences were observed in blood pressure, BDNF or TNFα concentrations between conditions (**Table 1**).

No significant main effects between conditions were observed for MVIC torque, $T_{75\ ms}$, VA% or EMG/M for VL or VM (**Table 2** and **Figure 1**) before or immediately following DHY compared with CON. Furthermore, no significant differences between conditions were observed in contractile properties ($T_{tw,p}$, $t_{tw,p}$, and $t_{1/2}$) or E-C coupling efficiency (20:VFT or 20:80) (**Table 2**).

Fewer contractions were completed during the fatiguing exercise protocol at 3 h after DHY (17 ± 7) than CON (23 ± 8) [P < 0.001, CI (−11.84, −0.16), g = 0.775]. MVIC torque and $T_{75\ ms}$ decreased after exercise (P < 0.001) but this change was not statistically different between conditions (**Table 3**). Additionally, no significant between-condition effects were observed in VA% or in VL or VM EMG/M before or immediately following fatiguing exercise (**Figure 1**). No significant differences were observed between conditions in contractile properties ($T_{tw,p}$, $t_{tw,p}$, and $t_{1/2}$) or E-C coupling efficiency (20:VFT or 20:80) in DHY when compared CON (**Table 3**).

Perception of tension was greater immediately after DHY when compared with CON (10 ± 5 and 7 ± 1, respectively; P = 0.036) but was not different at any other time point.

TABLE 1 | Mean (± SD) body mass and cardiovascular, urine and blood markers.

Physiological marker	Condition	Pre-intervention	Immediately post-intervention	3 h post-intervention
Body mass (kg)	DHY	80 ± 10	77 ± 10	79 ± 10
	CON	80 ± 11	80 ± 11***	80 ± 11
Serum osmolality (mOsm/kg)	DHY	286 ± 5	293 ± 5	290 ± 6
	CON	288 ± 4	285 ± 4**	287 ± 5
Haematocrit (%)	DHY	42 ± 2	45 ± 3	43 ± 2
	CON	42 ± 3	43 ± 3*	42 ± 2
Urine osmolality (mOsm/kg)	DHY	448 ± 318	740 ± 249	728 ± 326
	CON	654 ± 321	338 ± 159***	654 ± 321***
Urine specific gravity (SG)	DHY	1.013 ± 0.009	1.027 ± 0.008	1.022 ± 0.01
	CON	1.018 ± 0.01	1.008 ± 0.004***	1.01 ± 0.005***
Resting heart rate (bpm)	DHY	66 ± 10	117 ± 21	68 ± 12
	CON	69 ± 11	56 ± 9***	64 ± 10
Systolic blood pressure (mmHg)	DHY	126 ± 12	122 ± 13	127 ± 11
	CON	125 ± 14	124 ± 14	126 ± 10
Diastolic blood pressure (mmHg)	DHY	69 ± 10	69 ± 7	71 ± 9
	CON	69 ± 9	72 ± 5	69 ± 7
Core temperature (°C)	DHY	37 ± 0.5	39 ± 0.5	37 ± 0.5
	CON	37 ± 0.5	37 ± 0.2***	37 ± 0.2
Tympanic temperature (°C)	DHY	36 ± 0.5	39 ± 0.5	36 ± 0.5
	CON	36 ± 0.5	36 ± 0.5***	36 ± 0.5
Brain-derived neurotropic factor (μg/ml)	DHY	188 ± 149	161 ± 100	106 ± 65
	CON	114 ± 44	202 ± 212	115 ± 115
Tumor necrosis factor alpha (pg/ml)	DHY	0.93 ± 0.23	1.09 ± 0.3	1.02 ± 0.34
	CON	1.13 ± 0.49	1.08 ± 0.32	0.94 ± 0.22

*P < 0.05, **P < 0.01, ***P < 0.001 when compared with DHY in at the same time point.*

TABLE 2 | Mean (± SD), 95% confidence limit and Hedges *g* effect size EMG and torque data before and 3 h after DHY and CON.

Neuromuscular variable	Condition	Pre-intervention	3 h post-intervention	DHY vs. CON Pre-intervention	DHY vs.CON 3 h post-intervention
MVIC (Nm)	DHY	295 ± 48	297 ± 49	(−37, 39), 0.02	(−43, 35), 0.08
	CON	296 ± 50	293 ± 52		
$T_{75\ ms}$ (Nm)	DHY	74 ± 26	72 ± 24	(−13, 23), 0.324	(−12, 24), 0.214
	CON	79 ± 20	78 ± 22		
20:VFT (Nm)	DHY	0.998 ± 0.04	0.976 ± 0.05	(−0.04, 0.02), 0.291	(−0.01, 0.07), 0.643
	CON	0.986 ± 0.04	1.006 ± 0.04		
20:80 (Nm)	DHY	0.698 ± 0.11	0.705 ± 0.07	(−0.07, 0.08), 0.04	(−0.06, 0.06), 0.013
	CON	0.702 ± 0.09	0.704 ± 0.08		
$t_{tw,p}$ (s)	DHY	0.078 ± 0.002	0.077 ± 0.016	(0.00, 0.01), 0.401	(−0.01, 0.02), 0.258
	CON	0.083 ± 0.017	0.081 ± 0.014		
$T_{tw,p}$ (Nm)	DHY	70 ± 12	71 ± 12	(−8.94, 8.94), 0.000	(−11.32, 7.32), 0.162
	CON	70 ± 11	69 ± 12		
$t_{1/2}$ (s)	DHY	0.074 ± 0.022	0.073 ± 0.021	(−0.02, 0.02), 0.042	(−0.02, 0.01), 0.231
	CON	0.075 ± 0.024	0.068 ± 0.021		

Differences are displayed as 95% CI (LL, UL) and effect size, no P-values are displayed as no significant main effects were observed.
Results displayed are maximal voluntary isometric contraction (MVIC), torque at 75 ms ($T_{75\ ms}$), 20 Hz:variable frequency train ratio (20:VFT), 20 Hz:80 Hz ratio (20:80), time to peak twitch ($t_{tw,p}$), peak twitch torque ($T_{tw,p}$), peak twitch half relaxation time ($t_{1/2}$).

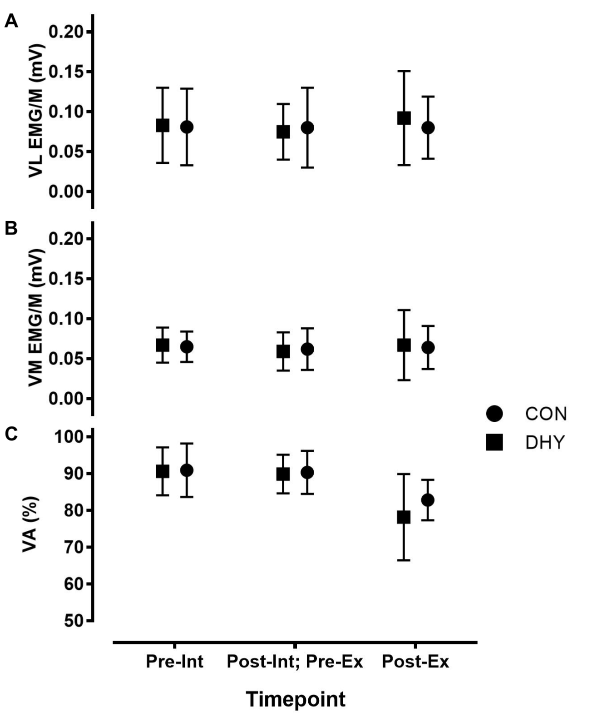

FIGURE 1 | Maximal vastus lateralis (VL) **(A)**, vastus medialis (VM) **(B)** EMG_{RMS}/M and voluntary activation (VA%) **(C)** before and 3 h after DHY/CON (Int), as well as before and immediately after fatiguing exercise (Ex). No significant differences were observed between VA (%) or vastus lateralis and medialis EMG_{RMS}/M, indicating no influence on central neuromuscular function resulting from acute dehydration.

Depression (13 ± 6 and 8 ± 1, respectively; $P = 0.033$) and anger (13 ± 8 and 7 ± 1, receptively, $P = 0.036$) were only greater immediately after DHY when compared with CON. Confusion was also greater only immediately after the DHY when compared with CON (10 ± 5 and 6 ± 3, respectively; $P = 0.045$). However, perception of fatigue after DHY was greater than CON immediately (17 ± 6 and 6 ± 2, respectively; $P = 0.003$) and 3 h after (11 ± 5 and 6 ± 2, respectively; $P = 0.012$). No main effects were observed in vigor at any time point (**Table 4**).

The majority of cognitive function measures were unchanged except for one-card accuracy and speed. One card learning accuracy was significantly ($P = 0.02$) less immediately following DHY compared with CON (68 ± 9 and 75 ± 8%, respectively) but no differences were observed at 3 h (**Table 5**). Additionally, one card learning speed immediately following DHY was significantly ($P = 0.048$) faster than CON (865 ± 192 and 909 ± 184 ms, respectively) but no differences were observed after 3 h (**Table 5**).

DISCUSSION

In the present study, the influence of the acute loss of 3% of body mass, achieved by 3 h of passive dehydration in the heat, on muscular strength and endurance, central and peripheral neuromuscular function, psychological profile, cognitive function, body temperature and markers of hydration status was examined. While previous research has examined the influence of acute dehydration on muscular performance, a novel aspect of the present study was investigating the central and peripheral neuromuscular mechanisms underpinning such performance. The main observations were that: (i) 3 h of *ad libitum* fluid and food consumption following acute dehydration of 3% resulted in the recovery of body mass, serum osmolality and haematocrit but not urinary markers of hydration status, (ii) the number of isometric knee extensor contractions performed at 85% of MVIC was less 3 h after DHY when compared with CON, (iii) peak torque or $T_{75\ ms}$ were not significantly different between conditions at any time point,

TABLE 3 | Mean (± SD), 95% confidence limit and Hedges *g* effect size EMG and torque data before and immediately after fatiguing exercise for both DHY and CON.

Neuromuscular variable	Condition	Pre-exercise	Post-exercise	DHY vs. CON Pre-exercise	DHY vs. CON Post-exercise
MVIC (Nm)	DHY	297 ± 49	236 ± 33	(−43, 35), 0.08	(−31, 15), 0.026
	CON	293 ± 52	228 ± 27		
$T_{75\ ms}$ (Nm)	DHY	72 ± 24	58 ± 20	(−12, 24), 0.214	(−18, 14), 0.0
	CON	78 ± 22	56 ± 21		
20:VFT	DHY	0.976 ± 0.05	0.98 ± 0.03	(−0.01, 0.07), 0.643	(−0.04, 0.02), 0.302
	CON	1.006 ± 0.04	0.969 ± 0.04		
20:80	DHY	0.705 ± 0.07	0.678 ± 0.08	(−0.06, 0.06), 0.013	(−0.08, 0.05), 0.194
	CON	0.704 ± 0.08	0.661 ± 0.09		
$t_{tw,p}$ (s)	DHY	0.077 ± 0.016	0.087 ± 0.019	(−0.01, 0.02), 0.258	(−0.02, 0.01), 0.377
	CON	0.081 ± 0.014	0.08 ± 0.017		
$T_{tw,p}$ (Nm)	DHY	71 ± 12	62 ± 11	(−11, 7), 0.162	(−9, 75), 0.211
	CON	69 ± 12	60 ± 7		
$t_{1/2}$ (s)	DHY	0.073 ± 0.021	0.085 ± 0.028	(−0.02, 0.01), 0.231	(−0.02, 0.02), 0.104
	CON	0.068 ± 0.021	0.088 ± 0.028		

Differences are displayed as 95% CI (LL, UL) and effect size, no P-values are displayed as no significant main effects were observed.
Results displayed are maximal voluntary isometric contraction (MVIC), torque at 75 ms ($T_{75\ ms}$), 20 Hz:variable frequency train ratio (20:VFT), 20 Hz:80 Hz ratio (20:80), time to peak twitch ($t_{tw,p}$), peak twitch torque ($T_{tw,p}$), peak twitch half relaxation time ($t_{1/2}$).

(iv) there were no significant differences between DHY and CON conditions in markers of either central and peripheral neuromuscular function at any time point, and (v) acute dehydration increased perception of fatigue both immediately and 3 h after DHY when compared with CON.

As anticipated, participants in this study lost more body mass during DHY (3.2% of body mass) than CON (0.4%), with the decrease during DHY being similar to that previously observed prior to competition in combat sports (Barley et al., 2017a). Physiological markers of hydration status (USG, urine osmolality, haematocrit, serum osmolality and body mass) indicated a greater level of dehydration immediately after DHY than CON (**Table 1**). *Ad libitum* fluid and food consumption resulted in body mass, serum osmolality and haematocrit returning to baseline, however, urine osmolality and USG did not (**Table 1**). These results are similar to those reported previously in research examining acute dehydration in combat athletes (Barley et al., 2017b) and indicate the complexity of assessing dehydration and the limitations of using a single (or even multiple) measure of hydration status (Armstrong, 2007). Nevertheless, and of significant practical importance, acute dehydration followed by a rehydration period has been found to impair exercise performance even when several markers of hydration have returned to baseline (Barley et al., 2017b).

While studies have indicated that dehydration may (Minshull and James, 2012; Rodrigues et al., 2014) or may not (Cheuvront et al., 2006; Kraft et al., 2012) compromise anaerobic performance, few studies have examined measures of central and peripheral neuromuscular function and performance following rehydration. In the present study we observed no differences in maximal strength between conditions at any timepoint (**Tables 2**, **3**). However, we did observe a decrease in muscular strength-endurance 3 h after the DHY protocol, as evidenced by a 28% decrease in the number of contractions performed at 85% of MVIC (g = 0.775). It is important to consider that while there were no differences between conditions in maximum MVIC torque or $T_{75\ ms}$ immediately post-exercise, less total work (contractions at 85% MVIC) was completed after DHY than CON. Thus, the ability to generate maximal force was impaired after DHY because the loss of force generating capacity was the same after DHY and CON despite fewer contractions being performed. The lack of change in maximal strength alongside the decrease in muscular endurance supports the findings of previous research (Montain et al., 1998; Bigard et al., 2001). Such results suggest that the use of acute dehydration weight loss strategies in weight-restricted sports can impair physical performance even up to 3 h after the intervention and therefore may not be an optimal strategy for competitive performance. These results may also be applicable to occupational or military settings where dehydration is likely to occur during normal operational duties. Indeed, it appears that in such settings *ad libitum* rehydration may be effective in stabilizing several markers of hydration, but some markers of hydration and aspects of physical function might also remain compromised for at least 3 h.

Despite the negative influence of dehydration on muscular strength-endurance, no changes in markers of central or peripheral neuromuscular function were observed. Indeed, no differences in central drive (i.e., $T_{75\ ms}$, VA% or VL or VM EMG/M; **Tables 2**, **3** and **Figure 1**) were observed 3 h after DHY or CON or immediately following the fatiguing exercise. Likewise, no differences in peripheral neuromuscular function (i.e., $T_{tw,p}$, $t_{tw,p}$ or $t_{1/2}$; **Tables 2**, **3**) were observed between conditions at any time point. This aligns with previous research which has shown no change in the neuromuscular function of hypohydrated athletes (Evetovich et al., 2002). However, the present study contributes to the body of research by demonstrating that the lack of neuromuscular changes remain persistent after 3 h. These results differ from previous studies investigating the influence of thermal exposure on

TABLE 4 | Mean (± SD) mood state before and immediately and 3 h post-intervention.

Mood state	Condition	Pre-intervention	Immediately post-intervention	3 h post-intervention
Tension	DHY	8 ± 2	10 ± 5	8 ± 4
	CON	7 ± 2	7 ± 1*	7 ± 2
Depression	DHY	9 ± 2	13 ± 6	11 ± 5
	CON	9 ± 2	8 ± 1*	8 ± 0.5
Anger	DHY	8 ± 2	13 ± 8	9 ± 4
	CON	8 ± 2	7 ± 1*	8 ± 1
Fatigue	DHY	7 ± 3	17 ± 6	11 ± 5
	CON	7 ± 4	6 ± 2**	6 ± 2*
Confusion	DHY	7 ± 4	10 ± 5	8 ± 3
	CON	7 ± 5	6 ± 3*	6 ± 2
Vigor	DHY	16 ± 5	11 ± 4	13 ± 6
	CON	16 ± 6	13 ± 5	13 ± 5

*$P < 0.05$, **$P < 0.01$ when compared with DHY in at the same time point.*

TABLE 5 | Mean (± SD) cognitive marker immediately and 3 h post-intervention.

Cognitive assessment	Condition	Immediately post-intervention	3 h post-intervention
Detection – speed (ms)	DHY	343 ± 64	363 ± 64
	CON	359 ± 90	353 ± 74
Detection – accuracy (%)	DHY	97 ± 3	95 ± 6
	CON	96 ± 3	97 ± 2
Identification – speed (ms)	DHY	536 ± 118	544 ± 112
	CON	518 ± 92	522 ± 115
Identification – accuracy (%)	DHY	96 ± 3	94 ± 7
	CON	96 ± 4	95 ± 7
One card learning – speed (ms)	DHY	865 ± 192	918 ± 224
	CON	909 ± 184*	871 ± 194
One card learning – accuracy (%)	DHY	68 ± 9	70 ± 14
	CON	75 ± 8*	74 ± 9
One back – speed (ms)	DHY	724 ± 148	703 ± 172
	CON	703 ± 172	724 ± 148
One back- accuracy (%)	DHY	90 ± 8	94 ± 5
	CON	94 ± 6	95 ± 4
Maze errors (n)	DHY	45 ± 16	49 ± 27
	CON	46 ± 17	46 ± 14

$P < 0.05$ when compared with DHY in at the same time point.

neuromuscular fatigue (Ross et al., 2011; Goodall et al., 2015), however, the present study did not measure neuromuscular function during thermal stress but instead following a recovery period. Additionally, no differences in TNAα were observed at any timepoint indicating a lack of DHY-induced tissue damage. While DHY did not appear to influence E-C coupling processes, we did observe a non-significant moderate effect (g = 0.643) in the 20:VFT 3 h after DHY but not following fatiguing exercise (**Table 2**), potentially indicating a decrease in calcium sensitivity 3 h after DHY (Binder-Macleod and Kesar, 2005; Nielsen, 2009). It is important to consider that we did not measure all components of neuromuscular function [e.g., motoneuron facilitation (Heckman et al., 2005; Heckman and Enoka, 2012)] so it is possible that some changes exist that may partly explain the loss of function. Nevertheless, the lack of effect of DHY on RFD, EMG/M and VA% (central drive) or in twitch properties or tetanic torques (peripheral function) strongly indicate a lack of DHY-induced decrease in neuromuscular function. Alternatively, other non-neural mechanisms such as an elevated core temperature, reduction in muscle glycogen or impaired cardiovascular function may explain the impairment in muscular performance but the lack of difference in thermal or cardiovascular markers (**Table 1**) alongside the nature of the fatiguing exercise used in the present study makes such explanations unlikely (González-Alonso et al., 1997; Casa, 1999; Barley et al., 2017b). In addition, previous research investigating the mechanisms behind acute dehydration induced fatigue has found it is not the result of H^+ or P_1 concentration (Montain et al., 1998). Clearly, further research is needed to determine the mechanisms responsible for the impairment in exercise performance resulting from acute DHY.

Another possibility is that an increase in mental fatigue (e.g., A psychobiological effect characterized by subjective feelings of "tiredness and "lack of energy") may explain the observed decrease in muscular strength-endurance (Marcora et al., 2009). Indeed, mental fatigue has been previously linked to decrements in exercise performance despite markers of central fatigue (i.e., central motor drive specifically) remaining unchanged (Marcora et al., 2009; Pageaux et al., 2015). Consistent with the hypothesis of mental fatigue influencing performance, the present results show that perceptions of tension, depression, anger, fatigue and confusion were all greater than CON immediately after DHY, however, only the perception of fatigue remained greater than CON after 3 h (**Table 4**). Therefore, it seems plausible that altered fatigue perception contributed to the decreased number in knee extensor contractions performed during fatiguing exercise (Marcora et al., 2009; Tucker, 2009). These results are consistent with previous research examining the effects of lower levels of dehydration (<2%) and the combination of dehydration and food restriction on mood (Hall and Lane, 2001). In addition to the change in mood we also found evidence of compromised cognitive function after DHY. Indeed, decision-making time and accuracy in one card learning were both reduced immediately after DHY (**Table 5**); however, these were not different at 3 h post-intervention. While previous research has linked decreased fluid ingestion and hyperthermia to changes in blood BDNF concentrations and cognitive function (Roh et al., 2017), we observed no differences in blood BDNF at any time point although there was a large inter-subject variability in the results throughout the study which limits what interpretations can be made from the data (**Table 1**). Therefore it is plausible that the decreases in cognitive function observed in this study did not result solely from physiological changes but rather an influence of mental fatigue on task engagement (Van der Linden et al., 2003). Further research that takes a system biological approach to investigate the potential for mental fatigue to contribute to the performance decrements resulting from acute dehydration is required.

We report that acute heat-induced dehydration of 3% body mass impairs muscular strength-endurance and increases perception of fatigue without detectibly influencing markers of central and peripheral neuromuscular function at 3 h after the weight loss intervention despite *ad libitum* consumption of food and fluids. Such findings potentially indicate that an increased mental fatigue resulting from acute dehydration, which persists even after fluid and food consumption are resumed, influenced the muscle work achievable before "fatigue." In addition, we provide evidence suggesting that athletes may not achieve adequate rehydration when allowed *ad libitum* fluid and food consumption following weight loss, as evidenced by a greater USG and urine osmolality 3 h after. A strength of this study was that a wide range of scientifically valid markers of central and peripheral neuromuscular function were utilized following a highly controlled dehydration protocol. However, the present study intentionally did not standardize food and fluid consumption during the recovery period to maximize ecological validity at the cost of developing a greater understanding of how recovery practices following acute dehydration may influence exercise performance. These findings outline the need for further research into the mechanisms by which acute dehydration impairs exercise performance, specifically regarding the relationship between acute dehydration and mental fatigue.

AUTHOR CONTRIBUTIONS

OB, CA, and DC conceived and designed the study. OB performed the experiments and analyzed the data. OB, CA, DC and AB interpreted the results of experiments. OB and CA drafted the manuscript. All authors edited and revised the manuscript before approving the final version.

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Improved Neural Control of Movements Manifests in Expertise-Related Differences in Force Output and Brain Network Dynamics

*Christian Gölz[1], Claudia Voelcker-Rehage[2], Karin Mora[3], Eva-Maria Reuter[4], Ben Godde[5], Michael Dellnitz[3], Claus Reinsberger[1] and Solveig Vieluf[1]**

[1] Institute of Sports Medicine, Paderborn University, Paderborn, Germany, [2] Institute of Human Movement Science and Health, Chemnitz University of Technology, Chemnitz, Germany, [3] Department of Mathematics, Paderborn University, Paderborn, Germany, [4] Centre for Sensorimotor Performance, School of Human Movement and Nutrition Sciences, The University of Queensland, Brisbane, QLD, Australia, [5] Department of Psychology & Methods, Jacobs University Bremen, Bremen, Germany

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Edited by:
Lars Donath,
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Anne Krause,
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Germany
Dennis Hamacher,
Universitätsklinikum Magdeburg,
Germany

***Correspondence:**
Solveig Vieluf
vieluf@sportmed.upb.de

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It is well-established that expertise developed through continuous and deliberate practice has the potential to delay age-related decline in fine motor skills. However, less is known about the underlying mechanisms, that is, whether expertise leads to a higher performance level changing the initial status from which age-related decline starts or if expertise-related changes result in qualitatively different motor output and neural processing providing a resource of compensation for age-related changes. Thus, as a first step, this study aims at a better understanding of expertise-related changes in fine motor control with respect to force output and respective electrophysiological correlates. Here, using a multidimensional approach, we investigated fine motor control of experts and novices in precision mechanics during the execution of a dynamic force control task. On the level of force output, we analyzed precision, variability, and complexity. We further used dynamic mode decomposition (DMD) to analyze the electrophysiological correlates of force control to deduce brain network dynamics. Experts' force output was more precise, less variable, and more complex. Task-related DMD mean mode magnitudes within the α-band at electrodes over sensorimotor relevant areas were reduced in experts, and lower DMD mean mode magnitudes related to the force output in novices. Our results provide evidence for expertise dependent central adaptions with distinct and more complex organization and decentralization of sensorimotor subsystems. Results from our multidimensional approach can be seen as a step forward in understanding expertise-related changes and exploiting their potential as resources for healthy aging.

Keywords: fine motor expertise, EEG, task-related brain activity, sensorimotor network, force control

INTRODUCTION

The dexterous use of hands, including the precise modulation of fingertip forces, is required for many tasks of daily living. Normal hand functioning is realized by an elaborate and highly automated, and therefore efficient, system of neuromuscular control (Vaillancourt and Newell, 2002; Vieluf et al., 2015). The ability to precisely modulate fingertip force decreases with advancing age, starting in early middle adulthood and continuing through middle and old age (Lindberg et al., 2009; Diermayr et al., 2011). This decline may result in increasing demands on sensory, motor, and cognitive systems to maintain fine motor abilities during work and leisure activities. On the contrary, continuous and deliberate practice, leading to a domain specific expertise, can induce positive plasticity resulting in better performance and more efficient information transmission (Rosenbaum et al., 2006; Callan and Naito, 2014). Therefore, expertise might have a potential to postpone age-related decline in fine motor control (Vieluf et al., 2012). Up to now, however, the underlying processes that characterize expertise and expert performance are not sufficiently understood. Thus, it is necessary to characterize expertise-related changes in a suitable and standardized research context before exploiting its potential in terms of counteracting age-related decline or strengthening compensatory resources. In this study, we compared fine motor experts with novices in a force control task. Force control tasks require the precise adaptation of fingertip forces under visual control, as required in work routines of precision mechanics. To better understand expertise-related differences, we investigated behavioral markers of force control and electrophysiological correlates characterizing the neural control of movements in fine motor experts in comparison with novices within the working age range.

There is no unified definition of expertise, which already points to challenges in the investigation of this phenomenon. Experts are people who repeatedly, and not accidentally, perform excellently in a specific field or domain (Ericsson, 2006). This refers to a highly domain specific characteristic, skill, or knowledge that allows an expert to be distinguished from a novice (Ericsson and Lehmann, 1996; Ericsson, 2006). Based on this specificity, it is generally difficult to differentiate between experts and nonexperts in a scientific laboratory setting (Ericsson, 2014). The challenge in investigating the phenomenon-inherent properties and processes is to create a context that sufficiently reflects the specific field of expertise (Ericsson and Lehmann, 1996) and at the same time allows to define meaningful and standardized markers that differentiate between experts and nonexperts. Tasks requiring fine motor control provide such a research context in a suitable way. Fine motor tasks can be easily implemented in a laboratory setting and allow to characterize fine motor experts' force control and its electrophysiological correlates. Until now, only few studies have examined fine motor experts in work-related contexts (Krampe and Ericsson, 1996; Law et al., 2004; Vieluf et al., 2012) such as the field of music (Krampe and Ericsson, 1996; Krampe et al., 2002). All studies showed a superior performance in fine motor control tasks in experts compared with novices. Using a force maintenance task, Vieluf et al. (2012) found experts' performance to be less variable, more precise, and the time for force initiation to be shorter. Moreover, there are first indicators of a more complex or less regular performance output measured as center of pressure (CoP) fluctuations (Schmit et al., 2005; Stins et al., 2009) and for force maintenance (Vieluf et al., 2018) in experts. The higher complexity allows experts to be more adaptive to changes in task and environment indicating a higher motion automation and therefore fewer attentional control (Stins et al., 2009; Vieluf et al., 2018). Beyond behavioral improvements, based on the adaptation mechanisms inherent in the development of expertise, structural and functional changes occur in the brain (Rosenbaum et al., 2006). An enlargement of task-specific cortical areas in reaction to motor training was determined in monkeys by Nudo et al. (1996) and confirmed for humans in professional musicians (Elbert et al., 1995; Meister et al., 2005; Bangert and Schlaug, 2006) and Braille readers (Sterr et al., 1998). From a functional point of view, based on brain imaging studies, experts' information processing seems more efficient (Callan and Naito, 2014; Debarnot et al., 2014) as reflected by more focused activation in task-relevant areas and higher suppression of task-irrelevant activities (Krings et al., 2000; Haslinger et al., 2004; Kim et al., 2008). On an electrophysiological level, increased efficiency was reflected by a reduction of cortical potentials in relevant areas in relation to the rest condition (Del Percio et al., 2008, 2009; Babiloni et al., 2010). Furthermore, neural efficiency was accompanied by a changed network characteristic. The functional organization of the expert's brain was characterized by a stronger focus on communication between relevant areas and the simultaneous suppression of irrelevant connections (Bernardi et al., 2013; Binder et al., 2017) leading to a more efficient use and integration of information gained from several networks (Vieluf et al., 2018). Using a force maintenance task, Vieluf et al. (2018) point to the opposing relationship between expertise- and age-related processes, for which compensatory over activation of brain areas was reported earlier (Ward and Frackowiak, 2003; Reuter-Lorenz and Park, 2010) but could not clearly differentiate these processes.

In summary, these previous results indicate that experts perform better (i.e., more precise), with a less variable force output, and their behavioral performance is more complex, and while their information processing is characterized by increased neural efficiency. While these findings are based on investigations of simple or closed movements (Del Percio et al., 2008; Babiloni et al., 2011; Vieluf et al., 2018), phases before the execution of movements (Del Percio et al., 2011) or measurements during rest (Babiloni et al., 2010), it still remains open how expertise is characterized during the execution of a more complex domain-specific task with regard to its performance and information processing.

In extension of Vieluf et al.'s (2018), we aimed to identify expertise-related changes in the neural control of movements by examining experts of fine motor control in the execution of a complex domain-specific task. We proposed a dynamic force control task as a domain-specific task for precision mechanics. We used force and electrophysiological data from the Bremen-Hand-Study@Jacobs (Voelcker-Rehage et al., 2013) and selected markers that comprehensively describe force output and reflect the different electrophysiological characteristics of neural efficiency. In accordance with the studies mentioned above (Krampe and Ericsson, 1996; Krampe, 2002; Law et al., 2004; Vieluf et al., 2012), we expected fine motor experts to perform better in a force control task than novices in this domain. More specifically, in line with the studies of Stins et al. (2009) and Vieluf et al. (2012), we expected a more precise, less variable but more complex force output from the experts. Based on the study conducted by Brunton et al. (2016), for the analysis of the electrophysiological data, we used dynamic mode decomposition (DMD) to extract spatiotemporally coherent patterns of the captured electrical fields that represent the dynamic network characteristics of brain activity. Here, we expected to find indicators of increased neural efficiency, especially a changed network behavior, in the expert group. More precisely, owing to the change in information processing in experts as predicted by the neural efficiency hypothesis (Callan and Naito, 2014), we assumed a more focused activity in task-specific sensorimotor areas. We also expected neural efficiency to be reflected in a stronger activation of a task-specific sensorimotor network whose internal communication is more focused and thus more centered. Similar to our previous findings for a force maintenance task (Vieluf et al., 2018), we assumed this to be reflected in lower DMD values over sensorimotor relevant areas. Additionally, we explored the relation between electrophysiological markers of neural efficiency and force output markers.

MATERIALS AND METHODS

This work is based on the Bremen-Hand-Study@Jacobs (Voelcker-Rehage et al., 2013), which aimed to characterize age- and expertise-related differences in fine motor control in a sample of novices and experts throughout the working age-range. The data presented in this paper were collected during the fourth session of this study. In the previous sessions, participants underwent a series of behavioral tests including measures of somatosensory performance (Reuter et al., 2012) and force control (Vieluf et al., 2012, 2013a,b).

Participants

Data from 47 participants were analyzed in this study. All participants had given their informed consent to the procedures before participating. Participants were recruited via diverse communication media (flyers, telephone calls, and newspaper announcements) and received a reimbursement of eight Euros per hour. The study was approved by the ethics committee of the German Psychological Society and was in accordance with the ethical standards laid down in the Declaration of Helsinki.

As mentioned above, participants were divided into fine motor experts (exp: n = 25; age = 50 $\pm$ 9 years; 13 females, MVC = 59.36 $\pm$ 21.94 N) and novices (nov: n = 22; age = 51 $\pm$ 9 years; 13 females, MVC = 52.16 $\pm$ 21.94 N) based on their occupation and years of experience in a job requiring fine motor skills. Novices were defined as people whose daily work routines were hardly influenced by fine motor skills such as service employees (i.e., consultants, office clerks, insurance agents, and vocational trainees in these occupations). The expert group was comprised of participants with more than 10 years of experience in a field with high demands for fine motor control, here precision mechanics (e.g., optician, dentists, goldsmiths, watchmakers). A 10-year inclusion criterion was chosen based on the study conducted by Ericsson and Smith (1991), and the daily use of hands in the work context was verified with a questionnaire on the daily use of hands (Trautmann et al., 2011). In addition, clinical manual dexterity was assessed by using the Purdue Pegboard test performed with the right hand and in accordance with the manual (Model 32020, Lafayette Instruments, Lafayette, IN, United States). A questionnaire on demographic status and health identified the participants as healthy and free of neurological restrictions and limiting injuries of upper extremities. All participants had normal or corrected to normal vision and hearing. All participants were right-handed, which was assessed by the Edinburgh Handedness Inventory (Oldfield, 1971). Each participant had conducted more than 200 trials of various force control tasks within experiments during the previous sessions of the Bremen-Hand-Study@Jacobs. Consequently, all participants were highly familiar with the setup and tasks.

Experimental Procedure

Prior to the measurements, the maximum voluntary contraction (MVC) of each test subject was determined using the peak force achieved out of three maximum precision grip trials, 5 s each with 2 min of rest in between them (Vieluf et al., 2013b). After completion of the electroencephalogram (EEG) setup, we recorded 30 s of EEG at rest while participants sat on the chair with their eyes open. This resting state measure is required to relate the following analyses to the task and to normalize the differences between test persons. Afterward, participants performed a dynamic force control task, with their dominant right hand. Participants' task was to match a curve that represented their applied force as precisely as possible to a target force presented on a screen at a distance of 80 cm (19″, frame rate 60 Hz). For this purpose, test persons sat on a chair with arms resting quietly on the armrests and thumbs and index fingers gripping a force transducer (Mini-40 Model, ATI Industrial Automation, Garner, NC, United States) fixated on the armrest. The target force changed over time in the form of a sinusoidal time curve, so that the test persons had to constantly adapt their force output with their thumbs and index fingers to the target value. The force time curve of the target curve averaged to 7 N (minimum: 2 N, maximum: 12 N) and was presented at a frequency of 0.5 Hz. Both the target and the force produced were displayed on the screen in front of the participants. The time axis (x-axis) covered 5 s. The force (y-axis) was presented in a range from 0 to 14 N. The target curve shifted from right to left, and the presentation of the produced force moved from left to right on the screen. In this setting, the participants always saw 1 s of the upcoming target force curve in advance and 4 s of the already exerted force and target curve (see **Figures 1B,C**). In order to ensure that the target force level could be reached as quickly as possible, the start of the target curve had been set in minimum (i.e., in 2 N). Participants performed seven trials of 30 s each. Familiarization was not carried out, as participants were already familiar with the setup and task from their previous visits. While performing the motor task, EEG was recorded.

Data Recording

Grip force data were recorded with the force transducer with a sampling rate of 120 Hz with an amplitude resolution of 0.06

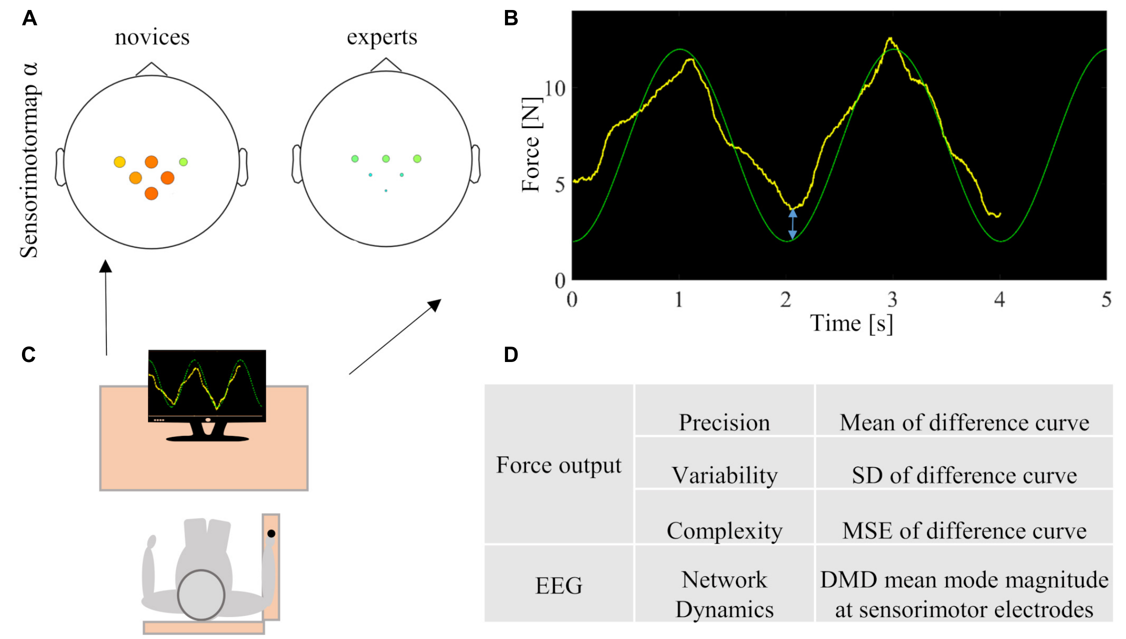

FIGURE 1 | Illustration of the multidimensional approach. **(A)** DMD mean mode extraction at sensorimotor relevant electrodes. **(B)** Analysis of the difference of target and applied force, indicated by the blue arrow. **(C)** Experimental setup. **(D)** Overview of force output and electrophysiological markers (SD = standard deviation, MSE = multiscale entropy, DMD = dynamic mode decomposition).

N using a customized LabView (National Instruments, Austin, TX, United States) program, which also provided online visual feedback on the screen.

Electroencephalogram data acquisition was done with a 32-electrode system with active electrodes (ActiveTwo, BioSemi, Amsterdam, Netherlands). The signal was recorded with a sampling rate of 2048 Hz and online band-pass filtered between 0.16 and 100 Hz. Electrodes were placed according to the 10–20 system (Jasper, 1958). In addition, the active common mode sense (CMS) electrode and the passive driven right leg (DRL) electrode were affixed next to Cz and used as reference and ground electrodes, respectively[1]. Vertical and horizontal eye movements as well as mastoid potentials were recorded with six facial electrodes designed for body-surface applications. Impedances were kept below 5 kOhm.

Data Analysis

For data analysis, MATLAB 2016b (MathWorks, Natick, MA, United States) and the additional EEGLAB package 14.1 (Delorme and Makeig, 2004) were used.

Analysis of Force Output

For analysis of the force data, only the z-component of the recorded force vector was further analyzed. Initially, the force data were filtered offline with a fourth-order lowpass butterworth filter with a cutoff frequency of 30 Hz. In order to exclude the initiation phase, the first 2 s of the force-time signals were excluded from further analysis. The analysis finally included the absolute mean difference (arithmetic mean of the deviation from the target force), the magnitude of variability (standard deviation of the deviation from the target force), and the complexity [multiscale entropy (MSE)], as described in the study conducted by Costa et al. (2005) (see **Figures 1B,D**).

Multiscale entropy allows to assess force output in the context of underlying neurophysiological processes, which can be assumed to indicate adaptability (Costa et al., 2005; Vieluf et al., 2015). In addition, MSE is the calculation of sample entropy values over several scales based on a coarse-graining procedure of the signal. As with this procedure, multiple coarse-grained time series are constructed by averaging the data points within non-overlapping windows of increasing length; there are different frequency ranges inherent in these time series. This allows to focus on relevant scales informing about process dependent changes in the signals' complexity (Morrison and Newell, 2012; Vieluf et al., 2015). To calculate the MSE, the vector length was fixed at 2, and the tolerance frame was 20% of the standard deviation of the signal (Costa et al., 2005; Vieluf et al., 2015). Based on the signal length and sampling rate, the entropy values of 60 scales were calculated. As an overall variable for the complexity of the signal, the arithmetic mean of the MSE values was determined over all scales (mean MSE). According to the inherent frequencies, the entropy values of functionally relevant scales were extracted based on the study conducted by Vieluf et al. (2015). These were scale 2 (inherent frequencies up to 30 Hz) representing the spectrum after filtering, scale 5 (inherent frequencies up to 12 Hz) representing the mechanisms of sensorimotor processing and physiological tremor (Elble and

[1]https://www.biosemi.com/faq/cms&drl.htm

Randall, 1976; Vaillancourt and Newell, 2003), and scale 15 containing frequencies up to 4 Hz most relevant for sensorimotor processing (Slifkin et al., 2000; Vaillancourt and Newell, 2003).

To identify outliers due to incorrect test execution, trials whose absolute mean force levels were below or above 2.5 times the standard deviation with regard to the group mean were excluded from further analysis (Frank et al., 2006; Vieluf et al., 2017). Thus, in the novice group, trials that were less than 4.7 N or more than 8.6 N in relation to the mean force were rejected, and in the expert group, trials that were less than 5.6 N or more than 7.6 N in relation to the mean force were rejected. In total, eight trials were excluded (experts: seven trials, novices: one trial). This included all seven trials of an expert who was consequently excluded from all further analysis.

Analysis of Electrophysiological Data

Electroencephalogram data were resampled to 200 Hz according to the Nyquist theorem and cut based on trial onset and length. Next, the data were re-referenced to the linked mastoids and band-pass filtered using an FIR filter (low cut off: 4 Hz, high cut off: 30 Hz). Later, the recordings were checked semiautomatically for artifacts. Signal components whose difference of maximum and minimum exceeded 120 μV within a window of 20 ms were marked as artifact. This criterion was chosen based on the common EEG analysis software (see BrainVisionAnalyzer, Brain Products, 12.2.5 – 4). Furthermore, the signals and markings were visually inspected by the authors of the study. Wrongly detected artifacts were discarded, and undetected artifacts were added. For further analysis, the time signals were divided into segments with a length of 0.5 s (100 data points). Segments in which artifacts were detected were rejected and remained unconsidered. Owing to a technical artifact, all trials of one subject (expert) had to be excluded from further analysis. On average, nine segments per test person (nov: 10, exp: 8) were rejected. Linear detrending had been applied to cleaned data to eliminate voltage shifts. In addition, the signals were amplitude normalized to reduce its contribution in further analysis.

We used the exact DMD algorithm described in Brunton et al.'s (2016), which was first proposed by Tu et al. (2014). This algorithm allows observation of the expression of the signals over the scalp detected by the EEG electrodes in relation to each other and thus to draw a conclusion on the dynamic network behavior of the brain. Herewith, not only the entire network behavior but also the activity of certain (sub-)networks, such as the sensorimotor network, can be approximated. In addition, DMD approximates the relationship between two data series X and X' in a time window. Time windows of 0.5 s length corresponding to 100 data points were analyzed. Linked spatial and temporal characteristics were approximated for each time window by a linear dynamical model given by $Y = \Phi \exp(\Omega t) z$, where Φ is the DMD mode matrix, Λ is a diagonal matrix with DMD eigenvalues along the diagonal from which $\Omega = \log(\Lambda)/\Delta t$ is obtained, t is time, $\Delta t = 0.005s$, and z is computed from the first data point x of X, that is, $x = \Phi z$. From the DMD eigenvalues, the oscillation frequency f was computed as $f = |\mathrm{imag}(\Omega)/(2\pi)|$. The mode matrix Φ represents the activation relationship between the electrodes for a particular frequency and indicates how much an electrode contributes to the dynamics of the network. The linear model illustrates how the spatial and temporal characteristics are linked. To increase the number of modes computed and thus the approximation accuracy, the delay embedding technique was applied, that is, data was stacked with a stacking depth of h. As a result of the error analysis on 100 randomly chosen windows of participant data, also described in Brunton et al.'s (2016), $h = 2$ was selected, as it revealed minimum error. The DMD analysis therefore was performed on an assembled data matrix, which stacked the first 99 data points on top of the last 99 data points. In addition, DMD mean mode magnitudes were calculated by averaging the mode magnitudes over all windows and associated with certain frequency ranges. A high DMD mode value indicates that the expression of the signal of a certain frequency recorded by each channel is high in relation to all other signals. By selecting the spatial distribution of the signal on seed electrodes, it is possible to draw conclusions about certain task-specific networks, such as the sensorimotor network. To characterize sensorimotor processes, we chose DMD mean modes at electrodes C3, C4, and Cz over sensorimotor regions. Furthermore, as Del Percio et al. (2011) and Binder et al. (2017) pointed out the importance of parietal areas in visuomotor tasks, we chose DMD mean modes at (centro-)parietal electrodes CP1, CP2, and Pz as seed electrodes and extracted the DMD mean modes, which are associated with the α- (8–12 Hz) and β- (12–30 Hz) frequency ranges. To obtain task-related values, the values of the rest condition were subtracted from the task condition in the same way as described in Brunton et al.'s (2016) (see **Figures 1A,D**). Finally, DMD mean modes of all valid trials were averaged per participant.

Statistical Analysis

The statistical analysis included the mean value of all valid trials (maximum seven) of all participants excluding the outliers. After data processing, this comprised 23 experts and 22 novices. Analyses were performed using SPSS statistics 22 (IBM, Armonk, NY, United States). Normal distribution was tested using the Shapiro–Wilk test. Screening results were compared among groups using t-test for two independent samples and Mann–Whitney U-test in case of violation of normal distribution.

Multivariate analysis of covariance (MANCOVA) with the between factor group (2; experts, novices) controlling for age and MVC was conducted to determine significant differences between experts and novices on the level of force output. For the analysis of the electrophysiological data, we added the within subject factor electrode (6; C3, C4, Cz, CP1, CP2, Pz) to the model. In case of violations of sphericity, the Greenhouse–Geisser adjustment was used, and corrected degrees of freedom and p-values were reported. Significant interactions and main effects were followed by Bonferroni corrected pairwise comparisons. In addition to normal distribution, we checked for homogeneity of error variances and covariance. Levene's test and Box's test showed no violation here (both $p > 0.05$). As analysis of variance was shown to be a robust statistical procedure in case of violation of normal distribution, especially with almost the same group sizes and group sizes over 10, we decided not to choose nonparametric methods in case of violation (Box, 1954; Glass et al., 1972; Schmider et al., 2010). Following Cohen (1988) and as suggested by Lenhard and Lenhard (2014), we considered effect size of $\eta_p^2 > 0.01$ to 0.06 as small effects (equivalent to Cohen's d of 0.2 to 0.4), $\eta_p^2 > 0.06$ to 0.14 as medium effects (equivalent to Cohen's d of 0.5 to 0.7), and $\eta_p^2 > 0.14$ as large effects (equivalent to Cohen's d of > 0.8).

To detect relationships between electrophysiological data and force output, variables of both levels were correlated using Pearson product-moment correlation, and in case of violation of the normal distribution Spearman rank correlation was used. False discovery rate (Benjamini and Hochberg, 1995) was used to correct the obtained p-values. As this analysis can be regarded as rather explorative, we report both uncorrected and corrected *p*-values. The correlation coefficients are judged according to the study conducted by Hopkins et al. (2009) with $r > 0.1$ to 0.3 indicating low, $r > 0.3$ to 0.5 indicating medium, $r > 0.5$ to 0.7 indicating strong, $r > 0.7$ to 0.9 indicating very strong, and $r > 0.9$ indicating perfect correlations.

RESULTS

MVC and Pegboard Results

A Mann–Whitney *U*-test indicated that the frequency of hand use was higher in experts (median = 35) than in novices (median = 16, $U = 12.5$, $p < 0.01$). No group differences were detected in the MVC values (exp: median = 52.63 N, nov: median = 49.18 N, $U = 196.00$, $p = 0.20$) and the pegboard results [exp: $M = 15.51$, $SD = 1.70$, nov: $M = 15.1$, $SD = 1.71$, $t(43) = 0.86$, $p = 0.40$].

Force Output Results

Force output results are illustrated in **Figure 2**. Analyses revealed significant differences between experts and novices. Precision was higher, as indicated by lower mean differences, in the expert group ($M = 0.97$ N, $SD = 0.20$ N) than in the group of novices [$M = 1.18$ N, $SD = 0.38$ N; $F(1,41) = 4.56$, $p = 0.04$, $\eta_p^2 = 0.10$]. Experts' force output was less variable ($M = 1.23$ N, $SD = 0.40$ N) than that of novices [$M = 1.23$ N, $SD = 0.40$ N, $F(1,41) = 4.80$, $p = 0.03$, $\eta_p^2 = 0{,}11$]. Differences in the complexity of the force output were found between the experimental groups. Experts' MSE was higher overall [exp: $M = 1.65$, $SD = 0.61$, nov: $M = 1.61$, $SD = 0.11$, $F(1,41) = 4.64$, $p = 0.04$, $\eta_p^2 = 0.10$] and on scale 15 representing sensorimotor processes [exp: $M = 1.43$, $SD = 0.10$, nov: $M = 1.36$, $SD = 0.61$, $F(1,41) = 4.29$, $p < 0.05$, $\eta_p^2 = 0.10$], while no significant differences were observed for MSE scale 2 representing the spectrum after filtering [exp: $M = 0.41$, $SD = 0.09$, nov: $M = 0.41$, $SD = 0.09$, $F(1,41) = 0.001$, $p = 0.91$, $\eta_p^2 = 0.00$] and scale 5 representing the mechanisms of sensorimotor processing and physiological tremor [exp: $M = 0.71$, $SD = 0.09$, nov: $M = 0.41$, $SD = 0.11$, $F(1,41) = 0.15$, $p = 0.67$, $\eta_p^2 = 0.004$].

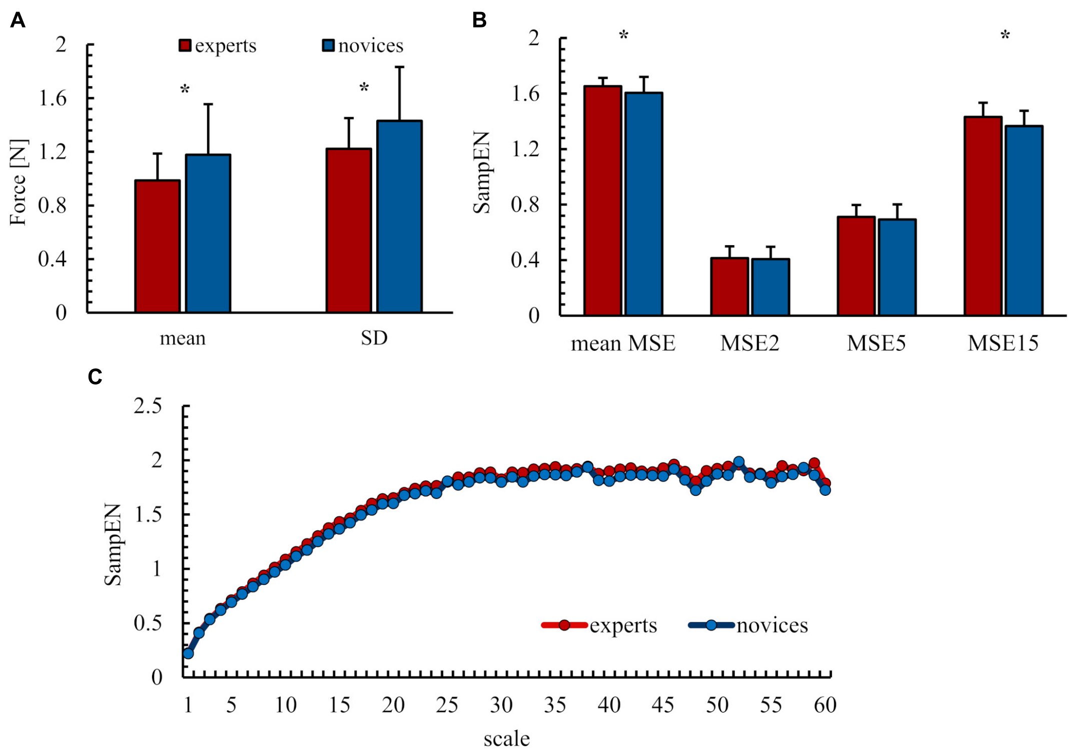

FIGURE 2 | Behavioral results **(A)** mean difference (mean) and variability (SD). **(B)** Entropy values as mean over all scales and of chosen scales 2, 5, and 15. **(C)** MSE curve of entropy values over all scales. All values as group mean and standard deviation. * Indicates significant differences.

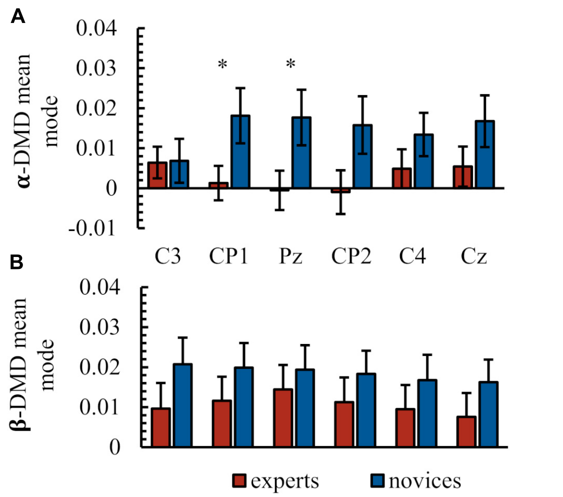

FIGURE 3 | Electrophysiological results. Task-related DMD mean mode at the electrodes of interest C3, CP1, Pz, CP2, C4 in the α-frequency band **(A)** and in the β-frequency band **(B)**. All values as group mean and standard error. * Indicates significant differences.

Electrophysiological Results

Figure 3 and **Table 1** summarize the electrophysiological results. Task-related DMD mean modes of seed electrodes representing sensorimotor network activity were compared between groups. In the α-frequency band, statistical analysis revealed neither a significant main effect of group [$F(1,41) = 2.99$, $p = 0.09$, $\eta_p^2 = 0.07$] nor of electrode [Greenhouse–Geisser: $F(3.38,138.71) = 0.47$, $p = 0.72$, $\eta_p^2 = 0.01$]. A significant interaction between electrode and group was found in the task-related DMD mean modes in the α-frequency [Greenhouse–Geisser: $F(3.38,138.71) = 3.72$, $p = 0.01$, $\eta_p^2 = 0.09$]. *Post hoc* comparisons revealed significant differences between groups in task-related DMD mean mode magnitudes at CP1 ($p = 0.05$) and Pz ($p = 0.04$) and marginally significant differences at CP2 ($p = 0.06$) with lower values in the expert group (see **Figure 2** and **Table 1**). No significant group differences were present in α- task-related DMD mean modes at central electrodes (C3: $p = 0.88$, C4: $p = 0.21$. Cz: $p = 0.14$). In the β-frequency band, statistical analysis revealed no significant main effect of group [$F(1,41) = 1.54$, $p = 0.22$, $\eta_p^2 = 0.04$] and a significant main effect of electrode [Greenhouse–Geisser: $F(2.89,118.54) = 3.03$, $p = 0.03$, $\eta_p^2 = 0.07$]. *Post hoc* comparisons revealed no significant differences between the electrodes here. No significant interaction between electrode and group was found in the task-related DMD mean modes in the β-frequency [Greenhouse–Geisser: $F(2.89,118.54) = 1.05$, $p = 0.37$, $\eta_p^2 = 0.03$].

Results of the Correlation Analysis

Significant positive correlations were found in the novice group for the correlation between task-related DMD mean mode in the α-frequency range at C3, CP1, and Pz and and mean difference [C3: $r = 0.49$, $p = 0.02$ ($p_{cor} = 0.12$); CP1: $r = 0.52$, $p = 0.01$ ($p_{cor} = 0.12$); Pz: $r = 0.49$, $p = 0.02$ ($p_{cor} = 0.12$)], SD [C3: $r = 0.58$, $p < 0.01$ ($p_{cor} = 0.06$); CP1: $r = 0.62$, $p < 0.01$ ($p_{cor} = 0.06$); Pz: $r = 0.58$, $p < 0.01$ ($p_{cor} = 0.06$)], and mean MSE [C3: $r = -0.47$, $p = 0.03$ ($p_{cor} = 0.12$); CP1: $r = -0.46$, $p = 0.04$ ($p_{cor} = 0.14$); Pz: $r = -0.47$, $p = 0.03$ ($p_{cor} = 0.12$)]. Furthermore, a significant positive correlation between task-related DMD mean mode in the β-frequency range at C4 and SD was found [$r = 0.49$, $p = 0.02$ ($p_{cor} = 0.12$)]. In the expert group, only β DMD mean mode at CP1 and MSE at scale 5 correlated significantly [$r = -0.42$, $p = 0.05$ ($p_{cor} = 0.14$)].

DISCUSSION

Based on the potential of domain specific expertise for delaying age-related decline in fine motor control, we aimed to identify expertise-related changes in the structure of the force output and the respective neural control processes. In an effort to meet the challenges in the investigation of expertise effects, we chose a dynamic force control experiment conducted in the context of the Bremen-Hand-Study@Jacobs (Voelcker-Rehage et al., 2013). The force control task resembles the dynamic grasping pattern frequently used by precision mechanics as part of their daily work routines. In summary, experts in this study showed a more precise and less variable force output. In addition, MSE analysis further revealed higher complexity of motor control output in experts. We further applied DMD to detect spatiotemporally coherent patterns within the EEG and found lower DMD values over sensorimotor relevant areas, which are indicators of different (i.e., more efficient) activities within the sensorimotor network for experts and novices in the α-frequency.

Characteristics of Experts' Force Output

In order to reveal differences in force control, the classical measures of motor performance, precision (absolute mean difference), and variability (standard deviation of the deviation from the target force) were applied. Similar to results gained in fine motor experts performing a finger tapping task (Krampe and Ericsson, 1996) or a task specific to the profession of surgeons (Law et al., 2004) and especially to findings from a static force control task using the same participants (Vieluf et al., 2012), experts were superior to novices in the dynamic force control task. This was reflected in a smaller deviation from the target force and a less variable performance, that is, the smaller amplitude in force fluctuations of the expert group.

In order to infer the underlying organization of the sensorimotor system, we further considered the complexity of the force output (MSE). Compared with novices, experts showed a more complex force output averaged over all scales (mean MSE) and particularly on the scale related to sensorimotor processes (MSE 15) but not on the scale related to tremor frequencies (MSE 5). Thus, extensive practice seems to alter neural processes of motor control but not general age-related characteristics. With this finding, we were able to, for the first time, describe the complexity of the force output in a dynamic force control task in the context of expertise and were thus able to extend the findings of a force maintenance task (Vieluf et al., 2018). Previous studies in healthy older adults and patients postulated the

TABLE 1 | TR DMD mean mode magnitudes at electrodes of interest in the α-frequency and β-frequency.

	α-Band					
	CP1*		CP2		Pz*	
	Mean	*SD*	Mean	*SD*	Mean	*SD*
Experts	0.0013	0.0207	−0.0010	0.0262	−0.0005	0.0235
Novices	0.0181	0.0330	0.0158	0.0344	0.0177	0.0334
	C3		C4		Cz	
	Mean	*SD*	Mean	*SD*	Mean	*SD*
Experts	0.0064	0.0190	0.0049	0.0233	0.0054	0.0241
Novices	0.0068	0.0264	0.0134	0.0260	0.0168	0.0311
	β-Band					
	CP1		CP2		Pz	
	Mean	*SD*	Mean	*SD*	Mean	*SD*
Experts	0.0097	0.0307	0.0116	0.0288	0.0144	0.0293
Novices	0.0207	0.0319	0.0199	0.0295	0.0194	0.0292
	C3		C4		Cz	
	Mean	*SD*	Mean	*SD*	Mean	*SD*
Experts	0.0113	0.0295	0.0095	0.0290	0.0076	0.0284
Novices	0.0183	0.0278	0.0168	0.0304	0.0163	0.0271

**Indicates significant group difference.*

loss of complexity hypothesis (Vaillancourt and Newell, 2002). According to this hypothesis, a decline in complexity is present with advancing age and diseases (Lipsitz and Goldberger, 1992; Vaillancourt and Newell, 2002). The higher complexity found in experts in this work would follow the principle of this hypothesis in the opposite way. Similarly, studies on force maintenance (Vieluf et al., 2018) and on postural control (Schmit et al., 2005; Stins et al., 2009) found higher complexity of the CoP oscillation pattern in experts compared with novices. A higher complexity in experts could be an indicator for greater motion automation (Stins et al., 2009). In other words, experts may need less attention and therefore less mental resources to accomplish the task. This would correspond to a higher efficiency in movement control. Furthermore, the increased complexity in experts could indicate a greater movement flexibility, as postulated by Schmit et al. (2005). Overall, we were able to confirm a superior performance of the expert group using the classical measures, precision, and variability. More interestingly, by assessing the complexity of behavioral performance via MSE, we provide first indications of a changed organization of sensorimotor control in experts. Such expertise-specific reorganization might allow for more adaptability when performing tasks.

Electrophysiological Markers of Experts' Force Control

To investigate electrophysiological correlates of sensorimotor processes, we used DMD to capture spatiotemporal patterns at sensorimotor relevant electrodes. Experts showed lower DMD mode magnitudes with significant differences in the (centro-) parietal electrodes in the α-frequency band. As DMD modes reflect the relation between all (sensorimotor relevant and irrelevant) electrodes, a lower DMD mode in the relevant frequencies above the sensorimotor areas could indicate a more focused activity within a sensorimotor network. This would further suggest a high specialization of this network. Owing to the localization of the seed electrodes over sensorimotor relevant brain areas, the α-frequency could be interpreted here as motor related (Pineda, 2005). Consequently, the results at this frequency could be considered in the context of a more efficient translation of sensory information into motor information modulated by a stronger sensorimotor network (Pineda, 2005). The importance of integration and conversion of sensory information could be reflected in the rather parietally localized differences between experts and novices. This could further suggest a higher network efficiency of the sensorimotor network in the expert group, especially regarding the processing of sensory information, which could be reflected in higher motion automation as described above. The force control task required the sensorimotor integration of visual information with force output. Participants were strongly dependent on visual feedback but could also use feedforward control based on the 1 s target preview. These findings are consistent with the study conducted by Binder et al. (2017), who described a stronger sensorimotor functional network in experts during the execution of various visuomotor tasks and emphasized the importance of the parietal areas herewith. Del Percio et al. (2011) also illustrated the coupling of parietal regions during the preparatory phase in shooting. Thus, it is conceivable that experts more effectively integrate visuomotor information. We failed to find any expertise-related effects for the β-DMD mean modes. The participants in our study came to the lab and performed force control tasks for the fourth time so that all participants (experts and novices) might have partly automatized the execution of the task and thus differences in the β-band might be reduced. While the superior behavioral performance of experts suggests that force control tasks are sensitive to expertise effects, the task may not have completely represented the respective expertise context (i.e., force modulation requirements at work). This might have influenced these results. Consequently, it is possible that expertise effects were lower than they would have been in even more specific tasks. Finally, a high between-subject variability could also be observed on the electrophysiological level in both frequency bands, which suggests a high individuality of the sensorimotor network.

Combined Reflection of Electrophysiological and Force Output Markers

The explorative correlation analysis revealed that for the novices, but not the experts, lower α-DMD mean modes at C3, CP1, and Pz were associated with the less variable and more precise force output. Furthermore, a lower mean MSE was associated with higher α-DMD mean modes at these electrodes in novices. In the β-band we further found that lower DMD mean modes at C4 were associated with a lower variability of the force output. In the expert group, there was only a negative correlation between the β-DMD mean mode at CP1 and the MSE of scale 5: the lower the β-DMD mean mode, the higher the MSE of scale 5. Speculatively stated, there could be a connection between the dynamic network characteristics (neural efficiency) and the performance level in the group of novices. Babiloni et al. (2011) found similar associations between the coupling of the electrodes in the α-band over sensorimotor relevant parietal areas and performance. Although force control is an expression of many different internal processes, a more efficient execution (lower variability and higher precision) could be partially reflected in altered brain activity patterns indicating neural efficiency. On the other hand, there are no results in the expert group that point to a simple relationship between force control and electrophysiological markers. Rather, this could indicate a more complex interaction of central and decentralized subsystems, which could also be reflected in a higher complexity of the force output, especially on the sensorimotor scale (MSE 15). This points to the importance of multidimensional approaches in the analysis and characterization of expert performance.

Taken together, the force output and electrophysiological data confirm that continuous and deliberate practice at work leads to domain specific plastic changes of the fine motor control system. Alterations of the neuromuscular control are opposing the commonly observed changes with aging, that is, increase in error and variability as well as loss of complexity. In addition, the interpretation of the electrophysiological findings is in line with the neural efficiency hypothesis that experts recruit smaller and more specific networks, opposing the changes of brain activity with increasing age. The dedifferentiation hypothesis states that with increasing age a loss of specificity occurs. Thus, structures and mechanisms that are specialized in young adults become less distinct or common to different functions in older age (Baltes and Lindenberger, 1997; Reuter-Lorenz and Park, 2010). The reversal effects on multiple levels suggest that continuous and deliberate practice has the potential to postpone or counteract age-related declines. These results might offer a foundation to design targeted interventions aiming to counteract age-related losses. Correlations between force output and electrophysiological markers were only present in the group of novices. Potentially, this indicates a more complex interaction between central and decentral systems in experts.

Methodological Considerations

This study can only provide first insights into expertise-related processes of the neuromuscular system as it is a cross-sectional study. A longitudinal study would expand the findings here and help to gain knowledge of how expertise is impacted by its development and maintenance into older age to elucidate the power of expertise in the context of aging. The chosen laboratory context and task can be considered as a suitable context since group differences are present in the force control task but not in the pegboard test. We used fixed force levels instead of making the force requirements relative to the MVC of the test persons. This had the advantage of mapping different requirements of the everyday task context and thus created a research context more similar to the expertise context. Nevertheless, the relatively low force levels could have had the disadvantage of different strength requirements for the participants. Furthermore, it should be noted that the participants here had already participated in three force control experiments, which could have had an influence on our results. Moreover, in this work, a resting measurement with open eyes was chosen as a baseline for the EEG signal. Such a baseline generally has the disadvantage of a higher exposure of the signals to artifacts caused by eye movements and blinking, which were removed during pre-processing. However, visual stimuli may have had a possible influence on resting activity. Nevertheless, such a rest condition was used in this study to weaken the effect of a dominant visual (sub-) network, which we expected to be engaged in the task. Therewith, we aimed to ensure that the examined sensorimotor network characteristics reflected a higher relation to the task itself.

In addition to traditional markers on the level of force output, we used the nonlinear method to sample entropy on different time scales. A general drawback of such methods is the dependency of input parameters (e.g., vector size, tolerance frame). As we chose these parameters in line with previous studies (Costa et al., 2005; Vieluf et al., 2015), we assume that our choice is valid. The main limitation for the interpretation of the electrophysiological data however is the small number of electrodes (n = 32) and the restriction to signal space. Thereby, the interpretation of the decrease of DMD mean modes as a decoupling remains speculative, though consistent with the literature. Increasing the number of channels, MRI co-registration, and transferring the signals into source space would overcome the general low spatial resolution of EEG in order to sharpen the results especially with regard to the sensorimotor regions.

At last, it should be noted again that the correlation analysis is explorative and the interpretation is based on the uncorrected p-values. Therefore, interpretation of the correlation results should be done with caution.

CONCLUSION

Here, we could confirm experts' performance to be more precise, less variable, and more complex, pointing to a superior performance and changed organization of sensorimotor control. The latter idea is supported by the finding that complexity is higher for the MSE scale representing sensorimotor processing but not for tremor. Electrophysiological correlates of force control further indicate that information processing might be more efficient in experts compared with novices. However, only in novices, we found a directional relationship between network characteristics and force output. This points to the importance of examining expertise with comprehensive multidimensional approaches. In summary, this study extends the knowledge in the field of expertise. Understanding the changes related to continuous and deliberate practice provides important insights into the characteristics of a fully developed expertise.

Considering these characteristics (i.e., neural efficiency, higher complexity) in connection with results from aging research suggests that expertise could be taken up as an opponent to age-related changes. Nevertheless, it still remains open whether expertise-related specificity and efficiency can be transferred to non-expertise tasks and whether expertise-related changes tend to favor reverse effects or the development of compensational resources for age-related decline. We suggest that further investigations are needed to understand how and to what extent age-related changes can be affected by continuous and deliberate practice. This study could provide a starting point here.

AUTHOR CONTRIBUTIONS

E-MR, CV-R, BG, and SV set up the experiments. CG, E-MR, CV-R, BG, and SV were involved in the conception of the work. E-MR and SV collected data. CG and KM analyzed data. All authors interpreted results, drafted parts of the work, approved the final version of the manuscript, and agreed to be accountable for all aspects of the work.

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Targeted Athletic Training Improves the Neuromuscular Performance in Terms of Body Posture From Adolescence to Adulthood – Long-Term Study Over 6 Years

Oliver Ludwig[1], Jens Kelm[2], Annette Hammes[3], Eduard Schmitt[4] and Michael Fröhlich[1]*

[1] Technische Universität Kaiserslautern, Fachgebiet Sportwissenschaft, Kaiserslautern, Germany,
[2] Chirurgisch-Orthopädisches Zentrum, Illingen, Germany, [3] MEDICOVER GmbH, Saarbrücken, Germany,
[4] Universitätsklinikum des Saarlandes, Klinik für Orthopädie und Orthopädische Chirurgie, Homburg, Germany

Poor posture in childhood and adolescence is held responsible for the occurrence of associated disorders in adult age. This study aimed to verify whether body posture in adolescence can be enhanced through the improvement of neuromuscular performance, attained by means of targeted strength, stretch, and body perception training, and whether any such improvement might also transition into adulthood. From a total of 84 volunteers, the posture development of 67 adolescents was checked annually between the age of 14 and 20 based on index values in three posture situations. 28 adolescents exercised twice a week for about 2 h up to the age of 18, 24 adolescents exercised continually up to the age of 20. Both groups practiced other additional sports for about 1.8 h/week. Fifteen persons served as a non-exercising control group, practicing optional sports of about 1.8 h/week until the age of 18, after that for 0.9 h/week. Group allocation was not random, but depended on the participants' choice. A linear mixed model was used to analyze the development of posture indexes among the groups and over time and the possible influence of anthropometric parameters (weight, size), of optional athletic activity and of sedentary behavior. The *post hoc* pairwise comparison was performed applying the Scheffé test. The significance level was set at 0.05. The group that exercised continually (TR20) exhibited a significant posture parameter improvement in all posture situations from the 2nd year of exercising on. The group that terminated their training when reaching adulthood (TR18) retained some improvements, such as conscious straightening of the body posture. In other posture situations (habitual, closed eyes), their posture results declined again from age 18. The effect sizes determined were between $\eta^2 = 0.12$ and $\eta^2 = 0.19$ and represent moderate to strong effects. The control group did not exhibit any differences. Anthropometric parameters, additional athletic activities and sedentary behavior did not influence the posture parameters significantly. An additional athletic training of 2 h per week including elements for improved body perception seems to have the potential to improve body posture in symptom free male adolescents and young adults.

Keywords: posture analysis, posture training, adolescence, maturation, posture development, body perception

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****Correspondence:***
Oliver Ludwig
oliver.ludwig@sowi.uni-kl.de

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INTRODUCTION

Upright posture is the result of a complex interaction of skeleton, musculature, and central nervous system (CNS) (Assaiante et al., 2005). Neuromuscular performance determines the quality of posture and movement control, which is key for mastering daily routine tasks and athletic activity (Punakallio, 2003). It changes in the course of time, which is based on increasing muscular weakness and control deficits in the CNS. Already in childhood and adolescence, deficits in posture control have an effect, usually in the form of weak posture. Typical weaknesses are, for example, lumbar hyperlordosis, hunchback, protracted shoulders, and protruded head. Depending on the definition of posture weaknesses, literature states their prevalence at 22–65% for 10- to 18-year-old children and adolescents (Kratenova et al., 2007; Gh Maghsoud et al., 2012; Wirth et al., 2013; Lee, 2016).

The connection between posture weaknesses and the occurrence of complaints in adolescence or in the course of adulthood has not been clarified to date. However, more and more studies suggest a link to back and neck pain during adolescence and in the course of later development (Dolphens et al., 2015, 2016; Noll et al., 2016). It is assumed that posture weaknesses may lead to a biomechanically unfavorable strain of tendons and joints, and that corresponding adaptations result in asymmetrical muscle activity, which, in turn, leads to muscular problems (Bruno et al., 2012).

If posture weakness in adolescents could lead to problems in adulthood, early intervention is required. Since posture weakness is often accompanied by weak muscle function (Buchtelová et al., 2013), the usual intervention - depending on the severity of problems – is physical therapy, rehabilitation sport, or a recommendation to increase athletic activity in general (Kim et al., 2015).

This recommendation becomes ever more important because the lifestyle of children in industrialized nations has changed to be more sedentary and less active (Claus et al., 2009; Drzal-Grabiec and Snela, 2012; Shan et al., 2014). Therefore, targeted athletic activity seems to be a suitable means to preventing posture problems. However, we need to critically question (i) whether targeted athletic activity in adolescence operates as a preventive or corrective factor in terms of posture deficits, and (ii) whether athletic activity in adolescence has positive effects reaching all the way into adulthood. According to current knowledge, targeted strength training can be performed already in childhood and adolescence [overview in (Matos and Winsley, 2007)].

The extent to which targeted posture training in adolescence carries positive effects into adulthood, i.e., the question whether it basically lays the foundation for a stable body posture, has not been clarified to date. Therefore, this study meant to verify whether a targeted training program in adolescence can carry positive effects into adulthood. The following hypotheses were to be analyzed:

(i) Targeted posture training improves selected posture parameters in adolescence.

(ii) Posture training performed on a regular basis in adolescence continues to have positive effects on body posture in adulthood.

MATERIALS AND METHODS

Sample

The study was carried out within the framework of an interdisciplinary research project (Kid-Check). As the primary goal was to examine intra-individual changes during a perennial training program, the minimal sample size was calculated with G*Power 2.1 (University Kiel, Germany) with alpha = 0.05, power = 0.95, effect size = 0.5, based on matched pairs T-test. We calculated a total sample size of 45, but increased this due to an expected high number of dropouts.

Between 2001 and 2018, a total of 84 adolescent male test persons with poor posture participated in the study (see flow chart, **Figure 1**). The adolescents (please refer to **Table 1** for anthropometric data) entered the study at age 14, and 67 of them were examined annually until the age of 20. The key criterion to be included in the study was a significant posture problem, defined by means of a posture index pertaining to habitual posture > 1.35 [**Figure 2** (Fröhner, 1998; Ludwig et al., 2016c)]. Exclusion criteria were acute complaints pertaining to the postural and musculoskeletal system, pathological changes in the spine, a BMI > 24 or an intensive (>3 h per week) additional athletic activity.

Twenty-three of the 84 test persons served as a control group. They were not randomly chosen. Instead, we added all persons to the control group who could not guarantee a regular participation in a two-times-per-week training. The main reasons were: too little time (caused by school, N = 4, caused by leisure activities, N = 5) and logistics problems regarding their transportation to the training location (N = 12). Two participants were not willing to perform a regularly strength training.

The later division of the training group at age 18 was not random, either. Initially, we planned to terminate the study when the participants came of age, but later decided to continue for two further years, if a sufficient number of participants would continue. The 28 test persons who stopped training did so for different reasons: too little time caused by school or studies (N = 16), changing their place of residence due to university studies (N = 6), having reached their personal fitness goals (N = 4), and physical injury (N = 2). According to them, none of them stopped because they had lost their interest in strength training or due to lack of motivation.

TABLE 1 | Anthropometric data and leisure behavior of the test groups at the start of the study.

	Mass (kg)	Height (cm)	Athletic activity (h/week)	Sedentary behav. (h/week)
TR20	60.3 ± 4.96	171.0 ± 5.49	1.87 ± 1.42	38.54 ± 5.32
TR18	60.9 ± 5.00	171.8 ± 4.28	1.86 ± 1.51	37.94 ± 4.50
CON	60.5 ± 5.76	171.6 ± 6.03	1.90 ± 0.87	39.37 ± 4.87

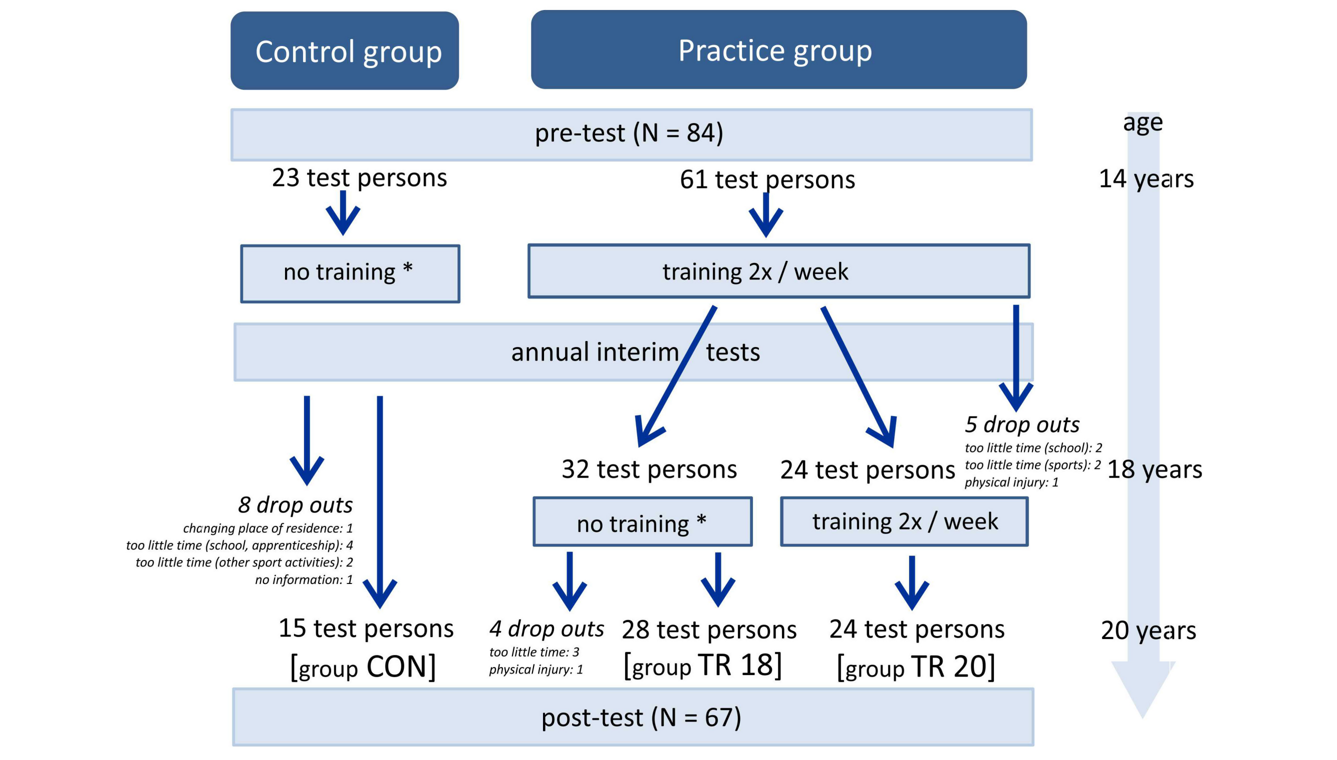

FIGURE 1 | Study flow chart. *Reasons see text.

During the 6 years, a total of 17 persons dropped out at different times. Their reasons for leaving the study are indicated in **Figure 1**. From that point onward, they were no longer available for any subsequent tests.Nevertheless, in order to reduce bias, we still included their data in the analysis up to the date of them having left the study.

The study's concept was based on the Helsinki declaration and conducted accordingly (World Medical Association, 2013). The university's ethics commission had approved the study (ref. no. 15-6). The test persons and their parents were informed on the order of the study and the content of the training and gave their written consent.

Posture Analysis

Since body posture is a complex phenomenon it was operationalized through a sum parameter [posture index (Fröhner, 1998; Ludwig et al., 2016b)] in three different posture situations in order to provide an overview of the general posture control (Woollacott and Shumway-Cook, 1990; Assaiante et al., 2005; Maurer et al., 2006; Ludwig et al., 2016d).

The following posture situations were registered for posture identification and comparison at an examination day once a year: the habitual, relaxed position (HAB), the active, upright position (ACT), and the active upright position with closed eyes (ECL) (Ludwig et al., 2016d). While active posture can be reproduced validly (Ludwig et al., 2016b), habitual posture changes during the day. To improve internal validity, all measurements were taken at the same time in the morning without any previous intensive physical activity, and followed a standardized test protocol (Ludwig et al., 2016d).

To determine the posture parameters, posture photographs in the sagittal plane were taken of the adolescents wearing swimwear or underwear. High-contrast marker balls were attached to anatomic landmarks as the caudal tip of the sternum, the point of the strongest lumbar lordosis, the point of the strongest thoracic kyphosis, as well as the spina iliaca anterior superior. A camera was mounted on a tripod at hip height (Olympus SP510UZ and Nikon Coolpix S33) and posture photographs (resolution 2304 pixels × 3072 pixels) of all three posture situations were taken in front of a calibration wall.

The adolescents first stood relaxed, feet at shoulder width, arms hanging down loosely, view straight ahead. A sideways photograph was taken (HAB). The adolescents were then instructed to actively change into an upright position without holding their breath. A second posture photograph was taken (ACT). The adolescents were then instructed to close their eyes while maintaining their active posture. After 60 s, the third photo was taken (ECL). The instructions were standardized and no optical or acoustic disruptive impulses occurred. All posture analyses were performed by an experienced researcher who was blinded to the group membership of the participants.

The horizontal distances between marker points and the perpendicular through the malleolus lateralis were calculated using the Corpus concepts® software (Fa. AFG, Idar-Oberstein, Germany), and the posture index was calculated based on those results. The posture index is a complex parameter that summarily evaluates the posture quality of the trunk (for details please see **Figure 2**). Values between 1.0 and 1.3 stand for a stable posture (Fröhner, 1998; Ludwig et al., 2016c). The test quality of this parameter has been confirmed in other studies (Ludwig et al., 2016b). The advantage of this parameter for the assessment of body posture is that it combines numerous individual posture parameters into one numerical value. Current studies show that the "global" posture parameters are associated with complaints, while "local" parameters, such as the pelvic angle, do not provide clear information (Dolphens et al., 2015).

Posture Training

From the start of the study, the 61 adolescents exercised for 60 min twice a week in a gym under qualified supervision. After a 6-min warm-up phase on a treadmill, they performed the strength-endurance exercises in the form of a set-of-3 training at the device (15 repetitions, 1 min break, see **Table 2**). During the 1st months, the participants familiarized themselves with the training devices. The weight load was kept low and the participants' awareness was led to focus on proper movement. The weights were chosen and adapted during the study so that the participants were able to carry out the 3 sets while adhering to the correct movement technique, but feeling definitely exhausted subjectively afterwards (Borg scale 7 of 10). The stretch exercises (active antagonist contract stretching) were executed three times each for 30 s for each side of the body (Youdas et al., 2010). As the main reason for poor posture is found in poor motor skills and weakness of the supporting musculature, e.g., due to sedentary day-to-day school life, neuromuscular performance can be improved via multiple approaches:

- Strengthening the weak core muscles (strength endurance)
- Improving the range of motion (ROM) of the movement-limiting muscles (mobility/flexibility)
- Improving the sensory-neuromuscular coordination.

The focus of the posture training therefore lied in three dimensions (**Table 2**):

(1) Strengthening the muscle groups that straighten up the pelvis (in particular, m. rectus abdominis, m. obliquus, hamstrings, and m. glutaeus maximus) because they are able to effect an active retroversion of the pelvis (Buchtelová et al., 2013; Jeong et al., 2015).
(2) Stretching the muscles involved in the forward tilt of the pelvis (m. iliopsoas, m. rectus femoris) in order to increase the ROM during pelvic retroversion (López-Miñarro et al., 2012; Konrad and Tilp, 2014). The antagonist-contract stretching (AC stretching) employed is known as an established method to increase the ROM and has the added advantage of easy and unsupported implementation (Konrad et al., 2017).
(3) Exercises for body perception (pelvis lift lying down, lordosis adjustment, pelvis retroversion, perception of pelvic movement) (Bruhn et al., 2004). These exercises especially trained the self-perception of body posture and active use of the muscle groups that straighten the pelvis.

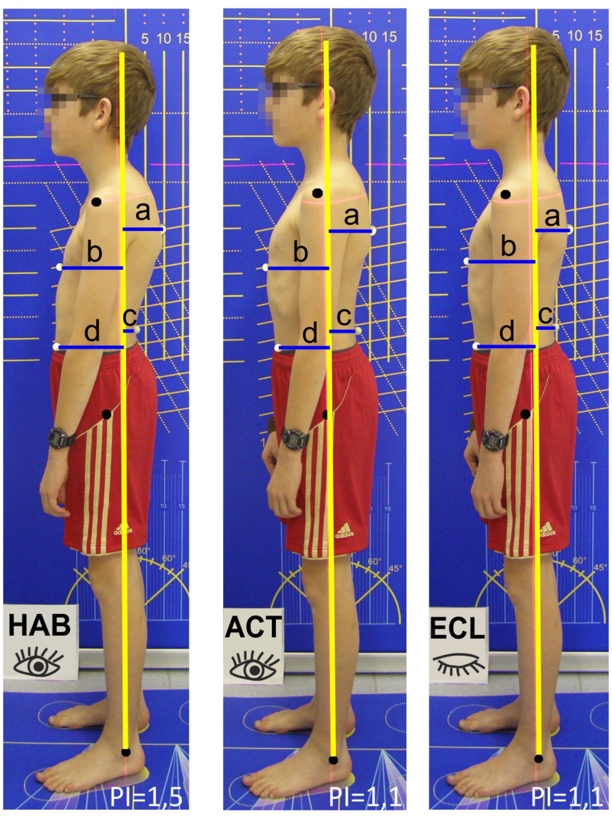

FIGURE 2 | Schematic posture analysis in the three posture situations habitual (HAB), active with open eyes (ACT), active with closed eyes (ECL). The posture indices HI are calculated as (a+d)/(b+c). Photomontage, courtesy of Elsevier publishing house.

Details of the athletic posture training were presented in an earlier study (Ludwig et al., 2016a). Every training session was protocolled in a person-specific, paper-based training protocol, in which the number of repetitions and the individual loads were noted. All training sessions were instructed by the same trainer (first author O.L.), who supervised correct movements and motivated the participants if necessary. Overall, the participants' training attendance was very good: as long as they participated in the study their average posture training workload was 1.81 h/week.

The posture measurements were repeated annually. After reaching adulthood, 32 adolescents left the posture training at their own request. 28 of these, however, still participated in the annual control examinations. 24 adolescents continued to exercise weekly until they reached age 20 (see flowchart in **Figure 1**). During the annual examinations, the weekly sedentary and standing behavior and athletic activity were recorded using a survey in form of a questionnaire (**Supplementary Data Sheet 1**). This served to identify and evaluate potential disturbance factors arising from the now non-school daily routines, which differed greatly among the test persons.

TABLE 2 | Exercises of the multi-dimensional posture training program.

	Training goals		Target muscle/*movement objective*	Equipment/Position	Movement
Strength	*Strengthening*	1	M. glutaeus max.	Glutaeus machine	Move a straight leg backward against resistance
		2	M. biceps fem.	Knee flexion, sitting	While seated with the knees at a right angle move the calves backward against resistance
		3	M. rectus abdom.	*Abdominal machine*, sitting	While seated bend the upper body forward against resistance
Stretching/Mobility	*Stretching/improving range of motion (ROM)*	4	M. iliopsoas, M. quadriceps fem.	Antagonist contract stretching in a lying position	Move the straight leg actively, then passively backward in the hip joint
		5	M. rectus femoris	Antagonist contract stretching in a lunge	Move the rear knee actively backward toward the lower back, then passively using a hand for support
Body perception	*Body perception*	6	*Control of pelvic position/Reducing lumbar lordosis*	Supine position	Actively neutralize lumbar lordosis under muscular tension
		7	*Control of pelvic position*	Supine position	Thigh vertical, knee bent at 90°, slightly (1 cm) lift pelvis from the floor
		8	*Control of pelvic position/Global posture*	Standing position	Tilt pelvis backward and forward, upper body and thighs remain motionless
		9	*Posture correction*	Standing position with mirror control	Targeted alignment of the body with the perpendicular

Statistics

Potential differences between the groups TR18 (training until the age of 18, then stop), TR20 (uninterrupted training until the age of 20), and CON (control group) before training start were examined by means of univariate variance analysis (ANOVA) for anthropometric parameters, athletic activity, sedentary parameters, and the posture indexes of the HAB, ACT, and ECL posture situations.

In order to identify potential confounding effects caused by anthropometric parameters or leisure behavior, we used a linear mixed model approach (mixed design ANOVA) with the posture indexes as dependent variables, and body weight, body height, hours of athletic activity per week, hours of sedentary activity, time, and group as model variables. We included all available data in this model. The homogeneity of the variances was verified using the Levene test, the heteroscedasticity was tested by the modified Breusch–Pagan-test.

Post hoc pairwise comparisons were performed according to Scheffé. The effect size was estimated based on Cohen's effect size measures based on partial eta square (η^2) and Cohen's *d* (Fröhlich and Pieter, 2009). A strong effect exists with $\eta^2 \geq 0.14$. The significance level was set at 0.05.

RESULTS

When the study started (age 14), there were no significant differences in the anthropometric data among the three groups (weight: $F = 0.27$, df = 2, $p = 0.77$, height: $F = 0.22$, df = 2, $p = 0.80$). The parameters describing leisure behavior, like sedentary behavior ($F = 0.74$, df = 2, $p = 0.48$) and athletic activities ($F = 0.01$, df = 2, $p = 0.99$) did not differ significantly, as well as the posture indices for habitual posture (HAB, $F = 0.84$, df = 2, $p = 0.44$), active posture (ACT, $F = 2.72$, df = 2, $p = 0.07$), and active posture with closed eyes (ECL, $F = 0.29$, df = 2, $p = 0.75$). Data are presented in **Tables 1**, **3** and the development over time is indicated in **Figure 3**.

After 2 years of training (age 16), we found significant improvements in the exercising groups TR18 and TR20 for all posture positions (posture index < 1.35, $F = 7.62$, df = 2, $p < 0.001$), both within the group and in comparison with the control group. From age 18 on, the posture parameters changed in different ways, which we will describe in the following paragraphs.

Habitual Posture

The linear mixed model showed a significant inter-individual effect for the *Group* factor for the habitual posture in the course of the study ($F = 52.107$, df = 2, $p < 0.0001$, $\eta^2 = 0.18$) and significant intra-subject effects for the *Time* factor ($F = 13.345$, df = 6, $p < 0.0001$, $\eta^2 = 0.14$) and the interaction *Group***Time* ($F = 7.059$, df = 12, $p < 0.0001$, $\eta^2 = 0.15$).

The other model variables did not show any significant effect (body weight: $F = 1.485$, df = 1, $p = 0.224$; height: $F = 0.008$, df = 1, $p = 0.927$; hours of athletic activity: $F = 3.532$, df = 1, $p = 0.061$; hours of sedentary activity: $F = 1.724$, df = 1, $p = 0.190$).

After the study was terminated, group comparisons of posture indexes between control group and TR 18 ($p = 0.06$) and between TR18 and TR20 ($p = 0.10$) showed no significant pair differences,

TABLE 3 | Posture parameter development in the three posture positions and test groups over time.

		TR20				TR18				CON			
		n	Mean	*SD*	C.I. (95%)	*n*	Mean	*SD*	C.I. (95%)	*n*	Mean	*SD*	C.I. (95%)
Habitual	14 years	26	1.42	0.06	1.39–1.44	31	1.42	0.05	1.40–1.44	23	1.41	0.06	1.36–1.43
	15 years	26	1.40	0.07	1.38–1.44	31	1.40	0.06	1.38–1.42	23	1.40	0.06	1.36–1.42
	16 years	26	1.30	0.08	1.27–1.34	31	1.33	0.07	1.31–1.36	22	1.40	0.07	1.36–1.43
	17 years	25	1.28	0.05	1.26–1.30	30	1.28	0.05	1.26–1.30	19	1.39	0.06	1.35–1.41
	18 years	24	1.28	0.05	1.26–1.30	28	1.27	0.09	1.24–1.31	16	1.37	0.05	1.34–1.40
	19 years	24	1.27	0.06	1.25–1.30	28	1.36	0.08	1.33–1.39	16	1.39	0.07	1.35–1.42
	20 years	24	1.27	0.05	1.25–1.30	28	1.37	0.08	1.34–1.40	15	1.40	0.06	1.37–1.43
Active, eyes open	14 years	26	1.35	0.06	1.33–1.38	31	1.37	0.05	1.35–1.39	23	1.32	0.05	1.30–1.36
	15 years	26	1.34	0.07	1.31–1.37	31	1.34	0.08	1.31–1.38	23	1.32	0.07	1.28–1.37
	16 years	26	1.20	0.08	1.16–1.23	31	1.23	0.06	1.20–1.25	22	1.34	0.05	1.33–1.38
	17 years	25	1.20	0.06	1.17–1.22	30	1.21	0.08	1.18–1.24	19	1.34	0.05	1.31–1.37
	18 years	24	1.18	0.09	1.14–1.21	28	1.19	0.10	1.15–1.23	16	1.34	0.05	1.31–1.37
	19 years	24	1.18	0.09	1.15–1.22	28	1.25	0.07	1.22–1.27	16	1.33	0.05	1.30–1.36
	20 years	24	1.19	0.07	1.16–1.22	28	1.24	0.11	1.20–1.28	15	1.33	0.05	1.30–1.36
Active, eyes closed	14 years	26	1.39	0.07	1.36–1.42	31	1.40	0.05	1.38–1.42	23	1.40	0.07	1.36–1.43
	15 years	26	1.36	0.06	1.33–1.38	31	1.38	0.07	1.35–1.41	23	1.41	0.06	1.38–1.44
	16 years	26	1.22	0.09	1.18–1.25	31	1.26	0.07	1.23–1.29	22	1.40	0.06	1.37–1.43
	17 years	25	1.23	0.06	1.20–1.26	30	1.26	0.06	1.23–1.28	19	1.39	0.04	1.37–1.41
	18 years	24	1.22	0.07	1.19–1.25	28	1.24	0.07	1.22–1.27	16	1.37	0.06	1.34–1.40
	19 years	24	1.25	0.08	1.21–1.28	28	1.33	0.07	1.30–1.35	16	1.37	0.05	1.34–1.40
	20 years	24	1.24	0.06	1.22–1.27	28	1.33	0.07	1.31–1.36	15	1.39	0.05	1.36–1.41

SD, standard deviation; C.I., 95% confidence interval.

while the control group and the TR20 group differed significantly ($p < 0.001$). During the joint training phase up to age 18, no significant differences between the groups TR18 and TR20 were identified. In the course of the study, the TR18 group came closer to the control group after having suspended their exercises, while the TR20 group continued to improve their habitual posture ($p < 0.001$, $d = 1.2$) (**Figure 3A**).

Active Posture

For active posture, significant inter-individual effects were identified for the *Group* factor ($F = 69.701$, df = 2, $p < 0.0001$, $\eta^2 = 0.225$), in addition to significant intra-individual effects for the *Time* factor ($F = 10.918$, df = 6, $p < 0.0001$, $\eta^2 = 0.120$) and the interaction *Group*Time* ($F = 9.171$, df = 12, $p < 0.0001$, $\eta^2 = 0.187$). *Post hoc* group comparisons after the end of the study showed significant differences between the two training groups and the control group ($p < 0.001$ each).

The linear mixed model did not show any significant effect for the other model variables (body weight: $F = 2.189$, df = 1, $p = 0.140$, height: $F = 0.259$, df = 1, $p = 0.611$; hours of athletic activity: $F = 0.017$, df = 1, $p = 0.897$; hours of sedentary activity: $F = 0.320$, df = 1, $p = 0.572$). During the intervention phase, the two training groups did not differ significantly (15 years: $p = 0.99$, 16 years: $p = 0.19$, 17 years: $p = 0.88$, 18 years: $p = 0.73$). After the training was suspended by TR18 at age 18, the groups TR18 and TR20 differed in the 1st year (19 years: $p = 0.01$), while no difference was found in the 2nd year (20 years: $p = 0.10$) (**Figure 3B**).

Active Posture With Closed Eyes

For active posture with closed eyes, significant inter-individual effects were found for the *Group* factor ($F = 111.517$, df = 2, $p < 0.0001$, $\eta^2 = 0.318$), in addition to significant intra-individual effects for the *Time* factor ($F = 22.427$, df = 6, $p < 0.0001$, $\eta^2 = 0.219$) and the interaction *Group*Time* ($F = 8.171$, df = 12, $p < 0.0001$, $\eta^2 = 0.170$). The two intervention groups differed from the control group ($p < 0.001$ each) in the active posture with closed eyes. From age 19, a significant difference was identified between TR18 and TR20 (19 years: $p < 0.001$, 20 years: $p < 0.0001$), while no difference was found between CON and TR18 (**Figure 3C**).

All other model variables did not show any significant effect (body weight: $F = 1.032$, df = 1, $p = 0.310$, height: $F = 0.307$, df = 1, $p = 0.580$; hours of athletic activity: $F = 0.293$, df = 1, $p = 0.589$; hours of sedentary activity: $F = 0.017$, df = 1, $p = 0.895$).

Sedentary Behavior and Athletic Activity

In the course of the study, the weekly time spent in a sitting position was comparable between the three groups (**Figure 4A**). At the end of the study, all three groups did not differ significantly in terms of sedentary behavior ($F = 0.46$, df = 2, $p = 0.63$) or standing behavior ($F = 0.11$, df = 2, $p = 0.90$) during school, work, or studies. Athletic activity resulted in significant differences between TR20 (3.59 ± 1.40 h/week) on the one hand, and TR18 (1.39 ± 1.19 h/week) and CON (0.53 ± 0.66 h/week) on the other at the end of the study. When athletic activity was adjusted so that only supplementary activity was analyzed – that is, any

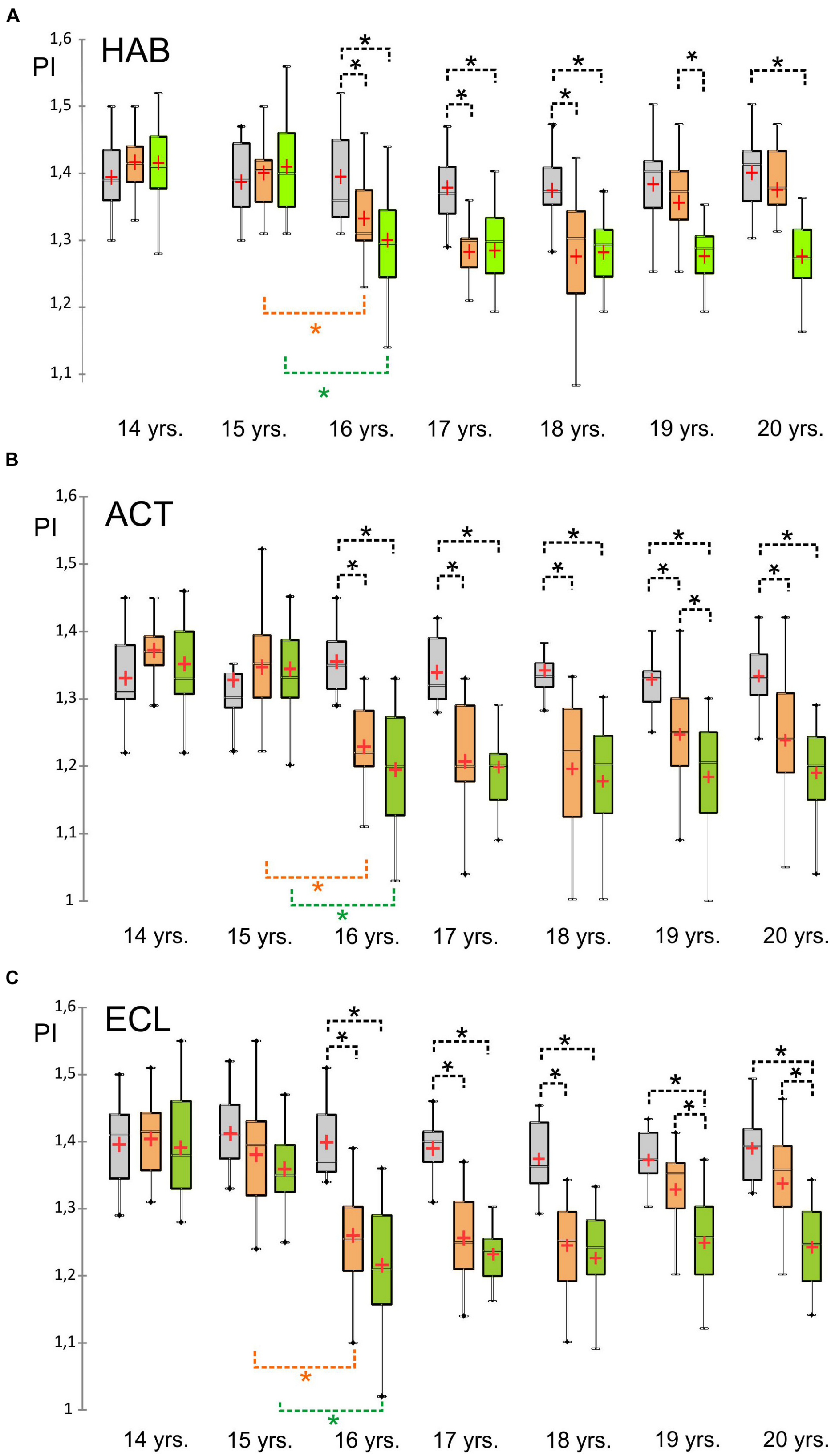

FIGURE 3 | Box plots of the posture index value development for the three groups over time. **(A)** (Above): habitual posture, **(B)** (middle): active posture with open eyes, **(C)** (below): active posture with closed eyes. *Y*-axes: posture index PI. Green: TR20, orange: TR18, gray: CON; * indicate significant differences.

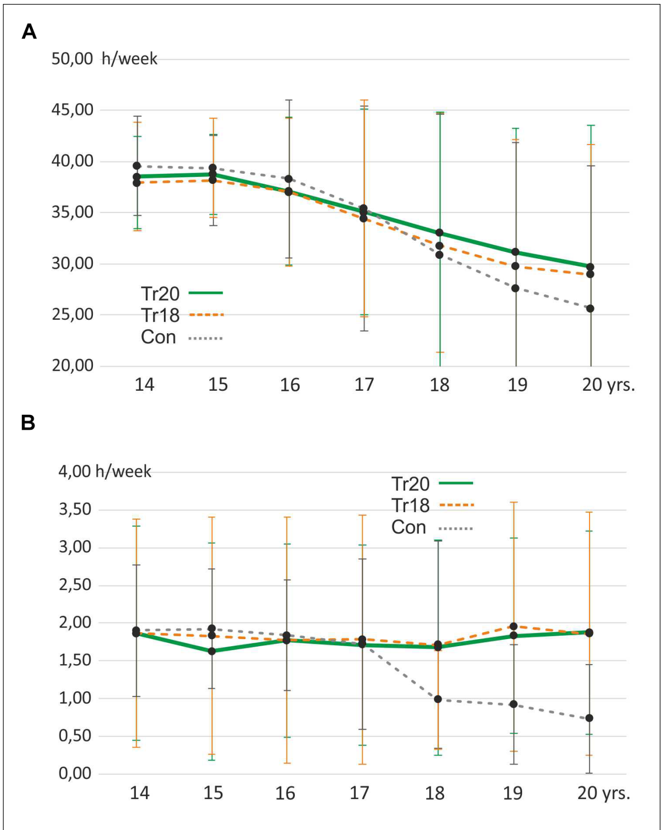

FIGURE 4 | Development of sedentary behavior **(A)** and supplemental athletic activities **(B)** in hours per week of the three groups over time (mean values, bars represent standard deviations).

activity in addition to the weekly posture training – there were no significant differences between the groups, although it was obvious that the participants of the control group spent less time with athletic activities from age 18 on (**Figure 4B**). Even though the differences did not reach significance level, their mean time spent with athletic activities was considerably lower.

DISCUSSION

The objective of this study was to analyze changes in the neuromuscular performance of posture regulation from adolescence to adulthood, and to determine the influence and potential effects of target-oriented athletic posture training.

Body posture is the result of a complex interaction of active structures (receptors and muscles) and passive elements (bones, tendons, fasciae, and ligaments) (Sousa et al., 2012). The quality of body posture control depends on the neuromuscular performance (Kilby et al., 2015).

Anthropometric Parameters and Leisure Behavior

Analyzing the changes of posture parameters during the long time span of 6 years may be susceptive to many confounders. Anthropometric parameters, such as body weight and height, as well as leisure behavior, such as athletic activities and sedentary behavior, change during the years of physical development and might constitute confounding variables. The latter are not easy to capture, much less to control. We did not find any significant influence of these factors, but we are aware that a survey once a year might not be optimal. Furthermore, more influencing variables like socioeconomic factors, developmental stage, environmental factors, and psychological variables may exist that we did not capture (Brodersen et al., 2005). Therefore, our results need to be interpreted with care.

Anthropometric variables seem not to have influenced posture development in a group-specific manner. The development of body weight and height was similar in all three groups and were within the normal national ranges (Neuhauser et al., 2013). Leisure behavior may also have had an important impact on the adolescents' physiological status as it is known that teenagers who prefer a more sedentary leisure behavior tend to have weaker muscles and a higher body mass index (Hanson and Chen, 2007). Once a year, we asked for leisure behavior in the form of a questionnaire. Up to the age of majority, this questionnaire was answered by children and parents together to avoid any type of bias that could have been produced by children who wanted to present themselves in a more positive light than was realistic. However, we had to rely on the truthfulness of the given statements. Furthermore, we must be aware that all statements had to be averaged over the period of 1 year. Changes in sedentary behavior during the year may, for example, have been caused by changing to another type of school or attending different classes with a higher or lower number of weekly lessons. For organizational reasons interviews in a shorter period were not feasible. Therefore, these statements need to be considered with caution.

Effects of Posture Training in Adolescence

Our first hypothesis that a supplemental targeted posture training program may improve selected posture parameters in adolescence seems to be confirmed. The exercising groups TR18 and TR20 showed significantly improved posture parameters after 2 years. The question arises whether the improvement was primarily due to the training program or confounded by other variables. We were able to exclude the influence of body mass and height by applying our mixed-model approach, as explained above. Other supplementary athletic activities did not significantly differ between the three groups until age 18, nor did sedentary behavior show a significant influence (compare **Figure 4**). We therefore suppose the additional posture training to be the main contributor to the group-specific posture improvement.

Studies that examined the effect of muscular training programs on posture in adolescents are sparse. Interrelationships between muscle strength and posture parameters were found by Barczyk-Pawelec et al. (2015), but they did not examine the direct influence of muscle strength on posture. Kim et al. (2006) found that an imbalance in trunk muscle strength could influence lumbar lordosis, which they assume to be a risk factor for low back pain. Pure maximum strength building of the core-stabilizing musculature does not necessarily lead to an improved posture. Studies by Klee showed an effectiveness of a combined strength and stretch program on the body posture of adolescents, but only few parameters and their changes were analyzed (Klee, 1994). Similar results were presented by Park et al. (2014) who could improve multiple posture deviations by a 6 months training program. Their program consisted of strengthening and stretching exercises, but they did not include body perception exercises. Other studies only focused on thoracic kyphosis and outlined its improvement by an athletic training (Betsch et al., 2015). Bansal et al. (2014) summarized in a review that exercises may result in a modest improvement of (thoracic) posture. However, their results were not homogeneous, as some studies could not identify any effect (Bansal et al., 2014). The multidimensional training program completed for this study focused on stretching, targeted strength training, and body perception. Special emphasis was placed on the area of senso-neuromuscular coordination, in which the perception of the own body position (Maravita et al., 2003) and, in particular, the conscious change of the pelvic position, was exercised. It is assumed that the joint positions subconsciously readjust via a neuromuscular balance, which leads to a change of habitual posture. Apart from that, a conscious posture correction is enabled when muscles are purposefully controlled (active posture). This, however, requires a good proprioceptive perception of joint positions, such as the pelvic position, and targeted and controlled muscle activation.

The participants of the control group exhibited a mean athletic activity of between 1.7 and 1.9 h/week during adolescence. This is comparable to the "basic" athletic activity (i.e., hours of independent weekly training besides the posture training) of both training groups. Nevertheless, we are not able to answer the question of whether the supplementary training (2 h per week) of the training groups *per se* was responsible for posture improvement, or whether the specific exercises were the reason. In other words, we cannot safely conclude that an unspecific training program of two supplementary hours per week would not have produced a comparable effect. Athletic activities of only 2 h per week, as found in the control group until the age of 17 (**Figure 4B**) obviously were not able to improve posture significantly. Other studies confirmed the benefit of a special posture training program, as well (Klee, 1994; Park et al., 2016). Our results are also in accordance with D'Amico and colleagues who found that self-correction maneuvers producing an improvement of body posture have to be learned with specific postural training (D'Amico et al., 2017). According to this, we assume that our multidimensional concept was the main reason why we were able to find such clear improvements. In an earlier study we confirmed the short-time effect of sensorimotor exercises as performed in the present study on posture improvement (Ludwig et al., 2016a). Other studies only examined short-term training programs. Our study was designed to find possible long-term effects, and we already identified significant effects after just 2 years. In general, positive effects of strength training will occur after three to 6 months (Klee, 1994; Ludwig et al., 2016a). In the present study, significant effects occurred later, but we must be aware that the 1st months of strength training were in part performed with very low weights. We initially trained 14-year-old children who had to learn the correct movements first. Simultaneously, they had to develop an awareness to be able to rate the strain on a modified Borg scale. This process developed slowly, and was even slowed down by the trainer sometimes in order to prevent any kind of physical overload. After 1 year, we were sure that all participants were able to perform their training in a targeted manner. Therefore, it is comprehensible why significant changes occurred only after 1 or 2 years.

Long-Term Effects in Adulthood

Our second hypothesis, i.e., that posture training regularly practiced in adolescence carries positive effects on body posture into adulthood, could be confirmed only in parts. In the continually exercising group (TR20), the habitual posture constantly remained in a good range (HI = 1.27 $\pm$ 0.06). **Figure 5** shows an example. In the group that stopped exercising from age 18 (TR18), the habitual posture index increased again. At the age of 20, it even deteriorated back to the initial value measured at the start of the study.

We interpret these findings that, if practiced continually, targeted posture training can sustainably improve the subconscious body posture. Since the habitual posture is maintained by means of subconsciously controlled neuro-motor processes (Sousa et al., 2012) and therefore decides on the strain placed on the musculoskeletal system in many everyday situations, it is especially important in terms of health aspects. More recent studies that identified an interrelation between posture deviations and the occurrence of back complaints in children and adolescents support this proposition (Dolphens et al., 2012, 2015, 2016). At a value of up to 65%, the prevalence of back pain in adolescence is noteworthy. To maintain a stable habitual posture, permanent training seems to be required.

In the continually exercising group (TR20), the active posture remained in a constant, good condition from age 16. The group that terminated their training from age 18 (TR18) also exhibited a good active posture, which differed significantly from the control group, but not from the TR20 group. We therefore conclude that the ability to control posture in a target-oriented manner seems to be preserved even after training breaks. Since in this posture position, a targeted deliberate muscular activation occurs, the development of a posture regulation skill can be viewed as the result of a learning process, during which not only strength endurance and flexibility were improved, but also the perception of one's own body and the targeted muscular activation (sensorimotor control) (Maurer et al., 2006; D'Amico

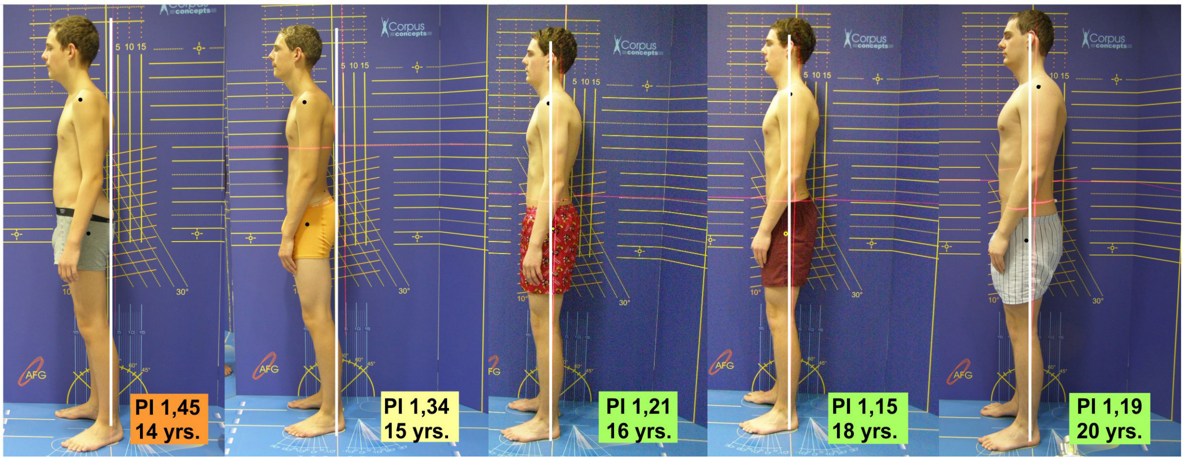

FIGURE 5 | Example of posture development in a continuously exercising test person over a period of 6 years. Posture index (PI) values < 1.30 indicate a stable body posture. Courtesy of Elsevier publishing house.

et al., 2017). Once learned, the motor programs required for this (i.e., targeted muscle activation in terms of time and amplitude) can be preserved over many years. Similar examples are known in swimming, biking, and skiing (Furley and Memmert, 2010). Like so many learned skills, their movement programs are presumably saved in the so-called 'procedural memory' (Squire, 2004). Once learned, these motor skills can be called upon after years without regular training being required. Therefore, we conclude that posture training performed in adolescence sustainably improves the ability of the test persons to correct their posture in a target-oriented and conscious manner in adulthood. We interpret this as positive (learning) effects that are carried over into young adulthood. However, the effects described are reproducible only when including the visual sense.

Comparing the active posture with closed eyes with the active posture with open eyes supplies an additional statement on the sensory information processing of the visual sense. When standing with closed eyes, body posture is controlled exclusively by proprioceptive sensory perception (Mallau et al., 2010; Chiba et al., 2016). Since the quality of posture regulation highly depends on targeted muscle control based on proprioception of the body, it needs to be trained.

The continually exercising group TR20 was able to maintain stable good posture values even without visual sensory information (i.e., their eyes were closed). Meanwhile, after the training stop of the TR18 group, a significant deterioration of their active posture occurred as soon as they closed their eyes. If posture deteriorates without the visual sense, the CNS is obviously not able to compensate for this loss in terms of posture control, e.g., by using only proprioceptive signals. Therefore, the changes of posture when closing the eyes give us a clue about the extent to which the CNS relies on the visual sense. It is known that proprioceptive information processing can be improved with exercises that are based on a targeted movement of certain body parts (e.g., the pelvic position) without vision (Bruhn et al., 2004; Zech et al., 2010). Such exercises were part of the weekly training program (see **Table 2**). Therefore, we interpret the fact of posture deterioration with closed eyes as suboptimal information processing of proprioceptive signals (Maurer et al., 2006; Mallau et al., 2010). That is, the CNS is not able to compensate the "switched off" visual sense by means of other sensory signals. According to our results, one may assume that this part of neuromuscular control obviously requires constant consolidation and is subject to deterioration if no adequate athletic activity takes place. This would explain why posture with closed eyes deteriorates in TR18 from the time when posture training was stopped. This finding corresponds to fundamental studies that had analyzed the adaptability of the neuromuscular control system (Asslander and Peterka, 2014). The CNS obviously changes the weighting of various sensor systems depending on the requirements in daily routine (Chiba et al., 2016). This could explain why a deterioration of posture regulation was observed without the visual sense when sensorimotor training was suspended. Nevertheless, we must be careful in interpreting these results as it is known that leisure behavior with strong visual components (e.g., playing computer games) may also influence visual signal processing in the brain (Oei and Patterson, 2013) and might also have confounded our results.

For practice, we can conclude that the proprioceptive perception of the body's position without the use of the visual sense should be trained regularly.

Interrelationships between poor posture and back problems are known. From a medical-preventive point of view, targeted posture training should therefore start in early adolescence and be continued throughout life. Adequate training elements can be found in many types of sport, particularly in martial arts, gymnastics, and gymnastic disciplines in general (Perrin et al., 2002; Tsaklis et al., 2008; Zech et al., 2010; Iunes et al., 2016). In principle, every athletic training program can be supplemented by the corresponding elements. The training program described in this study is set in a 'fitness sport' environment and enjoys the advantage of a high degree of acceptance, especially in the group of adolescents. Adolescence is the time when classical sports clubs usually lose their appeal and gym training becomes more attractive (Aarnio et al., 2002; Tammelin et al., 2003). Qualified exercise instructions are, of course, an important precondition.

LIMITATIONS

According to our knowledge, this study is the first that analyzed posture development and its trainability over a period of many years from adolescence to adulthood. However, it does have its limitations. For example, only symptom-free male test persons were examined. Statements on intervention options for back pain patients can therefore not be given. Nevertheless, from a preventive point of view, the target group with an increased pelvic tilt seems to be important, as students with posture weakness seem to have a greater prevalence of back pain in later adulthood (Dolphens et al., 2015).

Group Allocation and Dropouts

As mentioned above, group allocation was not random. Though, we asked for the reasons why potential participants did not want to join the training group. Most of them did not refuse strength training on principle, but stated time and logistics reasons. We therefore assume that the control group did not consist of some sort of a "negative selection," at least at the beginning of the study. At that time, their athletic activity and their sedentary time was comparable to both training groups (see **Figure 4**). However, at the end of the study, at age 20, they spent more time in a sedentary position than the training groups. Athletic activity of the control group was also going down at the end of the study. The fact that it did not reach significance level was possibly due to the low sample size and the high standard deviation.

Dropouts during such a long study period can hardly be avoided. We are aware that dropouts may produce a bias in the sense that unmotivated test persons leave the study at an early stage, leaving motivated test persons in the study and therefore generating more positive results. Although we could not totally avoid such a bias, we tried to keep its influence low. We included all test persons in our linear mixed model until the time of their dropping out. Furthermore, we asked in detail for the reasons of leaving the study. Most of them primarily stated logistics reasons or changes in their life situations (e.g., changing the place of residence, logistics problems/residence too far away from the fitness center, or too little time caused by their school situation or an apprenticeship). Only four participants left the study at age 18 because they had reached their individual training goals. All other participants of the TR18 group stated logistics reasons, as explained above. Therefore, we would not suggest a lack of motivation in any of these cases and view the resulting *bias* as acceptable.

Posture Measurement

In general, habitual posture shows a daily fluctuation depending, for example, on a subject's awareness and/or exhaustion. Earlier studies showed that the reproducibility of posture index measurements was good (Cronbach's alpha = 0.842, Ludwig et al., 2016b). In addition, we tried to improve the internal validity of posture measurements by means of a standardized measuring protocol. Furthermore, all measurements were performed in the morning when the participants were not physically or mentally fatigued. Nevertheless, we cannot fully exclude posture fluctuations as a potential source of bias.

Confounders

Even though potential disturbance variables, such as individual standing and sedentary behavior and supplementary athletic activities were surveyed using a questionnaire, they cannot entirely reflect differences in individual lifestyles. We did not find sedentary behavior to be a significant influencing factor during the course of the study. Nevertheless, we are aware that leisure behavior, that cannot fully be captured, might have had an influence on our output variables. Despite the control of potential disturbance variables, more complex factors that influence body posture, such as sporadic athletic activity or job-related factors, coming into play in adulthood, cannot be captured completely. We tried to evaluate possible influencing factors using a linear mixed model. Nevertheless, we are aware that there may be further factors that we are not aware of.

Since all test persons were supervised by one and the same tester, information regarding changes in leisure activity or additional athletic activity could be obtained between the yearly questionnaire surveys. For example, there was no test person who performed any additional athletic activities with strong balance components or sensorimotor training stimulus (e.g., gymnastics, skating, or martial arts), which could notably have improved their posture control. The "basic" athletic activity of all three groups was comparable, with most of the participants playing soccer, handball or performing cycling. In our yearly questionnaire we asked for the percentage of sensorimotor training or athletic training that in some sport clubs complements the practical training. Nevertheless, this factor was not susceptible of qualitative analysis, because we did not know the content and quality of these training sessions.

CONCLUSION

Supplemental targeted athletic training of 2 h/week during adolescence may improve the active straightening of the body posture in a pain free population. This ability transitions into adulthood, even if training is no longer regular. Subconscious body posture (habitual posture) and posture with closed eyes, both regulated particularly by proprioceptive receptor systems, can also be improved by supplemental adequate training, but do require continuous exercising if the improvement is to be maintained.

ETHICS STATEMENT

This study was carried out in accordance with the recommendations of the Saarland University Germany with written informed consent from all subjects and their parents, in accordance with the Declaration of Helsinki. The protocol was approved by the Ethic committee of Faculty 5 - Empirical Human Sciences.

AUTHOR CONTRIBUTIONS

OL, AH, and ES participated in the design of the study, carried out the experiments, and performed data analyses. OL wrote the manuscript. MF and AH helped to write the manuscript and supervised statistical analysis. JK contributed to the study design and supervised the entire project. All authors read and approved the final manuscript.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2018.01620/full#supplementary-material

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Effects of an Eight-Week Superimposed Submaximal Dynamic Whole-Body Electromyostimulation Training on Strength and Power Parameters of the Leg Muscles: A Randomized Controlled Intervention Study

*Florian Micke, Heinz Kleinöder, Ulrike Dörmann, Nicolas Wirtz and Lars Donath**

Department of Intervention Research in Exercise Training, German Sport University Cologne, Cologne, Germany

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***Correspondence:**
*Lars Donath
l.donath@dshs-koeln.de*

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The purpose of this study was to assess the effects of dynamic superimposed submaximal whole-body electromyostimulation (WB-EMS) training on maximal strength and power parameters of the leg muscles compared with a similar dynamic training without WB-EMS. Eighteen male sport students were randomly assigned either to a WB-EMS intervention (INT; $n = 9$; age: 28.8 (*SD*: 3.0) years; body mass: 80.2 (6.6) kg; strength training experience: 4.6 (2.8) years) or a traditional strength training group (CON; $n = 9$; age: 22.8 (2.5) years; body mass: 77.6 (9.0) kg; strength training experience: 4.5 (2.9) years). Both training intervention programs were performed twice a week over a period of 8 weeks with the only difference that INT performed all dynamic exercises (e.g., split squats, glute-ham raises, jumps, and tappings) with superimposed WB-EMS. WB-EMS intensity was adjusted to 70% of the individual maximal tolerable pain to ensure dynamic movement. Before (PRE), after (POST) and 2 weeks after the intervention (FU), performance indices were assessed by maximal strength (F_{max}) and maximal power (P_{max}) testing on the leg extension (LE), leg curl (LC), and leg press (LP) machine as primary endpoints. Additionally, vertical and horizontal jumps and 30 m sprint tests were conducted as secondary endpoints at PRE, POST and FU testing. Significant time effects were observed for strength and power parameters on LE and LC (LE F_{max} +5.0%; LC P_{max} +13.5%). A significant time × group interaction effect was merely observed for F_{max} on the LE where follow-up *post hoc* testing showed significantly higher improvements in the INT group from PRE to POST and PRE to FU (INT: +7.7%, $p < 0.01$; CON: +2.1%). These findings indicate that the combination of dynamic exercises and superimposed submaximal WB-EMS seems to be effective in order to improve leg strength and power. However, in young healthy adults the effects of superimposed WB-EMS were similar to the effects of dynamic resistance training without EMS, with the only exception of a significantly greater increase in leg extension F_{max} in the WB-EMS group.

Keywords: whole-body EMS, electrical stimulation, strength training, MVC, peak power output

INTRODUCTION

Health-related strength training recommendations regarding intensity, frequency and volume of strength training for maximal strength gains in trained individuals refer to 80–85% of One-repetition maximum, 2 days per week with a volume of 3–8 sets per muscle group (Peterson et al., 2005). If strength training is incorporated into fitness programs, it can also improve cardiovascular functions and psychological well-being, prevent osteoporosis and promote both weight loss and maintenance (Ratamess et al., 2009). Appropriate maximal strength and power training is also considered crucial for sport-specific physical development in terms of speed, dynamics and injury prevention (Reilly, 2007; Sander et al., 2013).

Electromyostimulation (EMS), an training technology for intensifying the training load, is known to be an effective and appealing complementary add-on training method to potentially further improve athletic performance factors (Filipovic et al., 2012). The benefits of EMS training can be attributed to the accentuated activation of fast motor units at relatively low force levels due to a non-selective recruitment pattern (Gregory and Bickel, 2005). Furthermore, EMS potentially supports the athlete in achieving greater strength and power adaptations by a synchronous recruitment of muscle fibers and an increased firing rate (Gregory and Bickel, 2005).

Most of the previous studies in the field of EMS examined isometric contractions of local muscles with maximal stimulation intensities at the individual pain threshold (Filipovic et al., 2011). Intervention studies using isometric EMS revealed considerable gains in isometric strength of about +32% in trained athletes (Filipovic et al., 2012). However, not only the increase in strength and power is essential. The functional transfer of these gains into sport-specific movements, especially in competitive sports is even more relevant. A sport-specific orientation of EMS training with dynamic exercises could account for the requirements of athletes, in order to achieve this functional transfer (Cormie et al., 2011a,b). Isometric EMS with maximal stimulation intensities does not meet the movement specificity required for the completion of sports movements (Paillard et al., 2005). When superimposed EMS is applied onto voluntary contractions, force was higher with respect to voluntary actions at eccentric actions (Paillard et al., 2005). In concentric and isometric actions, voluntary activation evoked higher force than EMS (Paillard et al., 2005). However, most recent evidence suggests that EMS superimposed onto voluntary contractions in a submaximal task could result in greater muscle fibers recruitment compared with voluntary or electrical stimulation alone and would be likely to generate greater gains of motor output after a training period (Paillard, 2018). Moreover, a low voluntary movement control exists at maximum stimulation intensities (Babault et al., 2007) and only submaximal contractions enable an efficient movement control with superimposed EMS (Bezerra et al., 2011). As a consequence, only stimulation intensities below the individual pain threshold allow dynamic movements with superimposed EMS. Taking the transfer of strength and power to sport-specific movements into account, it is well documented that a combination of separate EMS training and separate dynamic sport-specific exercises like jumping and sprinting could lead to these transfer effects (Maffiuletti et al., 2002; Herrero et al., 2006).

Compared to a local EMS, Whole-Body-Electromyostimulation (WB-EMS) stimulates several muscle groups like muscle-chains or agonist/antagonist simultaneously during dynamic movements. In non-athletic adults it is known that WB-EMS improves muscle mass and function while reducing fat mass and low back pain (Kemmler et al., 2018). At least in athletes, there is some evidence that locally applied EMS was slightly more favorable for increasing strength-related outcomes compared with WB-EMS (Kemmler et al., 2018). However, stimulation of muscle chains could support dynamic movements by compensating usual weak points like hip extensor (Lynn and Noffal, 2012) or lower back muscles (Hamlyn et al., 2007). It is assumed that a simultaneously and counterproductive activation of agonist and antagonist evokes additional demands on voluntary contraction, especially on a reduced co-activation of antagonistic muscles to continue dynamic exercises with superimposed EMS (Wirtz et al., 2016).

Due to the aforementioned background, the question arises whether a simultaneous combination of submaximal WB-EMS and dynamic strength and/or sport-specific exercises leads to improvements in both strength and power as well as jumping and sprinting performance. On the one hand, an advantage of this training approach is indicated by strength training over the entire muscle length. On the other hand it might induce beneficial effects due to the intensification of the technique training. Therefore, the aim of this study was to evaluate the effects of an 8-week, 16-session training program using dynamic submaximal WB-EMS training compared with traditional voluntary dynamic strength training without WB-EMS on (1) maximal strength and maximal power parameters and (2) on jumping and sprinting performance in male adult sport students. We hypothesized that the use of dynamic submaximal superimposed WB-EMS provides greater training adaptations and improves performance to a greater extent compared to traditional dynamic strength training without WB-EMS alone.

MATERIALS AND METHODS

Study Design

This study was designed as a 2-armed parallel-group, randomized controlled trial comparing the effects of submaximal superimposed dynamic WB-EMS (INT) with the effects of dynamic strength training without WB-EMS (CON) (**Figure 1**). The INT and CON groups completed 16 training sessions in 8 weeks twice a week. To determine training effects, isometric strength and isoinertial power diagnostics as well as jumping and sprinting tests were conducted under constant and stable lab conditions. Measurements of the primary and secondary outcome took place before the training period (PRE), after the training period (POST) and 2 weeks after the training period as follow-up (FU). PRE-, POST-, and FU-testings were intra-individually performed at the similar time of the day. After PRE, the subjects were randomly assigned (minimization method, strata: age, strength training experience) to either INT or CON.

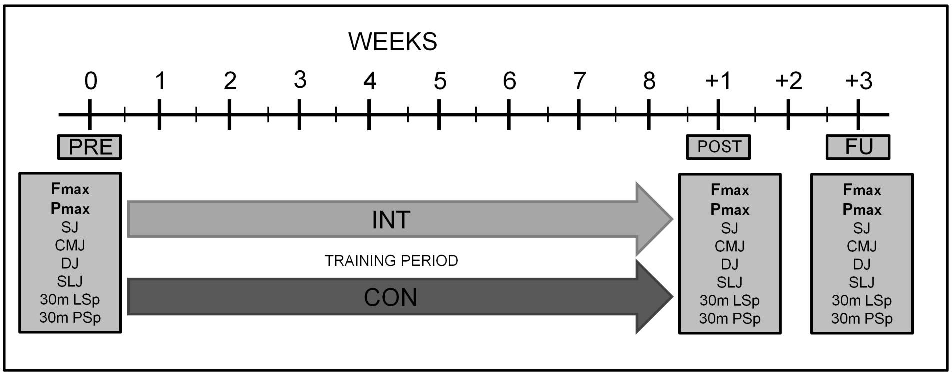

FIGURE 1 | Schematic representation of the experimental protocol. F_{max}, maximal isometric force; P_{max}, maximal isoinertial power; SJ, Squat Jump; CMJ, Counter Movement Jump; DJ, Drop Jump; SLJ, Standing Long Jump; 30 m LSp, 30 m Linear Sprint; 30 m PSp, 30 m Pendulum Sprint; INT, WB-EMS Intervention Group; CON, Traditional strength training Group.

In order to minimize influences of unspecific training loads, both groups were asked to refrain from any changes of their habitual physical activity behavior. Furthermore, all participants were instructed to maintain their normal dietary intake before and during the study.

Participants

Twenty male sport students volunteered to participate in the study. Inclusion criteria were: young adults between 18 and 30 years, who had a medical certificate attesting full physical fitness, had at least 2 years of strength training experience and had a sporting background in sports which requires performance abilities such as sprinting and/or jumping (e.g., soccer, handball, basketball, football, track and field, tennis). Exclusion criteria were any training experiences in WB-EMS. One week before the PRE tests, a familiarization session for testing was conducted. Thereby, the testing devices were adjusted and the participants were familiarized with the testing procedures. After the randomization, participants of the INT group were familiarized with WB-EMS and the training intensity was determined. Written informed consent was obtained from all participants after giving comprehensive study instructions. The study protocol was approved by the "Ethics Committee of the German Sport University Cologne" and complied with the Declaration of Helsinki. Two participants, one in each group, had to terminate study participation due to injuries that were not related to the study. Finally, 18 participants completed all tests and the attendance rate for the training sessions was 100% for both groups (participants characteristics are presented in **Table 1**).

Training Procedure

During the 8-week training period, participants of both groups similarly performed 16 training sessions (TS) twice a week with the only difference that the INT group performed all exercises with additional superimposed WB-EMS. Each session consisted of 5 training exercises in total, 2 strength exercises for either leg extension or knee flexion (e.g., split squats, glute-ham raises) plus 3 dynamic (a) jumping exercises (e.g., hurdle jumps, lateral jumps; in the first TS of the week) or (b) sprinting exercises (e.g., resistance band running, ABC-running drills; in the second TS of the week). Training variables like intensity, number of repetitions, repetition velocity or exercises changed after the first 3 weeks and after the second 3 weeks, due to the progression principles of strength training (see **Supplementary Tables S1–S6**). Three sets of each training exercise were conducted. The number of repetitions differed between the exercises and ranged from 5 to 10 repetitions. Movement velocity and range of motion (ROM) were predetermined for every single training exercise and were controlled with a metronome and with markings, respectively. The training intensity of each set was recorded using the Borg Rating of Perceived Exertion (RPE) and set to >16 (> "hard") (Tiggemann et al., 2010). A rest interval of at least 48 h between each training session was complied.

WB-EMS

The INT group performed all exercises with additional superimposed WB-EMS. Surface electrodes (miha bodytec, Augsburg, Germany) were applied to the leg and trunk muscles. An electrode vest with fixed surface electrodes provided the stimulation of the upper body including the chest (electrode size: 15 cm length × 4.5 cm width), the upper back (23 × 10 cm), the lower back (14 × 11 cm), the latissimus (14 × 9 cm) and the abdominals (23 × 10 cm). A belt system provided the stimulation for the lower body including the muscles of the glutes (13 × 10 cm), the thighs (44 × 4 cm), and the calves (27 × 4 cm). The sizes of the electrode vest and the belt electrodes (small/medium/large) were selected according to the body size of each participant.

The WB-EMS intervention was complied with the guidelines for a safe and effective WB-EMS training (Kemmler et al., 2016a). The intensity of EMS during the training was adjusted to 70% of the individual maximal tolerable pain [maximum tolerated amperage (0–120 mA)] as previously described in detail elsewhere (Wirtz et al., 2016). The maximal tolerated amperage was determined separately for each pair of electrodes

TABLE 1 | Demographic variables mean (SD).

	N	Age (years)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Strength training exp. (years)
INT	9	22.8 (3.0)	179.4 (5.1)	80.2 (6.6)	22.3 (1.8)	4.6 (2.8)
CON	9	22.8 (2.5)	184.9 (9.1)	77.6 (9.1)	20.9 (1.5)	4.5 (2.9)

before every training session. Firstly, maximum intensity was verified for simultaneous stimulation of all muscle groups. Then, the intensity was subsequently downregulated with the main controller at the WB-EMS device to an intensity of 70% to enable dynamic movements. Impulse frequency was set at 85 Hz, pulse duration at 350 μs, impulse type was bipolar and rectangle (Kemmler et al., 2014; Filipovic et al., 2016; Kemmler et al., 2016b; Wirtz et al., 2016). On/off-time was individually adjusted within each exercise (see **Supplementary Tables S1–S6**). In general, EMS was applied during all the execution time of each exercise and stopped during the rest period.

Testing Procedures

Strength and Power Testing

Maximal isometric strength (F_{max}) and maximal isoinertial power (P_{max}) diagnostics for the leg muscles were conducted on the leg extension (LE), leg curl (LC), and leg press (LP) machines (Edition-Line, gym80, Gelsenkirchen, Germany). All machines were equipped with the digital measurement technique Digimax (mechaTronic, Hamm, Germany). The multi-channel measuring system consisted of a force and distance sensor (megaTron, Munich, Germany), a PC-2-Channel-Interface, a computer with serial port and measurement/analysis software (IsoTest 2.0 and DynamicTest 2.0). F_{max} [N] and P_{max} [W] were calculated for statistical analysis and data presentation. Each participant performed 3 isometric and 3 isoinertial test attempts on the each leg machine. For F_{max} and P_{max}, the attempt with the highest value was subsequently used for further analysis. Isometric tests were conducted at an inner knee angle of 120 degree on the LE as well as LP and 150 degrees on the LC. Isoinertial test attempts were conducted with an additional load. This load was individually calculated as a percentage of the F_{max} in a further isometric test with the same angle as the starting position of the isoinertial test (LE and LP 90°; LC 170°). The attempts were conducted with 40% additional load on the LC and with 60% additional load on the LP as well as on the LE over a concentric ROM (inner knee ROM: LE and LP 90–180°; LP 170–80°). The instruction was to press against the lever arm "as hard and fast as possible" (Maffiuletti et al., 2016). Strength and power parameters were considered as primary endpoints.

Jumping Tests

Jumping performance was quantified using the Optojump system (Microgate, Bolzano, Italy). Therefore, jump height was assessed using the flight time method. After one familiarization jump trial, the participants performed 3 trials of each jump variation [1. squat jump (SJ), 2. counter movement jump (CMJ), 3. drop jump (DJ), and 4. Standing long jump (SLJ)] in a fixed order. For the respective jump task, the highest or longest jump was used for subsequent analysis. For the SJ, participants were instructed to start jumping from a static squatted position holding the knees at 90 degrees without any preliminary movement. For the CMJ, participants were instructed to start the jump from an upright standing position, squatting down to a knee angle of approximately 90 degrees in order to jump explosively as high as possible. DJs were performed from a 38 cm box (Bobbert et al., 1987). Participants were instructed to step down from the box and then to try to jump as high as possible after a short contact time on the ground. Hands remained akimbo for the entire movement of each vertical jump in order to eliminate any influence of the arm swing. The DJ height (DJH) and the DJ contact time (DJCT) were measured. For the SLJ, the participants started the horizontal jump from an upright standing position. They were instructed to gain adequate momentum by squatting down in order to jump as far as possible and to complete the jump with a controlled landing. Jump length was determined by measuring from tip to the participants' rear-most heel.

Sprinting Tests

Sprinting tests were conducted in an indoor hockey hall with a non-slippery floor. Performance was tested with a linear 30 m sprint (30 mLSp) and a pendulum sprint of 3 × 10 m (30mPSp) with 2 changes of direction of 180° (at 10 and 20 m) (Filipovic et al., 2016; Wirtz et al., 2016). Final sprint time was measured at 30 m. Starting position was 50 cm in front of the starting light beam for both sprint tests. Participants had 2 min recovery between the trials. Double infrared photoelectric barriers with a radio transmitter (DLS/F03, Sportronic, Leutenbach-Nellmersbach, Germany) were used for time measurement. The fastest time of 3 trials per sprint variation was used for subsequent analysis. Jumping and sprinting tests were conducted as secondary endpoints.

Statistical Analysis

Data were given as means with standard deviations (SD). Statistical analyses were performed by using a statistics software package (IBM SPSS Statistics, Version 25.0, Armonk, NY, United States). All parameters were normally distributed (Shapiro–Wilk test) and variances were homogeneous (Levene test). Then, separate 2 (group: INT, CON) × 3 (time: PRE, POST, and FU) repeated measures analyses of variances (rANOVA) were calculated. Thereby, PRE values of the respective primary or secondary outcome was included as covariate in order to adjust for possible baseline differences (Vickers and Altman, 2001). In case of a significant time × group interaction, Bonferroni *post hoc* tests and standardized mean differences (SMD) were calculated for pairwise comparison. The magnitude of SMD was classified according to the following scale: 0–0.19, negligible effect; 0.20–0.49, small effect; 0.50–0.79, moderate effect; and ≥0.80, large effect (Cohen, 1988). To estimate overall time and interaction effect sizes, partial eta squared (η_p^2) was computed with $\eta_p^2 \geq 0.01$ indicating small, ≥0.059 medium and ≥0.138 large effects (Cohen, 1988). The level of significance was set at $p < 0.05$.

RESULTS

Strength and Power Parameter

F_{max} and P_{max} values for both groups are provided in **Table 2**. All data are adjusted for baseline differences. A significant and large time × group interaction was merely observed for F_{max} on the LE ($p = 0.029$; $\eta_p^2 = 0.21$) where *post hoc* comparisons indicated higher improvements in the INT group from PRE to POST (INT: +6.9%, $p < 0.01$; CON: +0.5%) and PRE to FU (INT: +7.7%, $p < 0.01$; CON: +2.1%). Significant time-effects were observed for LE F_{max} ($p = 0.016$; $\eta_p^2 = 0.24$) and LC P_{max} ($p = 0.002$; $\eta_p^2 = 0.35$) (see **Supplementary Table S7**).

Jump and Sprint Parameter

Jump and sprint values for both groups are provided in **Table 3**. All data are adjusted for baseline differences. A significant and large time × group interaction ($p = 0.007$; $\eta_p^2 = 0.32$) was only observed for 30 mLSp where *post hoc* comparisons indicated a significant decline from PRE to POST and a significant improvement from POST to FU only in the INT group. Significant time-effects were observed during CMJ ($p = 0.038$; $\eta_p^2 = 0.20$), DJH ($p = 0.003$; $\eta_p^2 = 0.33$) and 30 mPSp ($p = 0.049$; $\eta_p^2 = 0.21$) (see **Supplementary Table S8**).

DISCUSSION

This study investigated the effects of an 8-week dynamic submaximal superimposed WB-EMS training on strength and power parameters of the leg muscles as well as on jumping and sprinting performance. It was hypothesized that WB-EMS training would provide greater training adaptations than a similar strength training without superimposed WB-EMS. The main findings of this study indicate that (a) superimposed dynamic WB-EMS does not provide greater benefits than dynamic resistance training alone and that (b) transfer effects on jumping and sprinting performance seem to be restricted in both groups.

Previous EMS studies solely compared WB-EMS training to traditional strength training with non-comparable standardizations. Our study used a standardization procedure with similar parameters for both groups regarding exercises, number of repetitions, number of sets, ROM, movement velocity and RPE. The present intervention program further revealed that WB-EMS training over 8-weeks twice a week led to improvements ranging from 8 to 15% in F_{max} and from 5 to 16% in P_{max} on average. The dynamic training program without superimposed WB-EMS led to improvements from 2 to 21% (F_{max}) and 8 to 11% (P_{max}) on average. The EMS-related

TABLE 2 | Maximal Strength (F_{max}) and Power (P_{max}) data for Leg Extension (LE), Leg Curl (LC), and Leg Press (LP) for both groups during PRE, POST, and FU testing, including change (delta).

	Parameter	Group	PRE	POST	% Delta PRE-POST	SMD PRE-POST	FU	% Delta PRE-FU	SMD PRE-FU	ANOVA p (η_p^2) Time Effect	ANOVA p (η_p^2) Time x Group Interaction
LE	F_{max} (N)	INT	2602 (367)	2781 (361)	+6.9**	0.49	2803 (291)	+7.7***	**0.61**	**0.016 (0.239)**	**0.029 (0.211)**
		CON	2502 (340)	2514 (229)	+0.5	0.04	2554 (280)	+2.1	0.17		
	P_{max} (W)	INT	1141 (236)	1199 (184)	+5.1	0.28	1225 (223)	+7.3	0.37	0.093 (0.146)	0.947 (0.004)
		CON	1036 (169)	1101 (190)	+6.3	0.36	1123 (189)	+8.4	0.49		
LC	F_{max} (N)	INT	1334 (192)	1422 (198)	+6.6	0.45	1455 (233)	+9.1	**0.57**	0.833 (0.012)	0.054 (0.177)
		CON	1348 (176)	1339 (219)	−0.7	−0.05	1364 (164)	+1.2	0.09		
	P_{max} (W)	INT	718 (205)	774 (130)	+7.8	0.32	832 (190)	+15.9	**0.58**	**0.002 (0.350)**	0.371 (0.064)
		CON	647 (97)	712 (114)	+10.1	**0.61**	719 (109)	+11.0	**0.69**		
LP	F_{max} (N)	INT	3255 (555)	3526 (769)	+8.3	0.40	3744 (829)	+15.0	**0.69**	0.513 (0.043)	0.690 (0.024)
		CON	2905 (566)	3339 (671)	+14.9	**0.70**	3517 (752)	+21.1	**0.92**		
	P_{max} (W)	INT	1674 (285)	1710 (324)	+2.2	0.12	1774 (279)	+6.0	0.35	0.229 (0.094)	0.326 (0.072)
		CON	1361 (251)	1498 (289)	+10.1	**0.51**	1497 (260)	+10.0	**0.53**		

*Values are presented as mean (±SD). Moderate to large standardized mean differences (SMD) have been highlighted in bold. Significance level was set at **p < 0.01 and ***p < 0.001. Eta squared for time effects and interaction are presented in brackets.*

TABLE 3 | Squat Jump (SJ), Counter Movement Jump (CMJ), Drop Jump Height (DJH), Drop Jump Contact Time (DJCT), Standing Long Jump (SLJ), 30 m Linear Sprint (30 mLSp), and 30 m Pendulum Sprint (30 mPSp) for both groups during PRE, POST, and FU testing, including change (delta).

	Parameter	Group	PRE	POST	% Delta PRE-POST	SMD PRE-POST	FU	% Delta PRE-FU	SMD PRE-FU	ANOVA p (η_p^2)	
										Time Effect	Time x Group Interaction
Jumps	SJ (cm)	INT	36.34 (6.09)	38.28 (5.73)	+5.3	0.33	40.17 (6.19)	+10.5	**0.62**	0.313 (0.075)	0.942 (0.004)
		CON	34.89 (3.02)	36.57 (5.59)	+4.8	0.37	38.48 (4.30)	+10.3	**0.97**		
	CMJ (cm)	INT	42.48 (7.46)	41.93 (6.21)	−1.3	−0.08	43.50 (6.93)	+2.4	0.14	**0.038 (0.195)**	0.664 (0.027)
		CON	39.18 (4.36)	41.06 (4.89)	+4.8	0.41	41.03 (4.92)	+4.7	0.40		
	DJH (cm)	INT	31.60 (4.94)	33.16 (3.74)	+4.9	0.36	33.63 (3.44)	+6.4	0.48	**0.003 (0.328)**	0.971 (0.002
		CON	31.50 (4.10)	32.74 (3.94)	+3.9	0.31	33.40 (3.34)	+6.0	**0.51**		
	DJCT (cm)	INT	178.89 (11.88)	177.22 (13.76)	−0.9	−0.13	177.67 (16.66)	−0.7	-0.08	0.995 (0.000)	0.834 (0.012)
		CON	166.78 (16.97)	161.89 (21.20)	−2.9	−0.26	165.22 (21.91)	−0.9	-0.08		
	SLJ (cm)	INT	225.33 (17.99)	232.22 (15.50)	+3.1	0.41	240.11 (17.97)	+6.6	**0.82**	0.429 (0.055)	0.691 (0.024)
		CON	224.00 (14.82)	235.33 (17.59)	+5.1	**0.70**	240.56 (21.02)	+7.4	**0.91**		
Sprints	30 mLSp(s)	INT	4.16 (0.10)	4.29 (0.10)	+3.1**	**1.30**	4.18 (0.13)	+0.5	0.17	0.119 (0.151)	**0.007 (0.319)**
		CON	4.24 (0.22)	4.23 (0.20)	−0.2	−0.05	4.19 (0.17)	−1.2	-0.25		
	30 mPSp(s)	INT	7.19 (0.19)	7.19 (0.30)	0.0	0.00	7.13 (0.29)	−0.8	-0.25	**0.049 (0.207)**	0.437 (0.062
		CON	7.25 (0.36)	7.11 (0.25)	−1.9	−0.45	7.11 (0.25)	−1.9	-0.45		

*Values are presented as mean (±SD). Moderate to large SMD have been highlighted in bold. Significance level was set at **$p < 0.01$. Eta squared for time effects and interaction are presented in brackets.*

improvements in F_{max} are comparable with previous results for local muscles with superimposed EMS obtained from Nobbs and Rhodes (1986) and Portmann and Montpetit (1991) who conducted dynamic training sessions. Studies which conducted isometric electrical stimulations at the maximal individual pain threshold could showed even higher improvements in F_{max} ranging from +33 ± 18% (Selkowitz, 1985; Cabric and Appell, 1987; Miller and Thépaut-Mathieu, 1993; Colson et al., 2000). Smaller improvements in our study might be attributed to the submaximal EMS intensity, which was selected to perform dynamic exercises. A significant time × group interaction effect was merely observed for F_{max} on LE. Despite a lack of significance level of the interaction effect, we at least observed moderate to large effect sizes for F_{max} on LC. However, subsequent pairwise comparison indicated that INT benefited from WB-EMS small to moderate extend (SMD = 0.57). Therefore, the use of superimposed WB-EMS might be a beneficial means to further improve maximal strength in the quadriceps and in tendency in hamstring muscles. Additional motor unit recruitment through EMS with (a) a continuous and exhausting contractile activity in the same pool of motor units during the entire exercise period, (b) a supramaximal temporal recruitment imposed by the high stimulation frequency chosen, and (c) a synchronous recruitment of neighboring muscle fibers might account for these strength gains (Requena Sánchez et al., 2005).

The EMS specific improvements of the hamstring muscles seem to be particularly important and interesting. Hamstring injuries are the most common muscle injuries in team sports (Ekstrand et al., 2011). A meaningful association between the susceptibility of hamstring injuries and a low hamstring/quadriceps ratio has been early proposed (Orchard et al., 1997). The activation of the hamstrings as well as quadriceps muscles were potentially higher with superimposed EMS during all exercises, especially during exercises like split squats or glute-ham rises in which the quadriceps or hamstrings have a primary agonistic or even antagonistic function. Therefore, the simultaneous stimulation of agonistic and antagonistic working muscles through WB-EMS does not seem to be as counterproductive as assumed in order to improve F_{max} and P_{max}. The antagonistic working muscles have to adapt to the EMS-induced resistance during eccentric and concentric contractions. This might be especially favorable during the concurrent application of EMS in eccentric muscular actions (Requena Sánchez et al., 2005).

The findings of the present study indicated that the highest improvements in F_{max} and P_{max} occurred 2 weeks after the training intervention. The calculation of SMD revealed larger effect sizes in all strength and power parameters for PRE-FU compared to PRE-POST for the INT group. The delayed adaptations caused by WB-EMS are also described in previous WB-EMS studies from Dörmann (2011) and Wirtz et al. (2016). Consequently, even a submaximal dynamic WB-EMS training seems to need prolonged regeneration periods after training to reach maximal adaptations. These prolonged adaptations might be explained with the accentuated activation of fast motor units at relatively low force levels (Gregory and Bickel, 2005) and the continuous and exhausting contractile activity in the same pool of motor units during the entire exercise period (Requena Sánchez et al., 2005). Based on these results, it can be speculate if a longer regeneration period after training leads to greater adaptations. If so, this would be an indication to apply EMS intermittently to realize similar improvements with a reduced training volume. Maffiuletti et al. (2000) were able to achieve and maintain significant improvements in various strength parameters within 4 weeks after a 4-week intervention with maximal isometric local EMS. Additionally, superimposed WB-EMS offers the opportunity to might shorten the duration of strength training through the intensification of traditional voluntary strength training without increasing the number of training sessions per week (Filipovic et al., 2012). The high attendance rate in the training sessions of the present study indicates that all participants were able to cope with the physical requirements. In particular the WB-EMS training with a submaximal EMS intensity of about 70% of the maximal individual pain threshold seems to be a beneficial compromise to achieve strength and power adaptations as well as to have an appropriate exertional tolerance.

Despite the considerable gains in F_{max} and P_{max}, the improvements in the jumping and sprinting tests were restricted. The only significant time × group interaction effect was observed for 30 mLSp with a significant decline in sprinting performance from PRE to POST and a significant improvement from POST to FU for the INT. These results are in consonance with further studies that observed a change in sprint performance after performing dynamic movements with superimposed WB-EMS (Filipovic et al., 2016). However, the sport-specific training orientation of the jumping and sprinting exercises with superimposed WB-EMS did not lead to the suspected results in the majority of the secondary endpoints. Nevertheless, previous studies indicate that dynamic EMS should be combined with additional athletic or plyometric training to better transfer the strength gains into movements like jumping or sprinting (Maffiuletti et al., 2002; Requena Sánchez et al., 2005; Filipovic et al., 2012). Maffiuletti et al. (2002) conducted a combination of separate isometric EMS sessions and separate plyometric jumping sessions during a 4-week training period and improved SJ performance by +21%. Studies that enhance sprint performance also used EMS in combination with separate sprint-specific or plyometric training (Brocherie et al., 2005; Herrero et al., 2006, 2010). In order to improve sport-specific abilities, the simultaneous activation of agonistic and antagonistic working muscles through WB-EMS in combination with jumping or sprinting exercises at the same time does not appear to be the most effective method as shown in the present study. It can be assumed that the recruitment pattern of WB-EMS disturbs the complex coordination of voluntary muscle activation during explosive performed jumps or sprints. The possible advantage of the simultaneous recruitment pattern of WB-EMS for maximal strength improvements with a reduced co-activation of antagonistic muscles consequently seems to be questionable. However, another explanation for the reduced adaptations of jumping and sprinting abilities could be the short regeneration time after the intervention. In our study, all parameters of the INT group increased within 2 weeks after completion of the WB-EMS intervention from POST-FU. It can be assumed, that strength and power improvements due to EMS training cannot be immediately transferred into complex sport-specific movements despite a sport-specific orientation of training exercises. Voluntary recruitment patterns of special movements are necessary and time for a conversion phase of WB-EMS induced improvements take at least 2 weeks. Furthermore, recent studies suggest that the strength gains achieved with WB-EMS may require a longer adaptation period (>7 weeks) or a higher number of EMS sessions per week to be transferred into sport specific movements (Filipovic et al., 2016).

Regarding future research, some limitations of the present study need to be addressed. Due to the relatively complex study protocol including 12 weeks of training/testing and supervised training twice a week for every participant, only a small sample size has been generated. A larger sample size could have increased the study power and might have provided more conclusive results. The second limitation is that the training stimulus of the exercise "Bulgarian Split Squat" was different between INT and CON. Training intensity of both intervention groups was controlled by the Borg RPE-scale and set to >16 (>"hard"). With regard to previous results from Wirtz et al. (2016) which showed that high mechanical loads in combination with WB-EMS do not lead to greater adaptations than a similar training without WB-EMS, training intensity should rather be intensified with the exercise itself or the intensification of the electrical stimulation than with additional weight. Even if this approach was feasible for the INT group, the participants of the CON group needed additional weights for the split squat exercise after 3 weeks of training because of the training progression reaching 16 at the Borg RPE-Scale.

CONCLUSION

In conclusion, it seems that dynamic submaximal superimposed WB-EMS training does not provide notable additional improvements in maximal strength and power parameters of the leg muscles of moderately trained, young athletes compared with a similar training intervention without superimposed WB-EMS. Only in the LE F_{max}, the INT group could achieve greater improvements than the CON group. Improvements in complex sport-specific movements like jumping or sprinting are restrictive despite a sport-specific orientation of the training exercises.

AUTHOR CONTRIBUTIONS

HK, FM, UD, and NW contributed to conception and design of the study. FM, UD, and NW organized and conducted the training intervention and data acquisition. FM and LD performed the statistical analysis. FM wrote the first draft of the manuscript. All authors contributed to manuscript revision, read, and approved the submitted version.

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FM orally presented parts of the data set at two conferences (23rd Annual Congress of the European College of Sport Science and 10th Annual Meeting of the Swiss Society of Sport Science) for dissemination reasons.

SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2018.01719/full#supplementary-material

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OPINION
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Attractive Gait Training: Applying Dynamical Systems Theory to the Improvement of Locomotor Performance Across the Lifespan

Bas Van Hooren[1,2], Kenneth Meijer[1] and Christopher McCrum[1,3,]*

[1] Department of Nutrition and Movement Sciences, NUTRIM School of Nutrition and Translational Research in Metabolism, Maastricht University Medical Centre+, Maastricht, Netherlands, [2] Institute of Sport Studies, Fontys University of Applied Sciences, Eindhoven, Netherlands, [3] Institute of Movement and Sport Gerontology, German Sport University Cologne, Cologne, Germany

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German Sport University Cologne, Germany

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Daniel Hamacher,
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****Correspondence:***
Christopher McCrum
chris.mccrum@maastrichtuniversity.nl

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INTRODUCTION

Healthy adult humans can walk and run with ease, yet it takes years to develop stable and economical locomotion. This apparent ease is the result of multiple degrees of freedom at dozens of joints being controlled by hundreds of muscles, all recruited and activated with precise timing and frequency by the neuromotor system (Turvey, 1990; Pandy and Andriacchi, 2010; Latash, 2012). Despite the multiple degrees of freedom resulting from this abundance, as well as the variation across individuals, bipedal gaits that emerge from this system (i.e., walking and running) are remarkably similar. According to dynamical systems theory, these similarities in behavior emerge because of attractors (Kelso et al., 1981; Kelso, 2012). Specifically, limit cycle attractors may be primarily responsible for the convergence of joint motion to form the periodic behavior in gait (Ijspeert, 2008; Broscheid et al., 2018).

Attractors represent coordination tendencies among system components (Davids et al., 2008), can be identified at multiple levels and emerge from the self-organization of the lower and higher-level components through circular causality (Haken, 1987). This means that the behavior of components at a higher level will be influenced (i.e., constrained) in a bottom-up manner by the behavior of components at the lower level and vice versa. With regards to locomotion, the two distinct human gaits, walking and running, represent two attractors at the macroscopic level (Diedrich and Warren, 1995; Lamoth et al., 2009), relative to the joint coupling of the ankle, knee and hip joints during these gaits that represents an attractor at a mesoscopic level (Diedrich and Warren, 1995), relative to the rhythmic neural activity of the central pattern generators that represents an attractor at a microscopic level (Cappellini et al., 2006; Ijspeert, 2008; Minassian et al., 2017) (**Figure 1**). Please note that we use micro-, meso- and macroscopic in relative terms, whereby the components at a mesoscopic level are macroscopic level components relative to the microscopic components.

Attractors in human locomotion may serve different purposes such as optimizing energetic and mechanical efficiency (Selinger et al., 2015; Kung et al., 2018), minimizing mechanical load (Kung et al., 2018), maintaining stability (Jordan et al., 2007) and increasing the robustness of the motion to perturbations resulting from internal (physiology) and external (environment) sources (Santuz et al., 2018). A decrease in stability of an attractor and an increased variability can induce a spontaneous phase transition to another attractor. An example of this in human locomotion is the walk to run transition, whereby a decreased stability of phase relationships of walking gait and increased variability in out-phase ankle-hip joint coupling with increasing walking speed results in a sudden self-organized transition around a speed of 2 m/s to the more stable running gait with a more in-phase joint coupling (Diedrich and Warren, 1995; Lamoth et al., 2009; Kung et al., 2018).

Other factors such as training (Zanone and Kelso, 1992; Kostrubiec et al., 2012), fatigue and aging (Stergiou and Decker, 2011) may also affect the magnitude of variability or the stability of the movement system and thereby induce a phase transition to a less optimal systems behavior. This may lead to reduced performance, injuries and pain. As effective locomotion is key to success in many sports, but also a necessity for functioning in daily life, understanding how the stability of the movement system in walking and running can be improved through training to enhance performance and reduce fatigue and injury would be beneficial. With the above points in mind, we propose that dynamical systems theory provides a framework for understanding locomotion and improving the effectiveness of interventions in locomotion-related problems. In this opinion article, we discuss two locomotor-related problems in populations of very different capacities, namely, falls among older adults and ankle sprain injuries in runners, and discuss how the application of dynamical systems theory can lead to novel approaches to intervention. Specifically, we hypothesize that applying small perturbations during locomotion may be effective for modifying the stability of specific locomotor attractors. Although using variability to improve performance is not novel [see studies on the variability of practice hypothesis (Schmidt, 1975), contextual interference effect (Magill and Hall, 1990) or differential learning (Beckmann and Schöllhorn, 2003; Schöllhorn et al., 2010; Serrien et al., 2018)], studies on these topics have mostly investigated discrete skills such as football kicking and shot-put and induced variability by practicing different movements rather than applying small unexpected perturbations during the actual movement to be improved. The use of small unexpected perturbations during movement may represent an effective way to further enhance performance.

DYNAMICAL SYSTEMS THEORY AND ANKLE SPRAIN INJURIES IN RUNNING

Running is a gait fundamental to many sports, but also an activity that is associated with a high injury incidence. A significant proportion of individuals participating in running or running-based sports sustain acute injuries such as ankle sprains (Van Gent et al., 2007; Fong et al., 2009). An ankle sprain can occur when high impact forces induce rapid inversion during ground contact, in particular when running on an irregular surface such as grass, sand or uneven sidewalks, or when changing direction. This inversion results in excessive stretch of the lateral ligaments that may lead to large strains or rupture (Fong et al., 2009).

Traditional approaches to ankle injury prevention and rehabilitation have applied training such as balancing on a wobble board or on one leg with the eyes closed (Schiftan et al., 2015). Although these approaches have been shown to be effective at preventing re-injuries in individuals with a history of ankle sprains, the evidence is less conclusive for ankle sprain prevention in individuals with no prior injury (Schiftan et al., 2015). Up to 70% of individuals with ankle sprain injuries report incomplete recovery and are therefore at a higher risk of re-injury (Anandacoomarasamy and Barnsley, 2005). One reason for the less conclusive evidence regarding primary prevention and incomplete recovery following an ankle sprain may be the different ways in which perturbations are corrected in traditional balance training and high-intensity movements such as running. According to the dynamical systems theory, a phase transition may occur from a combined reflex and preflex based correction of perturbations during tasks with no or minimum time pressure (e.g., traditional balance training on an unstable surface or slow walking) to a more preflex dominant correction in tasks with high time pressure such as the ground contact during high-speed running on uneven grass (Bosch, 2015). Reflexes may be strong and fast enough to correct smaller perturbations during traditional balance training, whereas they may be too slow and insufficient to prevent the effects of perturbations during (high-speed) running. Indeed, using a computational model of ankle inversion, DeMers et al. (2017) showed that reflexes took at least 80 ms to partly correct the perturbation, but failed to fully prevent excessive inversion. In contrast, moderate to high levels of co-activation were able to correct the perturbation within 60 ms due to the force-length-velocity properties of the muscle fiber and tendon elasticity, also known as preflexes (Loeb et al., 1999).

Applying small perturbations during running could potentially improve the robustness of running gait and in particular the motions of the ankle to perturbations by modifying the stability of the attractor via mechanisms such as alterations in step width and muscle activation at a mesoscopic level (Santuz et al., 2018). In the long-term, these acute mechanisms may translate into a more robust running gait pattern that is more prone to injuries via alteration in contact times and stride frequencies. We hypothesize that applying small perturbations during running may therefore be more effective for prevention and rehabilitation of ankle sprains compared to traditional balance training without time pressure, although further research is required to substantiate this notion. Also note that both approaches can complement each other and are not necessarily mutually exclusive.

DYNAMICAL SYSTEMS THEORY AND FALLS PREVENTION IN OLDER ADULTS

Walking is an essential gait for daily life but is also accompanied with an increased risk of falls with increasing age (Berg et al., 1997; Talbot et al., 2005). If we exclude environmental influences, we can address an individual's falls risk by looking either at the stability of their steady-state gait or at their behavior when they are brought out of balance to the extent that the locomotor behavior is altered. The latter has been the focus of much research, as many falls occur due to slipping or tripping (Berg et al., 1997; Talbot et al., 2005). Research on perturbation-based balance training has proliferated recently, during which the recovery reactions to sudden perturbations to balance or gait are trained (Mansfield et al., 2015; Okubo et al., 2016; Gerards et al., 2017; McCrum et al., 2017). However, as well as studying the reactive responses following balance loss, it is important to consider how the balance loss comes about, and if the robustness of the gait pattern to perturbations can

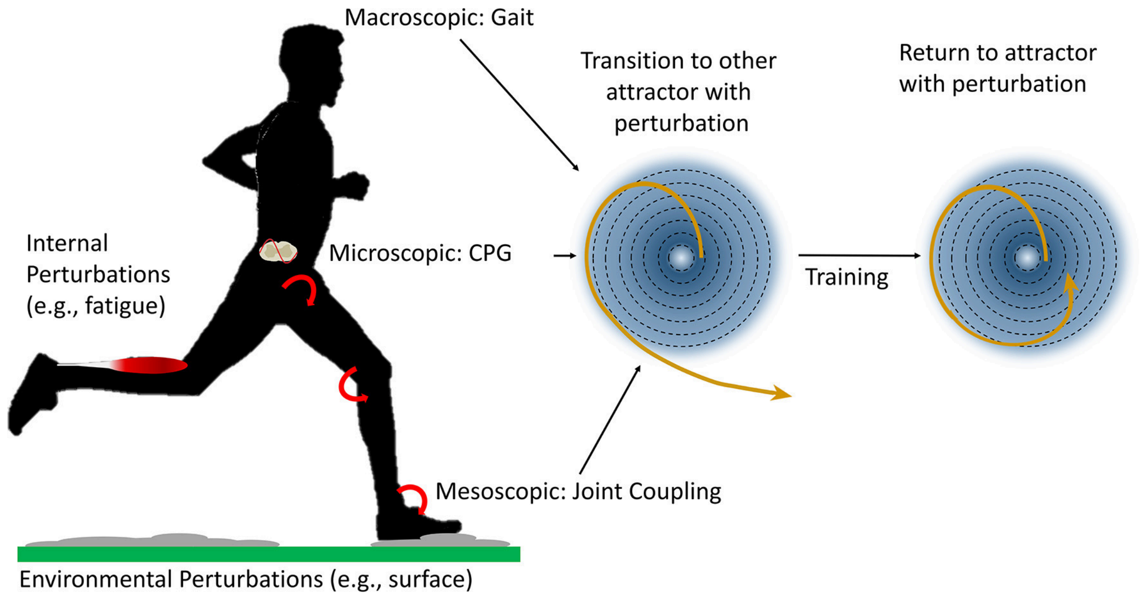

FIGURE 1 | In bipedal gaits such as walking and running, gait, joint coupling and central pattern generators (CPG) may represent limit cycle attractors (coordination tendencies that cyclically repeat, simplified represented on the right in two dimensions based on Kitano, 2004) at a macroscopic, mesoscopic and microscopic level relative to each other when viewed from a top-down perspective, respectively. Internal and external perturbations from for example fatigue or an uneven surface can lead to a phase transition to another potentially less effective or less efficient (limit cycle) attractor. Large perturbations may for example lead to problems such as falls during walking or an ankle sprains while running. Training may increase the stability of attractors so that perturbations of a larger magnitude or perhaps greater frequency or unpredictability can be accommodated without a loss of stability.

be improved. In this context, rather than just using large perturbations that bring people out of balance, dynamical systems theory would suggest that applying smaller perturbations during gait that do not require a complete switch of locomotor behavior (phase transition in attractors) may also lead to positive improvements in gait stability via an increased robustness of the movement patterns. Both coping with small perturbations without a significant change in gait behavior and with large perturbations that do require some explicit recovery movements have previously been suggested as key requirements for stable gait (Bruijn et al., 2013), and both show significant declines with increasing age (Maki and Mcilroy, 2006; Süptitz et al., 2013; Terrier and Reynard, 2015; McCrum et al., 2016). The importance of studying stability during steady-state gait, in addition to reactive stability during larger perturbations, is supported by evidence of the relationship between decreased stability during steady-state gait and falls incidence (Hausdorff et al., 2001; Van Schooten et al., 2016; Bizovska et al., 2018).

Through the application of small perturbations during steady-state walking, the stability of specific locomotion attractors may be modified. One study has demonstrated alterations in motor primitives while walking and running on uneven, compared with even surfaces, creating activation patterns that were more robust to the perturbations (Santuz et al., 2018). If the basins of attraction of limit cycle attractors could be modified in older adults, this could mean that perturbations of a larger magnitude (or perhaps greater frequency or unpredictability) could be accommodated without significant loss in dynamic stability. For example, while walking over uneven ground, more frequent or larger undulations in the surface could be negotiated without loss of dynamic stability and without the need for subsequent large reactive balance corrections. One recent study had older participants walk on a treadmill with stable and unstable (water) loads in a backpack (Walsh et al., 2018). As would be expected, step variability was increased, and mediolateral dynamic stability decreased in the unstable load condition and electromyography activity was also increased to cope with the load (Walsh et al., 2018). If practiced over longer time periods, a more robust gait pattern may be the result via alterations such as step width or time, joint moments at the ankle to control center of mass velocity and muscle activation and motor primitives at a mesoscopic level. Further research is needed to examine the training effects of walking with small continuous unexpected perturbations and whether this translates to a more robust response to large perturbations and subsequently reduced falls risk, but such training represents one interesting avenue for future falls prevention interventions.

CONCLUSION

Human locomotion can be conceptualized as a behavior of a dynamical system, with attractors that serve different purposes such as optimizing energetic and mechanical efficiency, minimizing mechanical load, maintaining stability and increasing the robustness of the motion to perturbations resulting from internal (physiology) and external (environment) sources. We have proposed that through the application of small perturbations during walking and running, the basin of attraction for specific locomotion attractors may be modified, which may have benefits for both maintaining gait stability and for reducing injury risk. Further research is required to elucidate the effectiveness of such interventions in different populations.

AUTHOR CONTRIBUTIONS

BVH and CM: Conception of the work; drafted the article. BVH: Prepared figure. All authors participated in discussions of the literature and concepts, reviewed and revised the article and approved the final version of the article.

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Gait Stability and Its Influencing Factors in Older Adults

Daniel Hamacher[1]*, Dominik Liebl[2], Claudia Hödl[1], Veronika Heßler[1], Christoph K. Kniewasser[1], Thomas Thönnessen[1] and Astrid Zech[1]

[1] Institute of Sport Science, Friedrich Schiller University of Jena, Jena, Germany, [2] Department of Statistics, University of Bonn, Bonn, Germany

A stable gait pattern is a prerequisite to successfully master various activities of daily living. Furthermore, reduced gait stability is associated with a higher risk of falling. To provide specific intervention strategies to improve gait stability, gaining detailed knowledge of the underlying mechanism and influencing factors is of utmost importance. The effects of relevant influencing factors on gait stability are poorly examined, yet. Therefore, the aim of the current study was to quantify the effects of various influencing factors on gait stability. In a cross-sectional study, we assessed dynamic gait stability and relevant influencing factors in 102 older adults (age >65 years). In addition to dynamic gait stability (largest Lyapunov exponent [LLE] and gait variability measures) during normal over-ground (single-task: ST) and dual-task (DT) walking, we registered the following influencing factors: health status (SF12), pain status (painDETECT, SES), fear of falling (falls efficacy scale), depression (CES-D), cognition performance (Stroop test), physical activity (Freiburger Fragebogen zur körperlichen Aktivität), proprioception (joint position sense), peripheral sensation (mechanical and vibration detection threshold), balance performance (static balance on force plate) and muscular fitness (instrumented sit-to-stand test). We used a principal components regression to link the identified principal components with the gait stability and gait variability responses. The four principal components "strength and gender" (e.g., $p = 0.001$ for LLE during ST), "physical activity" (e.g., $p = 0.006$ for LLE during ST), "pain" (e.g., $p = 0.030$ for LLE during DT) and "peripheral sensation" (e.g., $p = 0.002$ for LLE during ST) were each significantly associated with at least two of the analyzed gait stability/variability measures. The dimension "balance" was a significant predictor in only one gait measure. While "proprioception" tends to correlate with a gait variability measure, we did not find a dependency of mental health on any gait measure. In conclusion, the participants' ability to recover from small perturbations (as measured with the largest Lyapunov exponent) seems to be related to gender and strength, the amount of physical activity the participants spent every week, peripheral sensation and pain status. Since the explained variance is still rather low, there could be more relevant factors that were not addressed, yet.

Keywords: balance, muscular fitness, physical activity, pain, peripheral sensation, gender, proprioception, gait variability

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Lars Donath,
German Sport University Cologne,
Germany

Reviewed by:
Sridhar Poosapadi Arjunan,
RMIT University, Australia
Eduard Kurz,
University Hospital in Halle, Germany

****Correspondence:***
Daniel Hamacher
daniel.hamacher@uni-jena.de

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INTRODUCTION

Falls in the elderly occur quite frequently. 30–60% of older adults fall at least once a year (Rubenstein, 2006; Inouye et al., 2009) and up to 20% of falls result in injuries (Rubenstein, 2006), hospitalization or death in older adults (Rubenstein, 2006; Pfortmueller et al., 2014). Western health care systems spend 0.85 to 1.5% of their total health care expenditures (Heinrich et al., 2010) on the consequences of falls, rendering this issue socio-economically relevant. Considering the effect on the individual's quality of life, the prevention of falls is of utmost importance.

Risk factors for falls and fall-related injuries can be categorized into (I) environmental risk factors, (II) behavioral risk factors, (III) biological risk factors, and (IV) socioeconomic risk factors. However, besides standing quietly and sitting down, walking is one of the most common daily activities leading to falls (Robinovitch et al., 2013). Thus, it is not surprising that an unstable gait pattern is associated with a higher risk of falling in older adults (Hamacher et al., 2011; Bruijn et al., 2013). In addition, walking is a prerequisite to handle a wide range of daily activities (Bramble and Lieberman, 2004). Since gait measures are predictors of mobility (Brach et al., 2007), a stable gait also ensures social participation not only in the elderly. This highlights the importance of a stable gait in various contexts.

According to the literature, age is a strong predictor of falls (Lord et al., 2003; Ambrose et al., 2013; Pfortmueller et al., 2014) as well as of an unstable gait (Terrier and Reynard, 2015). Other (predominantly age-related) intrinsic determinants are frequently discussed as risk factors for falls. These intrinsic determinants include gender, muscle strength or muscle power, balance, peripheral sensation (proprioception, vibration sense, tactile sensitivity), cognition, and diseases, such as diabetes mellitus (Lord et al., 2003; Ambrose et al., 2013; Pfortmueller et al., 2014). Even depression seems to be a factor related to an altered gait pattern and the probability of falling (Paleacu et al., 2007; Ambrose et al., 2013). Furthermore, gait characteristics (Ambrose et al., 2013) and in particular gait variability and gait stability measures (Hamacher et al., 2011; Bruijn et al., 2013) or the ability to dual-task during gait (Ambrose et al., 2013) are also highly associated with the risk of falling. Although this may indicate interactions between the above-mentioned person-related determinants and walking abilities in older adults, the effects of intrinsic factors on gait variability or gait stability itself were barely reported in the literature. Since knowledge of relevant factors affecting a stable gait is a prerequisite to (1) develop and to evaluate specific intervention strategies, (2) improve gait stability, (3) reduce the number of falls, or (4) ensure save social participation in older adults, we aimed to assess the effects of intrinsic factors associated with falling on gait stability and gait variability as well as on dual-task walking in older adults.

In general, exercise interventions are capable to improve gait variability (Wollesen et al., 2015) and, furthermore, an improved gait pattern enhances physical functioning, physical activity and social participation (VanSwearingen et al., 2011). However, to be able to deduce individually adjusted fall prevention programs (Pfortmueller et al., 2014) that are more efficient and to better understand the underlying mechanisms of gait stability, influencing factors must be identified.

In a broader sense, a stable gait is a gait pattern that does not lead to falls. There are different types of gait stability (Bruijn et al., 2013): for example, dealing with small internal (e.g., neuromuscular noise) and external perturbations (e.g., surface friction) during normal overground walking or recovering from larger perturbations (e.g., a trip or a slip). This study will focus on the former type. Here, regarding fall risk, variability measures and the largest Lyapunov exponent depict the best construct, predictive, and convergent validity (Bruijn et al., 2013) and are, thus, chosen for this study.

Therefore, the aim of the current study was to explore the effect of all above-mentioned intrinsic risk factors on gait stability and gait variability during normal and dual-task walking in older adults. Additionally, factors that interfere with a stable gait, e.g., pain (Hamacher et al., 2016), osteoarthritis and having a joint replacement (Yakhdani et al., 2010; Hamacher, 2014), were also considered.

MATERIALS AND METHODS

Study Design and Participants

In a cross-sectional study, 102 (52 female and 50 male) healthy older adults with a mean age of 72 years (SD = 4.5 years) and a mean body mass index of 27 (SD = 3.6) were recruited using a newspaper announcement. Inclusion criteria were an age of at least 65 years and the ability to walk for 10 min without any aids. Acute neurological, orthopedic or cardiovascular diseases lead to exclusion. Participants with common age-related diseases, such as diabetes mellitus, osteoarthritis, high blood pressure or an implanted prosthesis, were included in the study. This study was carried out in accordance with the recommendations of the Declaration of Helsinki with written informed consent from all subjects. The protocol was approved by the Ethical Commission of the Faculty of Social and Behavioral Sciences, Friedrich Schiller University of Jena (no. FSV 16/05).

Testing Procedure

Each participant came at 2 different days within 2 weeks to complete the tests. At the first test day, a standardized gait analysis was conducted. At the second test day, influencing factors were registered. In total, all tests lasted 3.5 h. A summary of all outcomes is given in **Tables 1**, **2**.

Anamnesis

To check the inclusion and exclusion criteria, diseases, motor-functional complaints, and medication were registered. The participants were explicitly asked if they have diabetes mellitus, osteoarthritis in any joint of the lower extremities or any kind of prosthesis at the lower extremities. To be able to better describe our cohort, we also asked the participants how frequently they have fallen (while walking or standing) within the last 12 months.

Gait Analysis

To assess gait parameters, a standardized gait analysis was conducted in an empty sports hall. Thereto, inertial sensors

TABLE 1 | Overview of the dependent variables.

Dependent variables			Abbreviation
Primary analysis	Gait stability	Local dynamic stability (LDS) of the foot during single-task (ST) walking	$LDS_{foot,ST}$
		Local dynamic stability (LDS) of the foot during dual-task (DT) walking	$LDS_{foot, DT}$
		Local dynamic stability (LDS) of the trunk during single-task (ST) walking	$LDS_{trunk, ST}$
		Local dynamic stability (LDS) of the trunk during dual-task (DT) walking	$LDS_{trunk, DT}$
Secondary analysis	Gait variability	Stride-to-stride standard deviation (SD) of stride length during single-task walking	$SD_{StrideLength,ST}$
		Stride-to-stride standard deviation (SD) of stride length during dual-task walking	$SD_{StrideLength, DT}$
		Stride-to-stride standard deviation (SD) of stride time during single-task walking	$SD_{StrideTime, ST}$
		Stride-to-stride standard deviation (SD) of stride time during dual-task walking	$SD_{StrideTime, DT}$

(MTw2, Xsens Technologies B.V., Enschede, The Netherlands, range of measurement of angular velocity: ±1,200 deg/s, sampling rate: 100 Hz) were attached to the dominant forefoot with tape and to the thorax (strap underneath the arms, thus sensor at upper thoracic spine) with an elastic strap. The dominant foot was identified using the German version of the Lateral Preference Inventory (Ehrenstein and Arnold-Schulz-Gahmen, 1997). To improve the reliability of gait measures, the participants walked on a 25 m track up and down once with their comfortable walking pace to familiarize to the test setup (Hamacher et al., 2017). Thereafter, the participants completed the following conditions in randomized and balanced order: (a) Motor single-task condition: walking up and down the 25 m track with their comfortable walking pace for 4 min; (b) Dual-task condition: Walking with comfortable walking pace while reciting serial three subtractions (starting from a random three-digit number) for 4 min.

From the kinematic walking time series, the following gait parameters were calculated: stride length and stride time as well as the intra-individual stride-to-stride variability (standard deviations) of stride length and stride time as measures of gait variability. The reliability of the measurement system is verified (Hamacher et al., 2014).

As a measure of local dynamic gait stability (LDS), the short-time largest Lyapunov exponent was determined for foot and trunk kinematics separately using an evaluated algorithm (Hamacher et al., 2015). Since for normal overground walking, the highest effects comparing young vs. older adults were observed when analyzing time series derived from three-dimensional angular velocity data of the foot (Hamacher et al., 2015), we used those time series, too. For each participant, the three-dimensional angular velocity data of 100 strides were time-normalized to 10,000 samples, resulting in an average of 100 samples per stride. Thereafter, a state space was built upon on the time-normalized data using the method of time-delayed embedding. The time delay and embedded dimension were chosen based on the first minimal mutual information (Fraser and Swinney, 1986) and the false nearest neighbors analysis (Kennel et al., 1992), respectively. A fixed time-delay τ (mean across all participants: $\tau_{foot} = 9$, $\tau_{trunk} = 11$) and embedded dimension dE (maximum across all participants: $dE_{foot} = dE_{trunk} = 12$) was used for all participants. The largest Lyapunov exponent was calculated using Rosenstein and coworkers' algorithm (Rosenstein et al., 1993). Thereto, the Euclidean distances of each point in state space of initially nearest neighbors were tracked in time and the mean of the logarithm of this divergence curves was calculated. The largest Lyapunov exponent was defined as the slope of the linear fit through approximately 0–0.5 strides. Larger values indicate lower local dynamic gait stability. The largest Lyapunov exponent quantifies the ability of a dynamic system (human gait) to recover from small perturbations (Bruijn et al., 2013).

Compared to the gait variability measures, the largest Lyapunov exponent depict a slightly better construct validity (Bruijn et al., 2013). Therefore, the outcomes local dynamic stability (LDS) of trunk and foot during single-task and dual-task walking were considered primary criterions to be predicted by the assumed influencing outcomes. In a secondary analysis, the gait variability parameters SD of stride length and SD of stride time during single-task and dual-task walking were analyzed.

Muscular Fitness

To measure muscular power of the lower extremities, the sit-to-stand transfer has already been successfully conducted in older adults (Lindemann et al., 2003; Zech et al., 2011; Zijlstra et al., 2012). We used an instrumented version to assess muscular power (Zijlstra et al., 2010). Thereto, an inertial sensor (MTw2, Xsens Technologies B.V., Enschede, The Netherlands, sampling rate: 100 Hz) was fixed to the back of the pelvis. Compared to a force plate based approach, power calculated from inertial sensor data fixed to the pelvis depict high correlations ($r = 0.95$ for fast movements, Zijlstra et al., 2010). The participants were placed on the front part of a chair (height: 0.47 m, no armrests, arms crossed over the chest). They were asked to stand up as fast as possible. The participants were asked to sit/stand motionless immediately before and after the sit-to-stand transition. As described below, this was used as a boundary condition for parameter calculation.

Using the sensors orientation (quaternions) and the three-dimensional accelerometer data, the vertical component of the acceleration data was extracted, and gravitational acceleration was removed by subtracting 9.81 m/s^2. Vertical movement velocity was calculated by numerical integration (Heun's method). Prior to and after the sit-to-stand transition, the movement velocity is considered zero. Any deviations from zero (e.g., due to the numerical integration) were removed by subtracting a straight line which was fitted through two points:

TABLE 2 | Overview of all independent variables.

Variables	Abbreviation	
Anamnesis	Age	
	Gender	
	Body mass index	BMI
	Osteoarthritis	
	Any kind of joint replacement	Prosthesis
Muscular fitness	Normalized Peak Power during the sit-to-stand test	$S2S_{PeakV}$
	Normalized Mean Power during the sit-to-stand test	$S2S_{MeanV}$
Balance	Sway during double-leg (dl) stance with eyes open (eo)	$Sway_{dl,eo}$
	Sway during semi-tandem (st) stance with eyes open (eo)	$Sway_{st,eo}$
	Sway during double-leg stance (dl) with eyes closed (ec)	$Sway_{dl,ec}$
	Sway during semi-tandem stance (st) with eyes closed (ec)	$Sway_{st,ec}$
Pain	Neuropathic component (painDETECT)	$Pain_{neuro}$
	Affective component (SES)	$Pain_{affect}$
	Sensory component (SES)	$Pain_{sens}$
	sub-scales rhythmicity of the sensory component (SES)	$Pain_{sens,rhythm.}$
	sub-scales local depth of the sensory component (SES)	$Pain_{sens,depth}$
	sub-scales temperature of the sensory component (SES)	$Pain_{sens,temp.}$
Cognition	Time needed for incongruent stimulus condition (Stoop test)	Cog_{ink}
	Time costs for the incongruent stimulus (compared to the ink-naming condition, Stroop test)	Cog_{relink}
Fear of Falling	FES-I score	FES-I
Depression	CES-D score	CES-D
Health status	Physical component summary score of the SF-12	$SF12p_{hysical}$
	Mental component summary score of the SF-12	$SF12m_{ental}$
Peripheral sensation	Vibration detection threshold of the “quantitative sensory testing” battery	$Sens_{VibDT}$
	Mechanical detection threshold (MDT) of the “quantitative sensory testing” battery	$Sens_{MechDT}$
Proprioception	Mean of the absolute error of an active/active angle reproduction test	$Prop_{MeanErr}$
	Standard deviation of the error of an active/active angle reproduction test	$Prop_{SDErr}$
Physical activity	Total sum of physical activity (FFkA questionnaire)	$FFkA_{total}$
	Basic (common daily activities) physical activity (FFkA questionnaire)	$FFkA_{basic}$
	Extracurricular physical activity (FFkA questionnaire)	$FFkA_{Extracurr}$
	Sports activity (FFkA questionnaire)	$FFkA_{Sports}$

the pre-test and the post-test vertical velocity value. The Power was calculated as the arithmetic product of the vertical movement velocity, gravitational acceleration g, and the body weight. Power was then normalized to the subject’s body weight. Based on the resulting power-time curve, peak power and mean power (mean power during the sit-to-stand transition where the vertical movement velocity was at least 0.1 m/s) were calculated as outcomes.

Balance

Balance performance was assessed during double-leg stance (feet together) and semi-tandem stance (toe of the dominant foot slightly touching the contralateral heel) on a force platform (type 9260AA6, Kistler Instrumente GmbH, Winterthur, Swiss). During the test, the participants did not wear shoes but socks. The two stance conditions were completed with eyes open and eyes closed for 30 s each stance condition. In all stance conditions, hands were held on hips. During the open eyes conditions, the participants were asked to look at a cross placed at eye level 1 m in front of the participant. With the aid of the MARS software (Kistler Instrumente GmbH, Sindelfingen, Germany) the mean velocity of the two-dimensional velocity of the center of pressure was calculated for each balance condition.

Pain

The pain status was assessed using the painDETECT questionnaire and the Pain Experience Scale (German: Schmerzempfindungsskala, SES). The painDETECT questionnaire quantifies the neuropathic component of pain. A higher score depicts a higher likelihood of neuropathic pain being present (Freynhagen et al., 2006).

The SES assesses the (a) affective and (b) sensory dimension of pain. Furthermore, information on (c) rhythmicity, (d) local depth, and (e) temperature are quantified as sub-scales of the sensory dimension of pain (Geissner, 1995).

Cognition

The Color-Word Interference Test (Stroop) is a frequently used test to quantify executive functioning. More specifically, the selective allocation of attention is rated (Lamers et al., 2010). The participants (a) must read color names, (b) name ink colors (no writings), and (c) name ink colors of words. However, in this last condition, the ink color and the color words do not match (incongruent stimulus). As outcomes, we used the time needed for condition c (incongruent stimulus) as well as the time difference of conditions b & c (time costs of the incongruent stimulus). Both outcomes are commonly assessed (Uttl and Graf, 1997).

Fear of Falling

We rated fear of falling using the Falls Efficacy Scale International (FES-I) (Dias et al., 2006). The questionnaire assesses the subjective fear to fall during various common daily activities. The German version of the FES-I was validated and depict good quality criteria (Delbaere et al., 2010). As the outcome, a single value reflecting the amount of fear of falling was calculated.

Depression

To register depression symptoms, we deployed the German version of the Center for Epidemiologic Studies-Depression Scale (CES-D) for screening purposes. This validated questionnaire is suitable for a self-assessment of depressive symptoms in the general population (Meyer and Hautzinger, 2001; Lehr et al., 2008). High outcome scores indicate a high probability of having a depression.

Health Status

The overall health status was assessed with the 12-Item Short-Form Health Survey (SF-12 Ware et al., 1996). Based on that 12 items a physical component summary, as well as a mental health summary score, were analyzed. High scores are interpreted as a good health status.

Peripheral Sensation

Mechanical and vibration detection thresholds were assessed as described in the "quantitative sensory testing" battery (Rolke et al., 2006). To test the mechanical detection threshold, we used Frey filaments (0.5, 1, 2, 4, 8, 16, 32, 64, 128, 256, 512 mN; MARSTOCKnervtest, Schriesheim, Germany). The force was applied to the lateral malleolus of the dominant foot. A series of descending, and ascending stimulus intensities was applied. The threshold was calculated as the geometric mean from five repetitions.

To test the vibration detection threshold, the Rydel–Seiffer graded tuning fork (64 Hz, 8/8 scale) was applied at a proximal elevation of the tibia (tibia tuberositas). The participants were asked to indicate once they did not feel any vibrations anymore. The threshold was then defined as the mean of three repetitions.

Proprioception

Proprioception was tested with an inertial sensor-based reproduction test using an active-active procedure as described

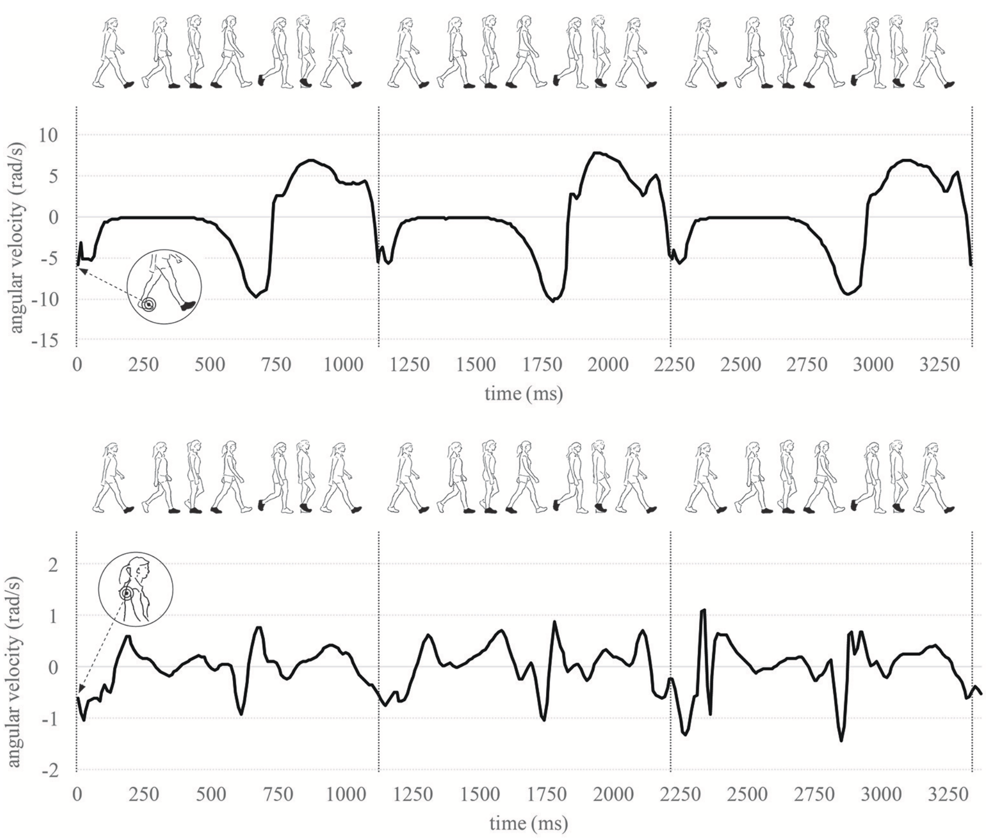

FIGURE 1 | Sensor placement at the foot and trunk. As an example, one part of the sensor signal across three strides is illustrated.

by Arvin et al. (2015). To measure the knee angle in real-time using an in-house software, inertial sensors (MTw2, Xsens Technologies B.V., Enschede, The Netherlands, sampling rate: 100 Hz) were fixed to the dominant shank (medial and distal to the tibia tuberositas) and to the iliotibial tract at the middle of the thigh. The participants stood on a wooden platform keeping their eyes closed. The participants were then asked to slowly flex their knee. The examiner said "stop" once a knee flexion angle of about 40° was reached. Deviations from the target angle up to ±5° were allowed and not corrected to reduce fatiguing effects in the older adults. The participants memorized the knee angle, returned to the start position and reproduced that angle. The mismatch error was registered. This procedure was repeated 10 times, but the first two times were considered learning trials and, thus, not analyzed. As outcomes, we calculated the mean of the absolute error and the standard deviation of the signed error.

Physical Activity

The "Freiburger Fragebogen zur körperlichen Aktivität" (FFkA Frey et al., 1999) is a validated instrument to register health-related physical activity (PA, in hours per week) during the last seven days. Since physical activities generally follow a yearly seasonal pattern (Cepeda et al., 2018), we adjusted, where appropriate, for this seasonal component using a sine/cosine regression with a yearly period. This seasonal regression model is often used in time series analysis and allows for an accurate description ($R^2 = 0.99$) of the yearly PA mean pattern of old-elderly adults (aged ≥75 years) reported in **Figure 1** of Cepeda et al. (2018). The data used to build and validate the seasonal regression model was extracted from **Figure 1** in Cepeda et al. (2018) using the open source software Engauge Digitizer (version 10.7 by Mark Mitchell).

Statistics

The number of predictor variables (**Table 2**) was relatively large in comparison to the sample size and collinearities between predictor variables would cause variance inflations in the estimators of a classical regression analysis. Therefore, we used a principal component regression (PCR) analysis to reduce the dimension of the predictor space, to gain orthogonalized predictors and to link the identified principal components (PC) with the gait stability responses (primary outcomes) as well as the gait variability responses (secondary outcomes, **Table 4**). The number of factors was determined using the Kaiser criterion (percentage of variance explained: 75.3%).

To facilitate the interpretation, we used VARIMAX-rotated PCs and considered their largest (≥0.5) standardized factor loadings (**Table 3**). Gender and muscular fitness are included in the same factor (PC three). To get further insights into the relationship of gender and muscular fitness with the response variables, we analyzed the correlations (Pearson's r) between each of the muscular fitness outcomes ($S2S_{PeakV}$ and $S2S_{MeanV}$) and each of the gait stability and gait variability outcomes separately for male and female participants (**Table 5**). Additionally, we tested for gender differences in the gait stability and gait variability outcomes using *t*-tests for independent samples (**Table 6**).

RESULTS

For 90 participants, all outcome measures were analyzed. Test data of 12 participants could not be included due to technical problems of the inertial sensor system ($n = 5$) or some participants did not want to answer all questionnaires ($n = 7$). Since only 9 participants self-reported having diabetes, this factor was not included in the statistical analysis. Of the 90 participants analyzed, 41 (46%) reported having at least one fall and 19 (21%) having two falls within the last 12 months.

Pronounced seasonal patterns were found and corrected for the overall PA [$F_{(2,99)} = 8.62$, $p < 0.001$], the basic PA [$F_{(2,99)} = 13.85$, $p < 0.001$], and the sports-related PA [$F_{(2,99)} = 3.55$, $p = 0.03$]. No significant seasonal pattern was found for the extracurricular PA.

The results of the principal components analyses are displayed in **Table 3**. Out of the 31 outcome measures, 10 factors were extracted in total: (1) "pain," (2) "balance," (3) "strength and gender," (4) "physical activity," (5) "cognition," (6) "proprioception," (7) "mental health," (8) "Osteoarthritis and prosthesis," (9) "BMI," and (10) "peripheral sensation."

The results of the regression analysis are depicted in **Table 4**. The factor "strength and gender" is a significant predictor for LDS foot ($LDS_{foot, ST}$: $p = 0.001$, $LDS_{foot, DT}$: $p = 0.050$) and trunk ($LDS_{trunk, ST}$: $p = 0.001$, $LDS_{trunk, DT}$: $p < 0.001$). A higher relative muscular fitness and/or being male correlated with a lower largest Lyapunov exponent (better LDS). Furthermore, more physical activity improved LDS of the foot during single ($p = 0.006$) and dual-task walking ($p = 0.040$) as well as to LDS of the trunk during single task walking ($p = 0.007$). A low peripheral sensation diminished the foot LDS ($LDS_{foot, ST}$: $p = 0.002$, $LDS_{foot, DT}$: $p < 0.001$) but there was not even a trend regarding the LDS of the trunk. BMI ($LDS_{foot, ST}$: $p = 0.043$), pain ($LDS_{foot, DT}$: $p = 0.030$) and the factor "Osteoarthritis and prosthesis" ($LDS_{trunk, DT}$: $p = 0.006$) were each a significant predictor but only in one model. Good balance abilities ($p = 0.094$) and a lower BMI ($p = 0.075$) tended to improve $LDS_{foot, DT}$.

Within the secondary analysis, Stride-to-stride gait variability was analyzed. Again, "strength and gender" ($SD_{StrideTime, ST}$: $p = 0.039$) and "physical activity" ($SD_{StrideLength, DT}$: $p = 0.014$, $SD_{StrideTime, ST}$: $p = 0.011$, $SD_{StrideTime, DT}$: $p = 0.003$,) were significant predictors. "Balance" ($SD_{StrideLength, ST}$: $p = 0.020$), "pain" ($SD_{StrideTime, ST}$: $p = 0.019$), "cognition" ($SD_{StrideLength, DT}$: $p = 0.007$) and "peripheral sensation" ($SD_{StrideTime, DT}$: $p = 0.039$) were only in one model of the secondary analysis significant predictors. Good "proprioception" or less "pain" tended to improve $SD_{StrideTime, ST}$ ($p = 0.077$) and $SD_{StrideTime, DT}$ ($p = 0.058$), respectively.

Knowing, that muscular fitness and gender were included into one factor, we determined the relation of muscular fitness (outcomes $S2S_{PeakV}$ and $S2S_{MwV}$) with the gait stability and gait variability outcomes for men and women, separately (**Table 5**). For men, a higher muscular fitness improveed $LDS_{foot, ST}$ ($S2S_{PeakV}$: $r = -0.33$, $p = 0.010$), $LDS_{foot, DT}$ ($S2S_{PeakV}$: $r = -0.30$, $p = 0.019$) and $SD_{StrideLength, ST}$ ($S2S_{PeakV}$: $r = -0.24$, $p = 0.049$; $S2S_{MeanV}$: $r = -0.25$, $p = 0.043$). Furthermore,

TABLE 3 | Results of the principal components analysis.

		Principle components									
		1 Pain	**2 Balance**	**3 Strength &Gender**	**4 Physical activity**	**5 Cognition**	**6 Proprioception**	**7 Mental health**	**8 Osteoarthritis/ prosthesis**	**9 BMI**	**10 Periperal sensation**
	Gender			−0.79							
	Age										0.51
	BMI									0.77	
	Osteoarthritis								0.70		
	Prosthesis								0.68		
	$S2S_{PeakV}$			0.91							
	$S2S_{MwV}$			0.94							
Balance	$Sway_{dl,eo}$		0.84								
	$Sway_{st,eo}$		0.90								
	$Sway_{dl,ec}$		0.90								
	$Sway_{st,ec}$		0.87								
	$Pain_{neuro}$ (Pain Detect)	0.64									
Pain SES	$Pain_{affect}$	0.85									
	$Pain_{sens}$	0.97									
	$Pain_{sens,rhythm.}$	0.65									
	$Pain_{sens,depth}$	0.82									
	$Pai_{sens,temp.}$	0.75									
Stroop	Cog_{ink}					0.97					
	Cog_{relink}					0.95					
	FES-I										
	CES-D							0.78			
SF12	$SF12_{physical}$									−0.62	
	$SF12_{mental}$						−0.88				
QST	$Sens_{VibDT}$										−0.66
	$Sens_{MechDT}$										0.69
Prop.	$Prop_{MeanErr}$						0.89				
	$Prop_{SDErr}$						0.88				
FFKA	$FFkA_{basic}$				0.73						
	$FFkA_{Extracurr}$				0.60			0.54			
	$FFkA_{Sports}$				0.58						
	$FFkA_{total}$				0.95						

*The principal components analysis was used to reduce the dimension of the predictor space (**Table 2**) for a subsequent regression analysis (**Table 4**). To facilitate the interpretation, we used VARIMAX-rotated PCs and considered their largest (≥0.5) standardized factor loadings. Variable abbreviations are outlined in **Table 2**.*

$S2S_{PeakV}$ ($LDS_{trunk,\ DT}$: $r = -0.21$, $p = 0.077$) and $S2S_{MeanV}$ ($LDS_{foot,\ ST}$: $r = -0.23$, $p = 0.052$; $LDS_{foot,\ DT}$: $r = -0.19$, $p = 0.092$) tended to increase LDS. For women, $SD_{StrideTime,\ ST}$ was correlated with $S2S_{PeakV}$ ($r = -0.36$, $p = 0.005$) and $S2S_{MeanV}$ ($r = -0.40$, $p = 0.002$). $LDS_{trunk,\ DT}$ ($S2S_{MeanV}$: $r = -0.23$, $p = 0.059$) tended to be correlated with muscular fitness measures.

Gait stability and gait variability measures of male vs. female participants are given in **Table 6**. Men depicted better LDS of foot ($LDS_{foot,\ ST}$: $p = 0.017$, $d = 0.48$; $LDS_{foot,\ DT}$: $p = 0.029$, $d = 0.45$) and trunk ($LDS_{trunk,\ ST}$: $p < 0.001$, $d = 0.82$; $LDS_{trunk,\ DT}$: $p < 0.001$, $d = 0.99$) but higher gait variability ($SD_{StrideLength,\ ST}$: $p < 0.001$, $d = 0.85$) than women.

DISCUSSION

The aim of the cross-sectional study was to explore influencing intrinsic factors on local dynamic gait stability and gait variability in an older population. The four dimensions (factors of a principal component analysis) (1) "strength and gender," (2) "physical activity," (3) "pain," and (4) "peripheral sensation" were each associated with at least two of the analyzed gait stability/variability measures. Dimension (5) "balance" was a significant predictor in only one gait measure. While dimension (6) "proprioception" tends to correlate with a gait variability measure, we did not find a dependency of mental health on any gait

TABLE 4 | Regression analyses were used to link the identified principal components (**Table 3**) with the gait stability responses (primary outcomes) and the gait variability responses (secondary outcomes).

	Primary analysis								**Secondary analysis**							
	LDS foot				**LDS trunk**				**SD Stride length**				**SD stride time**			
	ST		**DT**		**ST**		**DT**		**ST**		**DT**		**ST**		**DT**	
	beta	(*p*)	beta	(*p*)	beta	(*p*)	beta	(*p*)	beta	(*p*)	beta	(*p*)	beta	(*p*)	beta	(*p*)
Pain	0.155	(0.101)	0.214	(0.030)	−0.054	(0.593)	0.073	(0.462)	−0.033	(0.757)	−0.040	(0.700)	0.239	(0.019)	0.203	(0.058)
Balance	0.099	(0.297)	0.165	(0.094)	−0.162	(0.113)	−0.062	(0.532)	0.254	(0.020)	−0.016	(0.880	0.095	(0.348)	−0.004	(0.973)
Strength & Gender	−0.318	(0.001)	−0.188	(0.050)	−0.331	(0.001)	−0.432	(<0.001)	0.101	(0.338)	0.111	(0.277)	−0.207	(0.039)	−0.107	(0.305)
Physical activity	−0.260	(0.006)	−0.197	(0.040)	−0.272	(0.007)	−0.103	(0.287)	−0.124	(0.237)	−0.256	(0.014)	−0.255	(0.011)	−0.318	(0.003)
Cognition	−0.049	(0.597)	−0.116	(0.226)	0.025	(0.801)	−0.012	(0.902)	0.112	(0.286)	0.280	(0.007)	−0.047	(0.635)	−0.022	(0.833)
Proprioception	−0.014	(0.882)	−0.141	(0.144)	−0.005	(0.962)	−0.006	(0.950)	0.148	(0.162)	−0.022	(0.831)	0.177	(0.077)	−0.019	(0.855)
Mental health	0.105	(0.257)	−0.063	(0.518)	0.123	(0.216)	−0.113	(0.255)	−0.087	(0.410)	0.146	(0.155)	0.117	(0.240)	0.095	(0.370)
Osteoarthritis/prosthesis	−0.023	(0.800)	−0.032	(0.734)	0.135	(0.175)	0.273	(0.006)	0.064	(0.540)	−0.163	(0.111)	0.007	(0.941)	−0.098	(0.343)
BMI	0.191	(0.043)	0.172	(0.075)	0.098	(0.327)	−0.065	(0.501)	0.062	(0.557)	−0.112	(0.275)	0.137	(0.173)	−0.009	(0.933)
Periperal sensation	0.293	(0.002)	0.430	(<0.001)	−0.001	(0.995)	0.150	(0.123)	0.032	(0.763)	0.034	(0.740)	−0.030	(0.764)	0.216	(0.039)
Adjusted R^2	0.255		0.260		0.150		0.233		0.042		0.123		0.148		0.116	

ST, Single-task walking; DT, Dual-task walking; dark gray, $p < 0.05$ indicating a significant effect; light gray, $0.05 \leq p < 0.01$ indicating a non-significant tendency.

TABLE 5 | The correlations (Pearson's r) between each of the muscular fitness outcomes ($S2S_{PeakV}$ and $S2S_{MeanV}$) and each of the gait stability and gait variability outcomes were separately assessed for male and female participants.

		Men (*n* = 49)		**Women (*n* = 46)**	
		$S2S_{PeakV}$ *r* (*p*)	**$S2S_{MwV}$ *r* (*p*)**	**$S2S_{PeakV}$ *r* (*p*)**	**$S2S_{MwV}$ *r* (*p*)**
LDS foot	ST	−0.327 (0.010)	−0.232 (0.052)	−0.096 (0.256)	−0.185 (0.102)
	DT	−0.296 (0.019)	−0.193 (0.092)	0.013 (0.465)	−0.056 (0.356)
LDS trunk	ST	0.017 (0.455)	0.126 (0.192)	−0.094 (0.259)	−0.177 (0.111)
	DT	−0.207 (0.077)	−0.128 (0.191)	−0.129 (0.196)	−0.234 (0.059)
SD Stride length	ST	−0.237 (0.049)	−0.246 (0.043)	−0.179 (0.110)	−0.191 (0.094)
	DT	0.070 (0.314)	0.034 (0.408)	0.043 (0.386)	0.039 (0.397)
SD Stride time	ST	−0.099 (0.246)	−0.130 (0.183)	−0.363 (0.005)	−0.397 (0.002)
	DT	−0.109 (0.227)	−0.118 (0.209)	−0.201 (0.091)	−0.219 (0.072)

ST, Single-task walking; DT, Dual-task walking. dark gray, $p < 0.05$ indicating a significant effect; light gray, $0.05 \leq p < 0.01$ indicating a non-significant tendency.

measure. Hereafter, we will discuss these dimensions one after another:

1) The results suggest that participants with higher relative muscle performance or men walk more stable. Since both muscular fitness (sit-to-stand test) and gender were merged into one factor, we analyzed their individual contributions on gait stability and gait variability, separately. Comparing male and female participants, we observed significant differences in gait stability. Regarding the stability measures of the primary analysis, men walk more stable than women. This could be a reason why women are more likely to fall (WHO, 2008; Robinovitch et al., 2013). Despite this, the correlation of relative muscle performance with these primary measures was, on the one hand, stronger in men than in women regarding the gait stability measures but on the other hand, more pronounced in women as compared to men regarding the gait variability measures. Overall, the strongest correlations depicted only medium effects. The gait measures used in the current study reflect the system's capacity to recover from small perturbations (i.e., neuromuscular noise and wind, Bruijn et al., 2013). However, other types of stability, such as the recovery from larger perturbations (e.g., after tripping), might require more strength. This would also explain the comparatively low effect sizes of the correlation analysis. This result suggests that lower extremity muscular fitness is less relevant for LDS but could be more relevant for other kinds of gait stability (e.g., recovery from larger perturbations). This is in line with the recommendation to include strength exercises into fall prevention programs (Sherrington et al., 2011).
2) The amount of physical activity is also a strong predictor of most gait stability and gait variability measures during both, single-task and dual-task walking. This finding is in agreement with the broad evidence for beneficial effects of regular physical activity for enhancing and maintaining older

TABLE 6 | We tested for gender differences in the gait stability and gait variability outcomes using *t*-tests for independent samples (ST, single-task walking; DT, Dual-task walking).

				Male		Female		Male vs. female			
				Mean	**SD**	**Mean**	**SD**	***t***	**df**	***p***	***d***
Primary analysis	LDS foot		ST	1.58	0.15	1.66	0.17	−2.42	100	0.017	−0.48
			DT	1.72	0.19	1.81	0.22	−2.22	96	0.029	−0.45
	LDS trunk		ST	0.77	0.12	0.89	0.16	−4.12	100	<0.001	−0.82
			DT	0.83	0.13	0.98	0.17	−4.91	96	<0.001	−0.99
Secondary analysis	SD Stride length	[mm]	ST	29	7	24	5	4.30	100	<0.001	0.85
		[mm]	DT	31	9	29	8	1.32	98	0.188	0.26
	SD stride time	[ms]	ST	17	7	17	5	0.04	100	0.967	0.01
		[ms]	DT	25	11	28	18	−0.76	96	0.447	−0.15

adults' fitness as well as mental and physical health-related quality of life (Taylor et al., 2004; Netz et al., 2005; Nelson et al., 2007).

3) In our study, pain predicted a few gait outcomes. This is in line with previously reported data (Hamacher et al., 2016). In the paper of Hamacher et al. dual-task costs of gait depended on pain severity. In fact, pain is known to affect muscle activity and biomechanical behavior (Hodges, 2011) and pain disrupt cognitive functions and executive control (Keogh et al., 2013). Thus, it is surprising that the effects of pain on gait stability or gait variability were not more pronounced. A reason could be that the pain questionnaires were not assessed during the test day of the gait analysis which could have reduced the measured effect of pain on gait stability or variability measures.
4) Peripheral sensation was a significant predictor of foot LDS. In another study, touch and vibration sense were correlated with static balance performance (Lord et al., 1991) confirming our results. Interestingly, peripheral sensation did primarily effect foot LDS but not trunk LDS. In further studies, phase-dependent local dynamic stability (Ihlen et al., 2015) should be used to reveal if such effects are restricted to the stance or swing phase of gait.
5) We did not reveal any significant effect of balance in the primary analysis and only one within the secondary analysis. This is surprising since, in most studies, balance deficits being discussed to be a relevant risk factor for falls (Lord et al., 2003; Ambrose et al., 2013; Pfortmueller et al., 2014) and exercises to improve balance have been suggested to be included into fall prevention programs (Sherrington et al., 2008, 2011). Our results could imply that (static) balance is not that important for gait as it is for preventing falls in general (e.g., during quiet standing). It is known that posture control concepts are fundamentally different for standing and walking (Winter, 1995). Furthermore, it is known, that balance abilities are context-specific (Sibley et al., 2015; Kümmel et al., 2016). Our result could also be a limitation of the methods chosen. We only assessed static balance measures but no dynamic balance parameters and it is known that the predictive value for falls depends on the method (Muir et al., 2010) and that there is a high intrasubject variability in individual concepts maintaining postural stability (Pasma et al., 2014).
6) We did not find a significant dependency of the active joint position sense on gait stability or gait variability. This is in contrast with the result of Lord et al. (1991), who reported a relation of proprioception and static and dynamic balance measures. Non-questionable, proprioception plays a key role in motor control and functional joint stability (Riemann and Lephart, 2002; Proske and Gandevia, 2012) but the quality criteria (reliability and validity) have been questioned, in general (Riemann et al., 2002; Benjaminse et al., 2009; Hillier et al., 2015). Low quality criteria could explain the missing relationship between proprioception and gait stability or gait variability.

Overall, the explained variance of the regression model is rather low (adjusted R^2 range from 0.04 to 0.26). In models predicting static or dynamic balance, the multiple R ranged from 0.24 to 0.37 (Lord et al., 1991) which would be an R^2 of 0.06 to 0.14. These R^2 values are comparable to ours (between 0.15 and 0.26 for the models predicting LDS of the foot and trunk, **Table 4**). However, the low explained variance highlights the fact that there could be more relevant factors for gait stability and gait variability (as well as for balance) that were not addressed, yet. However, a strength of our study is the large number of considered parameters. Furthermore, the study sample seems to be sufficient to reveal effects relevant to the practice. On the other hand, this is a cross-sectional study. Thus, the results should be confirmed with experimental designs, for example with intervention studies designed to improve gait stability. Additionally, we did not assess vision, vestibular functioning, peripheral nerve tests and reflexes or Vitamin D deficiency and other variables that also have been discussed to be relevant risk factors for falls (Tinetti et al., 1988; Lord et al., 2003; Ambrose et al., 2013; Pfortmueller et al., 2014). At last, we assessed the participants' ability to recover from small perturbations. There are other types of gait stability that should be addressed in further studies.

In conclusion, the participants' ability to recover from small perturbations (as measured with the largest Lyapunov exponent) seems to be related to (1) gender and muscular fitness, (2) the amount of physical activity the participants spent every week, (3) peripheral sensation (mechanical and vibration detection threshold), and (4) pain status. No or minor effects were found for balance, proprioception, cognition or mental health. Since the explained variance is still rather low, there could be more relevant factors that were not addressed, yet.

AUTHOR CONTRIBUTIONS

DH and AZ planned and designed the study. DH, CH, VH, CK, and TT collected the data. DL and DH analyzed the data. DH, DL, and AZ drafted the manuscript. All authors critically revised the manuscript. All authors approved the final version of the manuscript.

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Validity and Reliability of a Novel Integrative Motor Performance Testing Course for Seniors: The "Agility Challenge for the Elderly (ACE)"

Eric Lichtenstein[1], Oliver Faude[1], Aline Zubler[1], Ralf Roth[1], Lukas Zahner[1], Roland Rössler[2], Timo Hinrichs[1], Jaap H. van Dieën[3] and Lars Donath[4]**

[1] Department of Sport, Exercise and Health, University of Basel, Basel, Switzerland, [2] Department of Public and Occupational Health, Amsterdam Public Health Research Institute, Amsterdam UMC, Vrije Universiteit Amsterdam, Amsterdam, Netherlands, [3] Department of Human Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam Movement Sciences, Amsterdam, Netherlands, [4] Department of Intervention Research in Exercise Training, Institute of Exercise Training and Computer Science in Sport, German Sport University Cologne, Cologne, Germany

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***Correspondence:**
Eric Lichtenstein
e.lichtenstein@unibas.ch
Lars Donath
l.donath@dshs-koeln.de

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Background: Assessing traditional neuromuscular fall risk factors (i.e., balance, gait, strength) in the elderly has so far mainly been done independently. Functional and integrative testing approaches are scarce. The present study proposes an agility course for an integrative assessment of neuromuscular and also cardiocirculatory capacity in seniors – and tests its criterion validity and reliability.

Methods: Thirty-six seniors (age: 69.0 ± 2.8 years; BMI: 25.4 ± 3.5 kg/m^2; sex: 19 males/17 females; weekly physical activity: 9.4 ± 5.5 h) participated. They completed four trials of the Agility Challenge for the Elderly (ACE)-course in two sessions separated by 1 week. The course consists of three segments focusing on different agility aspects. Cardiovascular capacity was assessed by 6-min walk test (6MWT), neuromuscular capacity by static, dynamic and perturbed standing balance tasks, habitual gait speed assessment, and rate of torque development testing. Parameters' predictive strength for the ACE performance was assessed by regression analysis. Reliability was calculated with intraclass correlation coefficients and coefficients of variation.

Results: Men completed the course in 43.0 ± 5.7 s and women in 51.9 ± 4.0 s. Overall and split times were explained by 6MWT time (η_p^2 = 0.38–0.44), gait speed (η_p^2 = 0.29–0.43), and to a lesser extent trunk rotation explosive strength (η_p^2 = 0.05–0.12). Static and dynamic balance as well as plantar flexion strength explained the performance in some segments to a very small extent (η_p^2 = 0.06–0.08). Good between- and within-day reliability were observed for total course and split times: The ICC for the between-day comparison was 0.93 (0.88–0.96) and ranged between 0.84 and 0.94 for split times. The within-day ICC was 0.94 (0.91–0.97) for overall time and 0.92–0.97 for split times. Coefficients of variation were smaller than 5.7% for within and between day analyses.

Conclusion: The ACE course reflects cardiocirculatory and neuromuscular capacity, with the three ACE segments potentially reflecting slightly different domains of neuromuscular (static and dynamic balance, ankle, and trunk strength) performance, whereas cardiocirculatory fitness and gait speed contribute to all segments. Our test can detect changes in overall performance greater than 5.7% and can thus be useful for documenting changes due to interventions in seniors.

Keywords: standing balance, elderly, seniors, exercise testing, risk of falling, balance, gait, strength

INTRODUCTION

Approximately 30% of the population in western societies will be aged >65 years until the end of the 21st century (Lutz et al., 2008). One-third of these seniors falls once a year and half of those people fall again within the subsequent year (Rubenstein, 2006; Lord, 2007). Falls are the leading cause of hospitalizations due to injury in this age-group (Jones et al., 2011). The resulting expenditures for the health care system are substantial (Bohl et al., 2010). In addition to extrinsic factors (e.g., poor lighting, bumps, ice, footwear) intrinsic factors, such as declines of lower limb strength (maximal and explosive strength) (Doherty, 2003) and impaired balance and gait performance (under single and dual task conditions) (Hytonen et al., 1993; Granacher et al., 2011a,b) contribute to increased individual fall risk.

These intrinsic fall risk factors have mostly been assessed independently (Miyamoto, 2008; Avelar et al., 2016; Donath et al., 2016a). Available evidence suggests that a lack of effectively integrating neuromuscular and cognitive function during difficult tasks might be an underlying reason for falls in seniors (Beauchet et al., 2009). The limitations of independently assessing different fall risk factors might be overcome if accelerations, decelerations, stop and go patterns, change in directions, eccentric and rotational movements and demanding spatial orientation tasks are integrated into a testing protocol. The need for such an integrative multicomponent testing approach, combining cognitive and motor inferences in functional tasks seems justified.

Recently, Donath et al. (2016b) proposed an own and novel agility-based framework for that purpose, potentially serving as a time-efficient and appealing method to assess the interplay and combination of several neuromuscular and cognitive fall risk factors. According to the authors agility comprises accelerations, decelerations, stop-and-go patterns, changes of direction, and eccentric loads, combined with demanding spatial orientation tasks. In line with this conceptual model, our "Agility Challenge for the Elderly" (ACE) attempts to integratively assess the different demands that are posed by fall-threatening real-life challenges.

The present study investigated whether and to which extent traditional neuromuscular (static and dynamic balance, gait, strength) and cardiocirculatory (aerobic endurance) performance variables relate to overall time and split time of this novel agility testing course in community-dwelling seniors. We hypothesize that different neuromuscular and cardiocirculatory capacity variables reflect different domains (i.e., stop-and-go, cutting manoeuvers, spatial orientation) of the ACE-course.

Exercise-based fall prevention studies require reliable detection of acute and interventional changes of neuromuscular performance or cardiovascular capacity. Large day-to-day variability of a neuromuscular fall risk factor due to biological- or device immanent "errors" can impede reliable detection of changes (Atkinson and Nevill, 1998). Absolute and relative reliability indices have been described for a variety of traditional balance- and strength-based fall risk factors (Muehlbauer et al., 2011; Roth et al., 2016). The present study, therefore, also assessed absolute and relative within-day and between-day reliability indices of our novel Agility-course.

MATERIALS AND METHODS

Study Design

The present study was conducted as a cross-sectional trial with a repeated measures design. Participants were tested on 3 days, 2–7 days apart. The first day was lab based applying several strength and balance tests. Lab testing took place in the following order: static balance, perturbed balance, dynamic balance, lower limb and trunk explosive strength, and lastly habitual gait speed assessment. Prior to these tests, anthropometrical data (BMI, leg length, and leg preference) were collected. Leg preference was determined by four established questions on leg dominance (Coren, 1993). The two subsequent testing days were conducted in a gym and comprised the ACE and the 6-min walk test (6MWT). The physical activity readiness questionnaire (PAR-Q) was used to determine participants' eligibility for test participation. Physical activity patterns were recorded utilizing the "Freiburg physical activity questionnaire" (Frey et al., 1999). The study was approved by the local ethics committee (Ethics Committee of Northwestern and Central Switzerland; approval number: 740/2016) and complied with the Declaration of Helsinki. All participants signed an informed written consent prior to the start of the study after receiving all relevant study information.

Participants

Healthy seniors, aged between 65 and 75 years, without clinical conditions were enrolled in the present study (**Table 1**). Participants could not suffer from chronic diseases, musculoskeletal impairments or cardio-vascular conditions that could affected testing. All participants were asked to refrain

TABLE 1 | Senior's anthropometric data, physical activity, and endurance capacity.

	All (36)		Men (19)		Women (17)	
	Mean	SD	Mean	SD	Mean	SD
Age (y)	69.0	2.8	69.0	2.6	68.9	3.1
BMI (kg/m^2)	25.4	3.5	25.2	2.3	25.7	4.3
PA/week (h)	9.4	5.5	9.3	4.6	9.4	6.4
sPA/week (h)	4.7	3.8	5.1	4.1	4.2	3.3
6MWT (m)	639	72	678	61	596	59

PA, physical activity; sPA, sportive physical activity.

from severe exercise within the last 48 h prior to exercise testing.

Testing Procedures and Data Processing

Balance and Gait Speed Testing

Static balance was assessed on a Kistler® force platform (KIS, Type 9286BA, Winterthur, Switzerland). Data collection lasted 30 s and three trials interspersed with 1 min of rest were conducted. All participants stood barefoot on their dominant leg with eyes open and were instructed to (a) remain as stable as possible, (b) focus on a marker on the wall (distance: 1.5 m; height: 1.75 m), (c) place the hands on the iliac crests (akimbo). Static balance performance was operationalized using the path length displacement of the center of pressure (CoP). Data were recorded at 120 Hz. Good reliability has been reported by Markovic et al. (2014) for static balance measurements under the mentioned conditions (ICC = 0.92–0.98).

The ability to deal with external perturbations was tested on the Posturomed® (Haider Bioswing, Pullenreuth, Germany). This tool consists of a movable platform attached to a solid frame with two dampened pendulums on each corner allowing the platform to move in all horizontal directions. The platform was initially locked in a stable position 2.5 cm away from its neutral position. Once the participants were stable in the starting body position (same as during static testing), the lock was released. Platform release was applied unexpectedly and the participants had to reduce the oscillation of the platform as fast as possible. An accelerometer (MicroSwing® 6, Haider Bioswing, Pullenreuth, Germany) was attached to the bottom of the platform and the platform's acceleration during the first 10 s after platform release was recorded and the sway path calculated. The three trials were conducted with 1-min rest between trials and data were recorded at 50 Hz. Schmidt et al. (2015) have reported acceptable reliability when assessing perturbation with this device and a similar protocol (ICC = 0.71–0.94).

Dynamic balance performance was tested using the Y-balance test (Functional Movement Systems, Chatham, MA, United States). This test comprises a Y-shaped plastic device where participants are instructed to push a plastic box as far as possible with one foot in anterior, posterior-medial, and posterior-lateral direction, respectively, while maintaining balance on the standing leg. Two familiarization trials were conducted for both legs and each direction. The participants were instructed to (a) place the hands on the hips, (b) only touch the box on the vertical surface, and (c) not kick the box. The distance between the furthest reaching positions of the box from the center was recorded and a composite score was calculated adjusted for leg length measured by the distance from the ground to the pubic bone during upright stance. Thus, we applied our own leg length measuring procedure instead of using anatomical landmarks. The composite score is the sum of the three reach distances divided by three times the leg length multiplied by 100 to obtain a percentage (Lai et al., 2017). The average composite score of three trials and both legs was used in the analysis. High reliability has been reported for the Y-balance test (ICC = 0.85–0.93) (Shaffer et al., 2013).

Habitual gait speed was assessed by instructing participants to walk in a 10 m corridor at their usual pace while time was measured with timing gates (Witty, Microgate, Bolzano, Italy). Participants started 2 m before and finished 2 m behind the timing gates to avoid the possible influence of acceleration and deceleration.

Strength Testing

To assess leg and trunk explosive strength, participants had to perform a series of isometric tasks on an isokinetic system (Isomed 2000®, D. & R. Ferstl GmbH, Hemau, Germany). Each test was preceded by one familiarization attempt. To obtain the maximum rate of torque development (RTD) participants were instructed to isometrically push as fast and hard as possible (Maffiuletti et al., 2016). Three trials were conducted for each movement.

During plantar flexion (PF) and dorsal extension (DE) testing, participants were positioned in a supine posture with hip and knee angles in a neutral position (0°) and the ankle angle at 10° plantar flexion. The working leg and feet were strapped to the device. Only plantar flexion and dorsal extension were possible in this position. Participants were instructed to cross their arms in front of their chest. Every leg and direction was tested, starting with the dominant leg.

To measure trunk extension and flexion, respectively, participants were placed on the Isomed® trunk adapter with a hip angle of 85° and a knee angle of 45°. They were fixed at the chest, knees and hip and pulled with their hands on a handle nearby their clavicular bone. Trunk extension was tested first, followed by flexion. Third, participants had to sit in the trunk rotation (TR) adapter with hip and knees at 90°. Their legs and pelvis were fixed with adjustable pads and they had to push with their shoulders against a pad in the left and right direction. The hands were placed loosely on their lap. Maximum RTD was calculated from the raw force data as the maximum rise of torque over 200 ms during every trial reflecting suggested time windows for RTD assessment (0–250 ms) but avoiding problems of force onset detection (Maffiuletti et al., 2016). The Isomed 2000® samples data at 200 Hz and filters the signal with a 6th order Butterworth filter with a cut off frequency of 200 Hz.

Endurance Testing

The 6MWT was used to measure endurance capacity (Bautmans et al., 2004). Seniors were instructed to briskly walk as far as possible during a 6-min period without running. Two cones were placed 30 m apart and participants shuttled between the cones. A marker was placed every 3 m and the participants had to stop at the nearest marker upon the stop signal. For logistical reasons, several participants (up to six) performed the test simultaneously, starting at 30-s intervals. The 6MWT was conducted after the first day of agility testing with at least 10 min of rest between the last agility test and the 6MWT.

Agility Testing (ACE)

All participants underwent a standardized 5-min warm up procedure prior to the ACE course testing. This warm-up phase consisted of slow and brisk walking, side stepping, knee lifting, backward walking and some hip rotations. The ACE is a course developed for a standard 9 m × 18 m volleyball court (**Figure 1**). The ACE course includes three segments. Each of the three segments aims at testing a certain aspect of agility. Participants completed the ACE four times interspersed with at least 3 min of rest. While not performing the tests, they were placed behind a wall to avoid observing other participants going through the course. The course was demonstrated twice by a study assistant. The first attempt of each testing day served as familiarization trial. Participants were instructed to walk as fast as possible without running. The fastest times for every segment from the second day were used for the validity analysis. Additionally, the number of trials where participants did not execute the tasks in the instructed order was recorded. Errors included a wrong sequence in the last segment, omission of the last cone, additional rounds around it or not following the indicated path at the start of the second segment. These trials were excluded from the analysis.

Segment A (**Figure 1A**) focuses on acceleration and deceleration by stop-and-go movement. This part covers 10 m with markers at 3, 4, 6.5, 8.5 m on the ground (10 cm × 50 cm). All participants were instructed to touch those markers with both feet simultaneously before continuing to the next marker. This segment starts in one corner of the back zone and finishes at the opposite corner of the back zone, where a cone is placed. Timing gates (TG) (Witty, Microgate, Bolzano, Italy) were placed at the start, perpendicular to the walking direction and at the 10-m mark. Seniors had to round the cone and then continue onward to the second segment. Segment B (**Figure 1B**) focuses on changes of direction, employing cutting maneuvers. Following the sideline, participants had to place their left foot in an area 50 cm × 50 cm before turning 45° and continue to the next foot zone. Six of these turns had to be completed with alternating foot placement while crossing the court. After the last turn, participants again walked along the sideline where a TG was placed on the center line ending the second segment. Segment C (**Figure 1C**) challenges spatial orientation. After following the sideline for 6 m, participants had to round a cone and were faced with three cones carrying the numbers 1, 2, and 3. The participants had to round the cones in the order from 1 to 3 and return to the base cone after rounding each cone. After rounding the last cone, participants had to return to the base cone one more time, round it and walk along the sideline through the final TG. The numbers on the cones were individually randomized for each trial.

Statistical Analysis

Data are provided as means with standard deviations (SD) as well as 95% confidence intervals (CI) for men and women separately. To explore the ACE's congruent validity with known physical capacity measures indicative of fall risk, multivariate linear regression models were constructed with the duration of every segment and the overall duration of the ACE as dependent variables. Firstly, gender and one predictor were used in the model to assess this latter predictor's strength. Then, all predictors were included and backward stepwise selection was done to determine the best model fit. The weakest models were discarded based on Akaike's information criterion (AIC) until the strongest model remained (Akaike, 1973). The chosen predictors in the starting model included all balance, explosive strength

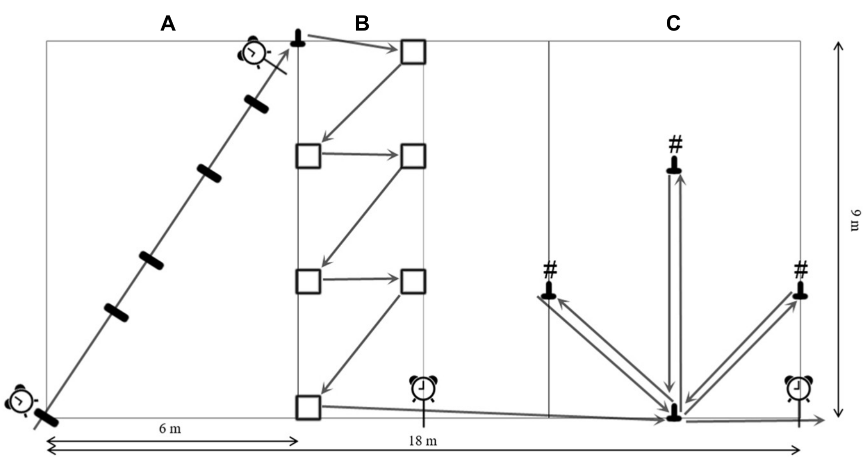

FIGURE 1 | The ACE-course layout with positioning of timing gates and cones with numbers (#), **(A)** stop and go segment; **(B)** change of direction segment; **(C)** spatial orientation segment.

and endurance parameters as well as the participant's gender. Every model's parameter's estimate (β), *p*-value (p), partial eta squared (η_p^2), and model strength [adjusted R-squared (R^2)] were calculated. In this context, η_p^2 serves as the magnitude of explained variance by the predictor excluding all other predictors. According to Cohen (1988) η_p^2 is interpreted as small when $0.01<\eta_p^2<0.06$, medium when $0.06<\eta_p^2<0.14$, and large when $\eta_p^2>0.14$. This procedure is similar to the initial analysis but controlled for the other potentially influencing predictors as well. Collinearity was assessed by calculating the variance inflation factors and normal distribution was checked with the Shapiro–Wilk test. The Software R (Version 3.5.1) was used to conduct the calculations utilizing the packages "car" (Version 3.0-2), "lmsupport" (Version 2.9.13), and "MASS" (Version 7.3-50).

Within-day and between-day reliability indices were calculated using a published spreadsheet and the typical error (TE), coefficient of variation (CV), and intraclass correlation coefficients (ICC, type 3,1) are reported with 95% CIs (Hopkins, 2015). The minimum detectable change was calculated as TE*1.96*2ˆ1/2 (Beaton, 2000).

RESULTS

Subject Characteristics

Thirty-six healthy seniors (17 women, 19 men) were recruited and completed the assessments. Their characteristics are summarized in **Table 1**.

Overall Agility Performance

Men's mean overall time was 43.1 s (5.7) and women's mean time was 51.9 s (4.0). Single predictor regression analysis revealed a small to moderate effect on overall performance for Y-balance composite score ($\eta_p^2 = 0.07$; $p = 0.13$), plantar flexion RTD ($\eta_p^2 = 0.07$; $p = 0.12$), and trunk rotation RTD ($\eta_p^2 = 0.08$; $p = 0.10$) albeit not statistically significant. A large effect was observed for 6MWT distance ($\eta_p^2 = 0.44$; $p < 0.001$) and self-selected speed during gait speed assessment ($\eta_p^2 = 0.43$; $p < 0.001$). Multiple predictor regression analysis revealed that the possible influence of plantar flexion ($\eta_p^2 = 0.02$; $p = 0.41$) and trunk rotational RTD ($\eta_p^2 = 0.01$; $p = 0.66$) disappeared. Y-balance composite score was discarded from this analysis because it was not part of the strongest model according to AIC. The model's strength including sex, gait speed, 6MWT distance, plantar flexion and trunk rotation RTD was $R^2 = 0.73$ (**Table 2**).

Split Times

The three different segments of the ACE were completed by men in 6.1 (0.7), 12.5 (1.8), and 24.2 (3.3) s, women took 7.5 (0.7), 15.4 (1.6), and 28.7 (2.2) s. 6MWT distance and gait speed strongly predicted all split times ($\eta_p^2 = 0.29$–0.43; $p < 0.01$). All other parameters' predictive strength did not reach statistical significance ($p > 0.05$). Trunk rotation RTD predicted all split times, but to a small extent ($\eta_p^2 = 0.05$–0.12; $p = 0.04$–0.19). Static balance performance was slightly but not statistically significantly associated with the first (stop and go) and second (cutting) segments' time ($\eta_p^2 = 0.06$–0.07; $p = 0.14$–0.16). Despite being not statistically significant but with moderate effect sizes, Y-balance composite score slightly predicted the times of the second and third segment ($\eta_p^2 = 0.06$–0.07; $p = 0.14$–0.17). The same holds true for plantar flexion explosive strength ($\eta_p^2 = 0.06$–0.08; $p = 0.1$–0.16). Additionally, the second segment time was predicted by perturbed balance ($\eta_p^2 = 0.06$; $p = 0.16$). In the multivariate analysis including all predictors, the influence of 6MWT remained for all splits, but gait speed did no longer predict stop-and-go split times ($\eta_p^2 = 0.06$; $p = 0.20$). The possible influence of the other included predictors was attenuated for all split times (**Table 2**). Model strength including sex, gait speed and 6MWT distance and the respective included factors for the split times ranged from $R^2 = 0.67$ to $R^2 = 0.71$.

TABLE 2 | Results of the multivariate analysis.

Variable	β	95% CI	*p*	η_p^2	Model R^2
ACE Overall					
Intercept	88.20	72.17; 104.23	0.00	0.79	
Sex (w)	3.40	0.30; 6.49	0.04	0.13	
6MWT (100 m)	−5.21	−7.28; −3.14	0.00	0.45	
PF RTD (kN/s)	−1.34	−2.77; 0.08	0.07	0.10	
CoP Path (cm)	−0.12	−0.27; 0.03	0.13	0.07	
					0.69
Stop and go (A)					
Intercept	10.42	7.90; 12.94	0.00		
Sex (w)	0.93	0.40; 1.46	0.00	0.29	
6MWT (100 m)	−0.49	−0.88; −0.10	0.02	0.17	
Speed (m/s)	−0.87	−2.19; 0.44	0.20	0.06	
TR RTD (kN/s)	−0.07	−0.22; 0.09	0.42	0.02	
CoP Path (m)	0.22	−0.18; 0.61	0.29	0.04	
					0.68
Cutting (B)					
Intercept	26.53	21.67; 31.40	0.00		
Sex (w)	1.36	0.24; 2.48	0.02	0.16	
6MWT (100 m)	−0.86	−1.73; 0.00	0.06	0.12	
Speed (m/s)	−4.44	−7.28; −1.60	0.00	0.24	
TR RTD (kN/s)	−0.27	−0.75; 0.20	0.27	0.04	
PF RTD (kN/s)	−0.16	−0.88; 0.55	0.66	0.01	
					0.71
Spatial orientation (C)					
Intercept	47.86	39.64; 56.08	0.00		
Sex (w)	1.98	0.09; 3.87	0.05	0.13	
6MWT (100 m)	−1.99	−3.45; −0.53	0.01	0.20	
Speed (m/s)	−5.50	−10.31; −0.70	0.03	0.15	
PF RTD (kN/s)	−0.55	−1.76; 0.66	0.38	0.03	
TR RTD (kN/s)	−0.05	−0.86; 0.76	0.91	0.00	
					0.67

β, beta-coefficient; 95% CI, 95% confidence interval; η_p^2, partial eta-squared; p, p-value; Model R^2, adjusted r-squared of the whole model; 6MWT, 6-min walk test; PF, plantar flexion; TR, trunk rotation; RTD, rate of torque development; CoP, center of pressure.

TABLE 3 | Results of the reliability analysis.

	Within day 1		Within day 2		Between day	
	ICC	**CV**	**ICC**	**CV**	**ICC**	**CV (%)**
ACE overall	0.94 (0.91; 0.97)	3.7 (3.1; 4.6)	0.98 (0.96; 0.99)	2.2 (1.9; 2.7)	0.93 (0.88; 0.96)	4.0 (3.3; 5.0)
Stop-and-go (A)	0.92 (0.86; 0.95)	4.1 (3.4; 5.1)	0.94 (0.91; 0.97)	3.6 (3.1; 4.4)	0.84 (0.74; 0.91)	5.7 (4.7; 7.2)
Cutting (B)	0.97 (0.94; 0.98)	3.3 (2.8; 4.1)	0.98 (0.96; 0.99)	2.7 (2.3; 3.2)	0.94 (0.90; 0.97)	4.3 (3.6; 5.4)
Spatial orientation (C)	0.93 (0.88; 0.96)	3.9 (3.3; 4.9)	0.96 (0.93; 0.98)	3.0 (2.5; 3.6)	0.92 (0.87; 0.96)	4.1 (3.4; 5.2)
Faulty trials	24		13			

ICC, intraclass correlation coefficient; CV, coefficient of variation.

Between- and Within-Day Reliability

Overall ACE performance was better on the second day (−1.96 s, p = 0.00) as well as all split times in segment A (−0.45 s, p = 0.00), segment B (−0.71 s, p = 0.00), and segment C (−0.84 s, p = 0.01). The ICC for the between-day comparison was 0.93 (0.89–0.96) and ranged from 0.84 to 0.92 for split times (**Table 3**). Absolute variability (CV) was 4.0% (3.3–5.0) for the between-day comparison and consistently around 5% for the within-day (between trial) comparison: stop and go: 5.7% (4.7–7.2), cutting maneuvers: 4.3% (3.6–5.4), spatial orientation: 4.1% (3.4–5.2). The standard errors of measurement (typical errors) were found to be 1.86 s (1.55–2.34) for total course time and ranged from 0.4 to 1.1 s for split times. A minimum detectable change of 5.2 s (4.3–6.5) for the overall course time, 1.1 s (0.9–1.4) for the stop and go segment, 1.7 s (1.4–2.1) for the cutting maneuvers segment and 2.9 s (2.4–3.7) for the spatial orientation segment was calculated. 24 out of 108 trials contained an error on the first day compared to 13 on the second day (**Table 3**).

DISCUSSION

To the best of our knowledge, this is the first study that explores the possibility of assessing cardiocirculatory fitness and neuromuscular fall risk parameters (surrogate parameters) by applying a time-efficient and integrative agility approach to healthy seniors. We aimed at investigating whether the novel ACE course for seniors reflects distinct domains of traditional neuromuscular and cardiocirculatory performance indices. We aimed at providing a feasible and integrative modular walking-based agility test battery that considers various aspects of motor performance relevant to daily living and fall threatening conditions, such as stop and go movements, changes in direction and spatial orientation. We found that the overall time of the ACE-course is mostly explained by cardiocirculatory fitness (walking time during the 6MWT) and gait performance (gait speed). However, detailed split time analyses revealed that performance in each of the three major domains might be predicted by different aspects of neuromuscular performance, even though the according associations were small to medium and not statistically significant. Besides gait speed and walking time during the 6MWT, the stop and go segment was potentially associated with static balance performance and, interestingly, trunk muscle performance. Moreover, the cutting maneuver segment might depend on plantar flexion explosive power, trunk rotation and different aspects of balance. The spatial orientation segment could also depend on these factors, except for static, and perturbed balance performance.

Few other performance tests focusing on agility in the elderly people have been proposed. Miyamoto (2008) introduced a test that also requires strength, balance and speed. Yet, challenge to spatial orientation, the ability to successfully perform changes of direction and stress to the cardiocirculatory system were underrepresented and their definition of agility for the elderly might omit certain challenges with fall-threatening character.

To improve agility and attenuate fall risk, Avelar et al. (2016) proposed a balance exercise circuit, separately training several aspects of agility and found beneficial effects on leg strength and power, balance and mobility. This approach tests and trains these aspects separately, which could be improved to better reflect situations where a combination of skills is required. The ACE-test has the potential to overcome this limitation and provides a blueprint for integrated assessment of all of these agility aspects without the need of an exhaustive test battery. Still, whether this agility approach can discriminate between future fallers and non-fallers remains to be elucidated in the future.

An interesting finding of our study was the association of trunk rotation explosive strength and performance in all segments of the ACE-course even though the effect sizes were small and lack statistical significance level. However, the predictors remained in the regression model and small effects can provide meaningful impact from an epidemiological perspective in the long run. Granacher et al. (2013) highlighted the importance of core strength and stability for the avoidance of falls. They noted that trunk muscle strength plays an important role in balance recovery. In line with this, Donath et al. (2013, 2015) found that slackline training reduced trunk muscle activity during highly difficult balance tasks. Additionally, it has been noted that a quick rotation of the trunk can help to avoid hip fractures by attenuating or dodging a direct shock on the hip

(Benichou and Lord, 2016). We herewith propose a method that might also reflect some functional aspects of core strength performance alongside other risk factors during integrative and functional tasks without the need for additional testing.

In order to enable a proper detection of intervention effects or discrimination between participants, the reliability of any test instrument should be documented (Atkinson and Nevill, 1998). We found the agility course to be acceptably reliable within and between days with coefficients of variations smaller than 5.7% for within and between day analyses for all segments and the overall performance. Considering that improvements of more than 5% due to a standard 6 to 12-week exercise intervention can be expected in seniors for parameters of strength and postural control (Frontera and Bigard, 2002; Granacher et al., 2009), our agility course could be suitable to detect such changes in unimpaired healthy individuals. Nevertheless, two familiarization attempts prior to testing are suggested to reduce the error rate and improve reliability. This is motivated by the fact that the within-day reliability appeared to be better on the second day, which was also supported by a drop in faulty trials (22.2–12.0%). This suggests an effect of familiarization that could mask potential intervention effects if not accounted for.

Some limitations have to be mentioned. The assessment of 6MWT was not done independently due to logistical reasons and the other participants' speed may have influenced participants performance. When constructing multivariate regression models the collinearity of several predictors can be of concern and there is an expected moderate correlation between 6MWT and habitual gait speed. Yet, the 6MWT was the most feasible assessment of cardiocirculatory capacity even though the limiting factor might have been brisk walking speed for some participants. The assessment of both parameters seemed reasonable, especially since gait speed is a strong predictor of sarcopenia in the elderly population. Additionally, the recruited population had an above average level of fitness and daily physical activity. Therefore, the ACE test might only be a reliable and suitable method for testing individuals that do not suffer from locomotor impairments. Subjects with poor physical performance and symptoms of frailty might not be suited to attempt the test. However, the ACE test is designed for a preventative setting in order to potentially detect people of risk for developing frailty or sustaining falls long before impairments are manifested. Whether the ACE test can actually estimate fall risk in the future remains to be elucidated. On the other hand, the magnitude of the presented associations might be small because of the very homogenous sample of highly active and fit individuals.

Due to the cross-sectional design of the study no causative link between the measured parameters can be established. Future studies should attempt to use a training method based on the ACE course's principal design to establish, whether it can relevantly improve fall risk factors, reduce the rate of falls or attenuate the progression of sarcopenia. To establish a causative link, training studies should be conducted training one of the fall risk factors, like balance or RTD, and changes in the ACE performance should be monitored.

Future modifications to the ACE test could broaden its application and refine the tasks used. Segment C was initially designed to pose a cognitive challenge to the participants but in practice, the order of the cones could be seen immediately when reaching the segment for the first time thus potentially eliminating the need to further challenge spatial orientation and perception. A task more aimed at reactive agility could be introduced. For example, the assessor could be placed at the finish and hold up numbers one to three or colored cones when the participant reaches Segment C's turning point. In this manner, subjects would have to react to multiple external stimuli rather than just one, improving the cognitive challenge. Yet, standardization for this utilizing an assessor seems difficult. As it stands, the biggest cognitive challenge of the course seemed to be the memorization of the exact order of tasks to be performed. The distance of stops in Segment A was designed to include variable distances, but it could be argued that the distance between the stops is too small to allow for proper acceleration and thus the velocity from which the deceleration has to be done is too low to mimic similar situations in real life. Fewer stops could be included and it is conceivable to also include stops on external stimuli as a task rather than pre-planned.

CONCLUSION

Our ACE course showed that distinct neuromuscular and cardiocirculatory components might differently contribute to agility. Overall agility performance was mainly explained by cardiocirculatory fitness (6MWT) and gait speed. These components could be either tackled by health-related exercise training with a special emphasis on endurance or by integrated agility training. We further conclude that agility performance relies on a broad range of distinct neuromuscular performance variables that should be integratively and functionally assessed. These performance variables included trunk strength, static and dynamic balance performance as well as ankle muscle power. These parameters might be predictive of ACE performance to varying degrees. Future studies could develop a training method based on our ACE approach and compare it with traditional training concepts in different settings and should also address whether this approach decreases fall rates among seniors.

AUTHOR CONTRIBUTIONS

EL and LD were responsible for the conception and design of the study. EL, AZ, and RaR were responsible for the acquisition of the data. RoR, EL, OF, and LD were responsible for interpreting and analyzing the data. EL, OF, and LD drafted the manuscript. LZ, TH, JvD, AZ, RoR, and RaR revised the content and gave important intellectual input to the discussion. All authors read and approved the final version of the manuscript.

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Maximal Eccentric Hamstrings Strength in Competitive Alpine Skiers: Cross-Sectional Observations From Youth to Elite Level

Martino V. Franchi[1,2*†], Lynn Ellenberger[1†], Marie Javet[3,4], Björn Bruhin[3,4], Michael Romann[3], Walter O. Frey[5] and Jörg Spörri[1*]

[1] Sports Medical Research Group, Department of Orthopaedics, Balgrist University Hospital, University of Zurich, Zurich, Switzerland, [2] Laboratory for Muscle Plasticity, Department of Orthopaedics, Balgrist University Hospital, University of Zurich, Zurich, Switzerland, [3] Section for Elite Sport, Swiss Federal Institute of Sport Magglingen, Magglingen, Switzerland, [4] Swiss-Ski, Muri bei Bern, Switzerland, [5] Balgrist Move>Med, Department of Orthopaedics, Balgrist University Hospital, University of Zurich, Zurich, Switzerland

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Edited by:
Lars Donath,
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Edith Cowan University, Australia

***Correspondence:**
Martino V. Franchi
martino.franchi@balgrist.ch
Jörg Spörri
joerg.spoerri@balgrist.ch

†These authors have contributed equally to this work as first authorship

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Competitive alpine skiers are subject to substantial risks of injury, especially concerning the anterior cruciate ligament (ACL). During "landing back weighted" episodes, hamstrings may partially counteract the anterior shear force acting on the tibia by eccentrically resisting the boot-induced drawer of the tibia relative to the femur. The aim of the present study was to provide novel descriptive data and sport-specific reference values on maximal eccentric hamstrings strength (MEHS) in competitive alpine skiers from youth to elite level, and to explore potential relationships with sex, age and biological maturation. 170 competitive alpine skiers were investigated: 139 youth athletes (51 females, 88 males; age: 13.8 ± 0.59 years) and 31 elite athletes (19 females, 12 males; age: 21.7 ± 2.8 years). MEHS was assessed by the (Vald Performance, Newstead, Australia). U15 female skiers presented lower MEHS compared to female elite skiers for both limbs (R = 210 ± 44 N vs. 340 ± 48 N, respectively, $p < 0.001$, and L = 207 ± 46 N vs. 303 ± 35 N, respectively, $p < 0.001$). Similarly, lower MEHS was observed in U15 male skiers compared to male elite skiers for both limbs (R = 259 ± 51 N vs. 486 ± 62 N, respectively, $p < 0.001$, and L = 258 ± 57 N vs. 427 ± 54 N, respectively, $p < 0.001$). Correlations between MEHS and chronological age were modestly significant only for the U15 group (r = 0.37 and $p < 0.001$). When the correlations for the U15 group were performed between MHES and maturity offset (obtained from the calculation of biological age, i.e., age at peak height velocity), statistical significance was reached by all the correlations run for 3 variables (Males < 0: r = 0.59, $p < 0.0001$; Males > 0: r = 0.70, $p < 0.0001$; and Females > 0: r = 0.46, $p < 0.0001$, start of maturity offset = 0). This cross-sectional description of MEHS in alpine skiers from youth to elite level highlights the importance of biological maturation for MEHS values in youth athletes and presents novel data that may offer insights into new approaches for injury prevention.

Keywords: conditioning, physical fitness, neuromuscular performance, testing, biological maturity status, athletes, injury prevention, alpine ski racing

INTRODUCTION

Competitive alpine skiers are known to be subject to substantial risks of injury (Spörri et al., 2017). Although the rates for some injuries have been recently reported to show a decline as stated by Färber et al. (2018), the possibility for skiers to sustain an anterior cruciate ligament (ACL) injury during their sportive career is still very high (Pujol et al., 2007; Flørenes et al., 2009, 2012; Westin et al., 2012, 2018; Bere et al., 2013a; Stenroos and Handolin, 2014; Haaland et al., 2016; Müller et al., 2017b). Most of the ACL-injuries occur while the skier is turning or landing from a jump (i.e., before or without falling) (Bere et al., 2011, 2014). Typical ACL-injury mechanisms include excessive knee joint compression, knee valgus and internal rotation, or a boot-induced anterior drawer of the tibia relative to the femur (Bere et al., 2011, 2013b; Jordan et al., 2017; Spörri et al., 2017).

Physical aspects of the athlete have been suggested to be among the top 5 key injury risk factors in alpine ski racing (Spörri et al., 2012) and fitness parameters have been shown to be associated with injury risk (Raschner et al., 2012; Müller et al., 2017a). During typical ACL-injury mechanisms, such as the "landing back weighted" mechanism, hamstring muscles may act as an ACL-synergist by producing a posteriorly directed shear force to the tibia (i.e., by eccentrically resisting the boot-induced anterior drawer of the tibia relative to the femur while landing).

Considering that both quadriceps and hamstrings muscle groups are significantly activated during jump landings (Färber et al., 2018), it is reasonable to enquire whether enhanced co-activation of such muscle groups contributes to prevention strategies (Oberhofer et al., 2017). However, previous research and opinion targeting quadriceps functional features and ACL-injuries has been controversial, as Färber et al. (2018) pointed out. Instead, hamstrings strength capacity may be of importance for many typical injury situations (e.g., jump landings or backward falls) (Read and Herzog, 1992; Herzog and Read, 1993; Gerritsen et al., 1996; DeMorat et al., 2004; Koyanagi et al., 2006; Semadeni and Schmitt, 2009; Bere et al., 2011, 2014; Yeow et al., 2011; Heinrich et al., 2018). In fact, if hamstrings are pre-activated fast and high enough (Färber et al., 2018), tibial anterior translation relative to the femur might be reduced, consequently diminishing the risk of ACL-injury.

Eccentric muscle actions are an inherent part of skiing (Berg et al., 1995; Kröll et al., 2015a,b), and specifically, sufficient eccentric hamstrings strength is considered to be important for ACL-injury prevention in skiers (Jordan et al., 2017; Spörri et al., 2017) and in athletes in general (Bourne et al., 2018). However, to date, there is no study that comprehensively investigated maximal eccentric hamstrings strength (MEHS) neither in youth nor in elite competitive alpine skiers. Thus, although it could be of significant interest for injury prevention strategies, to our knowledge, there is no presence in literature of any observations regarding relationships between sex, sportive level, chronological age/biological maturation and MEHS in competitive alpine skiers. Gaining further information on such parameters could help to identify potential new stratagems for ACL-injury prevention in youth and elite skiers and to better understand how to implement MEHS related prevention strategies effectively.

Accordingly, the sub-goals of the present study were: (1) to screen two distinct populations of competitive alpine skiers (including youth athletes and elite athletes) by assessing MEHS during Nordic Hamstrings Exercise (NHE), which has extensively been used in different sports such as Australian football, rugby, soccer and sprinting (Opar et al., 2013; Timmins et al., 2016); (2) to conduct a cross-sectional observation (from youth to elite level) on various relationships between sex, sportive level, age, biological maturation and MEHS. The overall aim of the present study was to provide novel descriptive data and reference values on MEHS in competitive alpine skiers, which could be of strategical interest for future novel injury prevention approaches starting from youth competitive level and age.

MATERIALS AND METHODS

Participants and Study Design

In total 170 competitive alpine skiers participated in the study: 139 U15 youth athletes (51 females, 88 males; mean age: 13.8 ± 0.6 years; range: 12.9 – 14.9 years) and 31 adult athletes (19 females, 12 males; mean age: 21.7 ± 2.8 years; range: 17.0 – 28.9 years). **Table 1** provides detailed anthropometric data separated by gender and groups of youth and adult elite skiers. Measurements were completed during the preseason (October 2017–November 2017) for youth elite alpine skiers and during off-season (May 2018–June 2018) for national level ski racers. This study was carried out in accordance with the recommendations of the institutional review board and local ethic committee with written informed consent from all subjects in accordance with the Declaration of Helsinki. Study approval was granted by the institutional review board and local ethic committee (KEK-ZH-NR: 2017-01395).

Maximal Eccentric Hamstring Strength During NHE

The maximal eccentric hamstring strength was assessed by using a NHE measurement device (Vald Performance, Newstead, Australia); its reliability and application on athlete populations is reported in several previous studies (Bourne et al., 2015; Opar et al., 2015; Timmins et al., 2016). Briefly, athletes knee on a padded board of the Norbord device with their ankles fixed by braces right above the lateral malleoli. The ankle braces contain integrated uniaxial load cells which are affixed to a pivot in order to ensure a constant force measurement through the longitudinal axis of the load cell. Directly prior to the measurement an investigator demonstrated the NHE to each athlete. The following verbal instructions were provided as previously described (Bourne et al., 2015; Opar et al., 2015): gradually lean forward at the slowest possible speed; maximally resist this movement with both legs; keep trunk and hips in a neutral position throughout the movement; hold hands crossed above the chest. A repetition was completed if the resulting forces overcame the athlete's resistance and pressurized a catch of the movement with the hands on the ground. All participants performed one set of three repetitions of NHE (5–10 s of rest between repetitions), whereby they were verbally encouraged to secure maximal exertion. Based on the previously described instructions a trial was considered valid if it demonstrated a constant increase of force progression culminating in a pronounced force peak, followed by a rapid decline. The best left and right maximum values of the three repetitions were used for further data analysis. The limbs asymmetry during MEHS production during NHE test was calculated as the difference between stronger and weaker leg expressed as percentage.

Biological Age and Maturity Offset Calculation

The biological age was calculated based on a formula by Mirwald et al. (2002) which provides a non-invasive and previously validated method to predict the age at peak height velocity (APHV) (Malina et al., 2007; Sherar et al., 2007) and moreover was validated for youth competitive alpine skiers (Müller et al., 2015). The gender-specific equations use anthropometric measures of body mass (0.1 kg, *Seca, Hamburg, Germany*), body height and sitting height (0.5 cm, determined by measuring tape), as well as chronological age at the time of measurement and sub-ischial leg length as the difference between body height and sitting height. Based on the collected data the prediction of an individual maturity offset is enabled, which marks a point in time before or after peak height velocity (PHV). The estimated APHV is given by subtracting the maturity offset from the actual chronological age (Mirwald et al., 2002).

Statistical Analysis

Data were reported as mean ± SD. Differences between groups were statistically analyzed for MEHS values using an unpaired Student's *t*-test. Correlations between sex, chronological age, biological age and MEHS and were tested by the Pearson's product moment correlation coefficient (r) and coefficient of determination (r^2). The level of significance was set at $p < 0.05$.

TABLE 1 | Anthropometric data for male and female athletes separated by groups.

	U15 athletes		Elite athletes	
	Female	**Male**	**Female**	**male**
	Mean (±SD) (min-max)	**Mean (±SD) (min–max)**	**Mean (±SD) (min-max)**	**Mean (±SD) (min-max)**
Age [y]	13.7 ± 0.6 (12.5–14.9)	13.9 ± 0.5 (12.9–14.8)	21.3 ± 2.7 (17–26.3)	22.4 ± 2.9 (18.3–28.9)
Body height [cm]	159.7 ± 6.3 (143–171.5)	161.4 ± 8.4 (145–185)	166.4 ± 5.7 (155–180)	176.3 ± 6.7 (166–189)
Body weight[kg]	48.4 ± 7.6 (35–66.5)	49.4 ± 10.3 (30–81)	65.5 ± 5.9 (50–82)	80.8 ± 6.2 (71–103)
BMI [kg/m^2]	18.9 ± 1.2 (14.4–25.9)	18.9 ± 1.2 (13–24.7)	23.6 ± 1.8 (20–26.3)	25.9 ± 1.3 (23.4–29)

SD, standard deviation.

RESULTS

Maximal Eccentric Hamstring Strength During NHE

A total of 170 competitive alpine skiers performed a maximal NHE-test. A comprehensive snapshot of the differences in strength between females and males for different limbs and at different age/sportive level is presented in **Figures 1A,B**. **Figure 1A** shows that U15 female skiers (n = 51) presented a significantly lower eccentric hamstrings strength compared to female elite skiers (n = 19) for both right and left limbs (R = 210 ± 44 N vs. 340 ± 48 N, respectively, $p < 0.001$, and L = 207 ± 46 N vs. 303 ± 35 N, respectively, $p < 0.001$). Similarly, **Figure 1B** shows that a significantly lower eccentric hamstrings strength was observed in U15 male skiers (N = 88) compared to male elite skiers (n = 12) for both limbs (R = 259 ± 51 N vs. 486 ± 62 N, respectively, $p < 0.001$, and L = 258 ± 57.2 vs. 427 ± 54 N, respectively, $p < 0.001$). Male skiers always presented significantly higher values of eccentric hamstrings strength irrespective of limb, age, or sportive level compared to female skiers ($p < 0.001$).

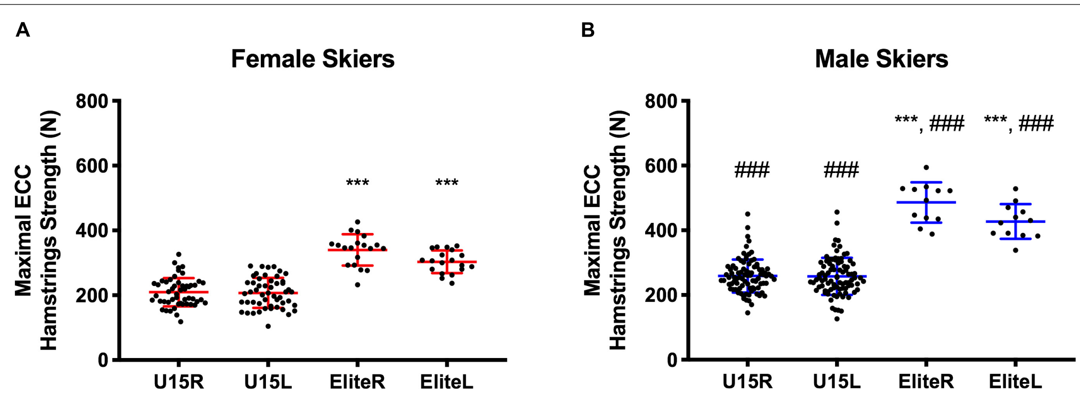

FIGURE 1 | Maximal eccentric hamstrings strength of right and left limbs for U15 and Elite athletes. *** = significantly different between Elite and U15 group of the same sex, $P < 0.001$; ### = significantly different between the same age/sportive level but of different sex, $P < 0.001$. **(A)** female, **(B)** male.

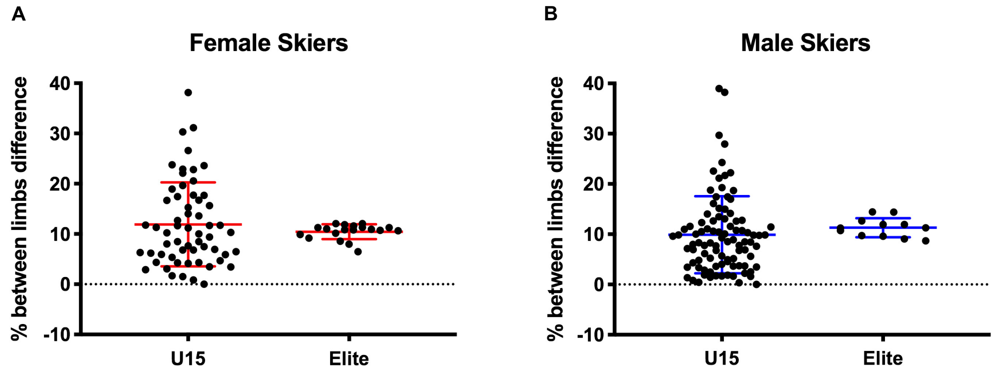

FIGURE 2 | Limbs asymmetry of MEHS production during NHE for female and male skiers of both sportive level groups. **(A)** female, **(B)** male.

Between Limbs Imbalance (Asymmetry) During NHE

Data for asymmetry of force production between right and left limb during NHE (difference between stronger and weaker leg expressed as percentage), are presented in **Figures 2A,B** for U15 and elite skiers, females and males, respectively. U15 female skiers presented similar values of asymmetry for eccentric hamstrings strength production compared to female elite skiers (11.91 ± 8.3% vs. 10.46 ± 1.47%, $p = 0.45$); in a very similar fashion, U15 male skiers showed no significant differences of asymmetry for eccentric hamstrings strength production when compared to male elite skiers (9.88 ± 7.67% vs. 11.31 ± 1.89%, $p = 0.52$). However, it is worth mentioning that compared to elite skiers, U15 skiers showed a higher variance in the asymmetry values observed.

Associations Between Maximal Eccentric Hamstrings Strength, Sex, Age and Maturity Offset

Correlations between MEHS and chronological age are presented in **Figure 3** for the elite skiers and in **Figure 4** for the U15 skiers, showed as grouped data (**Figures 3A**, **4A**), as well as the data obtained when accounting for sex differences (**Figures 3B,C**, **4B,C**). Pearsons' r and r^2-values, together with statistical significance, are shown in **Figures 3**, **4**.

Pearson's correlation between MEHS and chronological age did not reach any statistical significance in the elite group, neither for the grouped data ($r = 0.30$, $r^2 = 0.09$, $p = 0.1$) nor for female and male groups ($r = 0.14$, $r^2 = 0.01$, $p = 0.56$; $r = 0.40$, $r^2 = 0.16$, $p = 0.18$). Conversely, the correlations for the U15 group were observed to be statistically significant when the data were grouped ($r = 0.37$, $r^2 = 0.14$, $p < 0.001$) and when the data were expressed by sex (Females: $r = 0.26$, $r^2 = 0.07$, $p < 0.05$ and Males: $r = 0.40$, $r^2 = 0.16$, $p < 0.001$). When the correlations for the U15 group were performed between MEHS and maturity offset (**Figure 5**), statistical significance was reached by all the correlations run for 3 variables (Males < 0: $r = 0.59$, $r^2 = 0.35$, $p < 0.0001$; Males > 0: $r = 0.70$, $r^2 = 0.49$, $p < 0.0001$; and Females > 0: $r = 0.46$, $r^2 = 0.22$, $p < 0.0001$, where 0 represents the start of maturity offset).

Body Weight Normalized Maximal Eccentric Hamstrings Strength vs. Maturity Offset

In the group of U15 skiers, correlations between MEHS and maturity offset disappear when a normalization with body weight is performed (**Figure 6**) (Males < 0: $r = -0.1$, $r^2 = 0.01$, $p = 0.38$; Males > 0: $r = 0.20$, $r^2 = 0.04$, $p = 0.32$; and Females > 0: $r = -0.23$, $r^2 = 0.05$, $p = 0.08$, where 0 represents the start of maturity offset). Accordingly, a relative MEHS value-based ranking among the skiers is apparently different than an absolute value-based ranking (**Figure 5**).

DISCUSSION

The present investigation aimed to provide a cross-sectional observation of MEHS values in 170 competitive skiers (139 U15 athletes vs. 31 elite athletes). The main findings were the following: (1) greater MEHS during NHE was observed by the elite skiers compared to the U15 group and greater strength was developed by male compared to female skiers for both groups. (2) While no correlation was found between strength and chronological age in elite skiers, a weak to moderate association was found in the U15 group ($r = 0.37$ and $r^2 = 0.14$). However, when strength was correlated to maturation offset for the latter group, this association showed moderate to strong linear relationships in a gender dependent manner (Males < 0: $r = 0.59$, $r^2 = 0.35$; Males > 0: $r = 0.70$, $r^2 = 0.49$; and Females > 0: $r = 0.46$, $r^2 = 0.22$; where 0 represents the start of maturity offset). (3) In the U15 athletes, a body weight normalization of the MEHS values removes any relations to maturity offset.

Toward Alpine Skiing-Specific Reference Values of Maximal Eccentric Hamstrings Strength and Between-Limb Imbalance

The individual and average absolute values for MEHS for both limbs are presented in **Figure 1**. The mean value for the male elite

FIGURE 3 | Correlations between MEHS and chronological age for **(A)** Elite Swiss Teams skiers irrespective of sex, **(B)** Elite female skiers, and **(C)** Elite male skiers.

skiers was 486 ± 62.39 N for the right leg and 427.1 ± 53.62 N for the left one: compared to other previous studies, these values are considerably above the average found for elite Australian footballers (which did not sustain hamstrings injuries, average of left and right limbs = 301 ± 84 N, n = 159) (Opar et al., 2015), and for elite Rugby Union players (which did not sustain hamstrings injuries, average of left and right limbs = 367.7 ± 85 N, n = 158) (Bourne et al., 2015), and for football (soccer) players (which did not sustain hamstrings injuries, average of left and right limbs = 309.5 ± 73.4 N, n = 105) (Timmins et al., 2016). It must be stated that these higher force values may be also due to the high force production in the antagonists of the hamstring muscles (i.e., the knee extensors), which is typical for alpine ski racing and, therefore (Berg et al., 1995; Berg and Eiken, 1999), a priority in the conditioning of competitive alpine skiers. Lower values were observed for elite female athletes and U15 male and female skiers: however, to the best of our knowledge, no previous reports of MEHS (measured during NHE) were found comparable to these cohorts. The present study aimed purposely to provide the literature in sports medicine research with new sport-specific reference data on different cohorts of elite competitive skiers, also for multiple comparisons with different athletic populations.

Average values of between limb imbalance (asymmetry) for force developed during NHE was similar between the U15 and Elite groups and between sexes (F = 11.91 ± 8.3% vs. 10.46 ± 1.47%, respectively; M = 9.88 ± 7.67% vs. 11.31 ± 1.89%, respectively) (**Figure 2**). The present values are very similar to the ones previously showed for limbs in which no hamstrings injury had occurred in Australian footballers and rugby union players (Bourne et al., 2015; Opar et al., 2015). Moreover, it is worth highlighting that the individual values for U15 groups showed a great variability of between limb imbalance, possibly suggesting that U15 coordination strategies during NHE are not strongly consolidated.

The Unexplored Role of Maximal Eccentric Hamstring Strength and Between-Limb Imbalance for the Risk of ACL Injuries in Alpine Ski Racing

An indirect (i.e., etiology/injury mechanism-based) justification of why MEHS may be of importance for the purpose of ACL injury prevention in alpine ski racing can be found in the following theoretical considerations. During typical ACL-injury mechanisms, such as the “landing back weighted” mechanism, hamstring muscles may functionally counteract the anterior shear force acting on the tibia (i.e., by eccentrically resisting the boot-induced anterior drawer of the tibia relative to the femur while landing). This hypothesis is further supported by the simulation study findings of (Semadeni and Schmitt, 2009), the fact that hamstring muscle activation levels can be voluntarily increased during jump landing (Färber et al., 2018), as well as the evidence of multimodal neuromuscular injury prevention programs (and NHE in particular) being effective in the reduction of the risk of ACL injury in sports other than alpine ski racing (Petushek et al., 2018). Moreover, higher values of between-limb imbalance (i.e., ranging from 21.2 to a 13.1% between start and the end of pre-season) have been associated with a risk of hamstring injury in rugby (Bourne et al., 2015). At the same time, it is still unclear if MEHS and/or between-limb imbalance could represent a risk factor for (side-dependent) ACL-related injuries

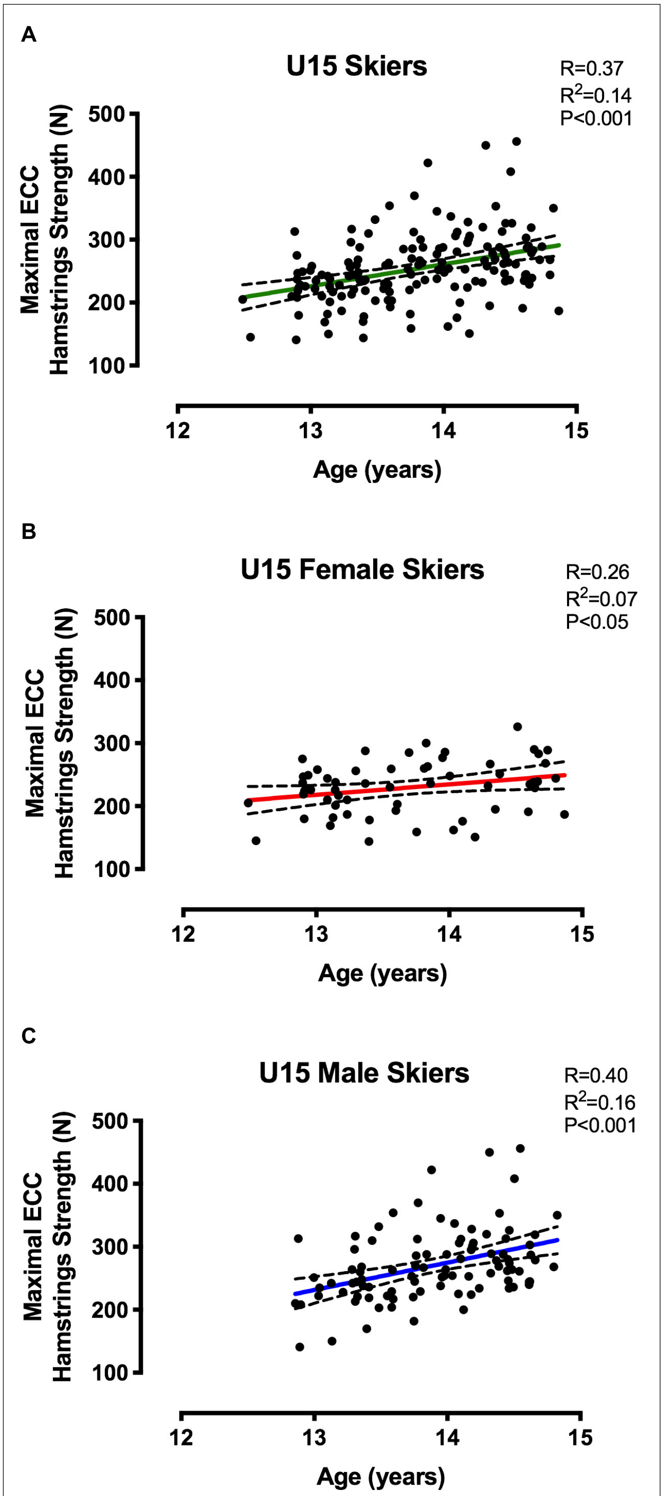

FIGURE 4 | Correlations between MEHS and chronological age for **(A)** U15 skiers irrespective of sex, **(B)** U15 female skiers, and **(C)** U15 male skiers.

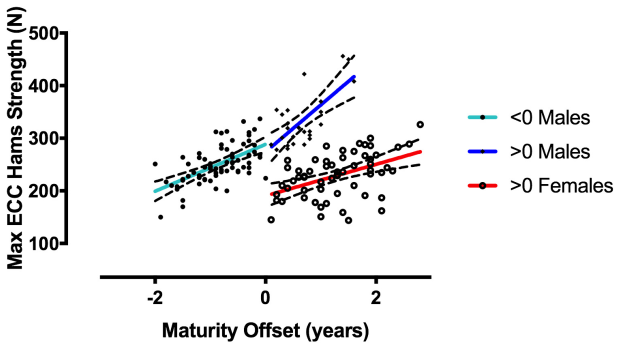

FIGURE 5 | Correlations between MEHS and maturity offset is represented by 0 value. Males < 0: $r = 0.59$, $r^2 = 0.35$, $p < 0.0001$; Males > 0: $r = 0.70$, $r^2 = 0.49$, $p < 0.0001$; and Females > 0: $r = 0.46$, $r^2 = 0.22$, $p < 0.0001$.

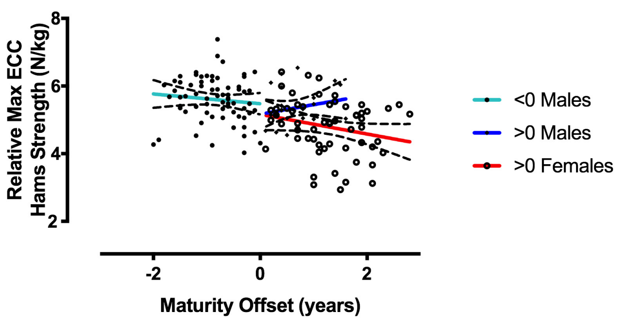

FIGURE 6 | Correlations between relative MEHS (N/kg) and maturity offset is represented by 0 value. No significant correlations observed.

in the sport of alpine ski racing. Accordingly, in a next step, longitudinal (i.e., epidemiology, etiology and/or intervention-related) studies are needed to verify the hypothesis of a direct association between MEHS, between limb imbalance and the risk of ACL injuries in the sport of alpine ski racing.

However, irrespective of these future aims, it is important to know about sport-specific reference values, potential age/maturity related influences and asymmetry problems in the corresponding populations, as being explored in the current study. Such information is essential for the interpretation of forthcoming longitudinal studies, and to better understand how to implement MEHS related prevention strategies effectively.

The Associations of Sex, Sportive Level, Chronological Age and Biological Maturation With Maximal Eccentric Hamstrings Strength

It was no surprise that elite skiers (ranging from 17 to 28 years old) showed greater MEHS compared to the younger cohort (ranging from 12 to 15 years old): however, we further aimed to clarify if such discrepancy was just due to the age difference or to the fact that the two groups belonged to two distinct sportive levels (**Figures 3–5**). MEHS was not associated to chronological age in elite skiers: this may indicate that the individual differences in strength between elite athletes were potentially more due to training-related than just temporal factors.

Conversely, U15 athletes showed significant correlations between MEHS and chronological age when subjects were grouped irrespectively of gender (r = 0.37) and when divided for sex (r = 0.26 for female skiers and r = 0.40 for male skiers). However, these correlations presented very low r^2-values (0.14, 0.07, and 0.16, respectively), thus explaining, in the best of the cases (i.e., for male skiers), only up to 16% of the variability of strength vs. age. Accordingly, we decided to investigate the relationship of eccentric hamstrings strength for female and male skiers in function of maturity offset (obtained from the calculation of biological age, i.e., age at peak height velocity). Interestingly, these relationships resulted in higher r^2-values (**Figure 5**): males presented the most significant relationships before (r^2 = 0.35) and after (r^2 = 0.49) peak height velocity, while females showed and r^2-value of 0.22 after peak height velocity. It must be specified that all of the female subjects have already reached their peak height velocity, so we could not present any relationship between maximal eccentric strength and maturity offset in the months/years before 0 value (i.e., the actual maturity offset). This is due to the fact that in this and other studies females reached their peak height velocity earlier than males (possibly around 11–12 years old) (Müller et al., 2017a,b).

The male subjects who already reached their peak height velocity were the ones that presented the higher absolute values for MEHS: while this was an expected finding, it is worth highlighting that few athletes within a year before PHV showed similar, if not greater, values of MEHS compared to other skiers which already passed PHV. In our opinion, these observations could be potentially regarded as selection criteria for either successive injury risk or general athletic performance, as young skiers that present such values of MEHS before complete maturation may start from a better overall condition compared to their peers, in a future perspective.

Future Perspectives

The aforementioned correlations between biological maturation and MEHS suggest that in younger cohorts is important to consider if an athlete has already reached her/his peak height velocity point, in order to better interpret force values in key of injury prevention and performance. In fact, the lower force values observed in the U15 male skiers' group were identified for athletes that were between 2 and 1 years before peak height velocity: if one would have considered just strength and/or chronological age for such subjects, potential misinterpretations could have been made for conclusions on risks of injury and/or selection criteria for high performance paths. One potential solution to better address the influence of biological maturation on MEHS in future might be found in a normalization with body weight (i.e., relative strength, N/kg). On the current study population such a normalization removed any relations between maturity offset and MEHS (**Figure 6**). However, for providing a deeper understanding of the long-term development of MEHS during the sportive career and/or over the entire life-span, future research should address the topic by the use of longitudinal study designs.

CONCLUSION

This study aimed on a cross-sectional description of MEHS in competitive alpine skiers from youth to elite level. It may provide reference values and background knowledge for the interpretation/implementation of future ACL-injury prevention and athletic conditioning studies/interventions in the sport of alpine ski racing. Moreover, it highlighted the importance of considering biological maturation for meaningful interpretations of force values of youth athletes that are close to their growth spurts. Future investigations of MEHS in the context of ACL-injury prevention and/or athletic conditioning should focus on longitudinal observations of the same athletes during their sportive career. More integrative approaches should be implemented, such as combining muscle function testing with ultrasound-based assessment of hamstrings muscle mechanical behavior, its architectural adaptations to longitudinal training and the investigation of potential underlying molecular mechanisms.

AUTHOR CONTRIBUTIONS

JS and WF conceptualized the study. JS, LE, and MJ conducted the data collection. MF and LE contributed to the analysis and interpretation of the data. MF, LE, and JS drafted the manuscript. All other authors revised it critically, and approved the final version and agreed to be accountable for all aspects of this work.

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