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Acute and chronic changes in postural control in response to different physiological states and external environmental conditions

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Acute and chronic changes in postural control in response to different physiological states and external environmental conditions

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Editorial: Acute and chronic changes in postural control in response to different physiological states and external environmental conditions

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Editorial on the Research Topic
Acute and chronic changes in postural control in response to different physiological states and external environmental conditions

Introduction

Human postural control is fundamental for motor skill learning and the performance of everyday and sports-related tasks (Gebel et al., 2020). Previously, postural control has been defined as the control of the body's position in space for stability and orientation to maintain and recover balance (Shumway-Cook and Woollacott, 2012). Postural balance is regulated through the complex processing of proprioceptive, vestibular and visual information on a spinal and/or supraspinal level within the central nervous system (Taube et al., 2008). Depending on the postural parameter under investigation, static and dynamic balance follow a *U*-shaped (postural sway) or inverse *U*-shaped curve (gait speed) across the lifespan. In other words, postural balance is not yet fully developed during childhood and adolescence which is for instance established in a larger dependency on visual information for the maintenance of upright stance (Granacher et al., 2011). In seniors, postural balance deteriorates due to biological aging of the central nervous system (e.g., desensitization of muscle spindles) and increased physical inactivity. When confronted with environmental stressors such as bad lightening, unstable surfaces, multitask situations, postural balance is impaired resulting in an increased risk of sustaining injuries and/or falls. Physical exercise or training induce sensations of fatigue resulting in acute balance declines (Gebel et al.). This is indicated for instance in an increased risk of sustaining injuries in the second half of soccer games (Hawkins and Fuller, 1999). While physical exercise can provoke acute negative effects on postural balance (Paillard, 2012), it has also positive effects when chronically applied over longer intervention periods (Paillard, 2017), irrespective of the population under investigation [youth, adult (athletes), seniors]. There is evidence from original research (Granacher et al., 2010a,b; Wälchli et al., 2018) and systematic reviews and meta-analyses (Lesinski et al., 2015a,b; Sherrington et al., 2017; Gebel et al., 2018; Al Attar et al., 2022) on the effects and dose response relations of balance-related interventions (balance training, slackline training, perturbation training etc.) on measures of balance and the risk of injuries and falls. Less is known on the underlying physiological mechanisms responsible for the exercise induced adaptations. With this Frontiers in Neuroscience Research Topic, we encouraged researchers to further explore the topic by providing knowledge on the acute and chronic changes in postural control in response to different physiological states and external environmental conditions to better understand how the human body responds to physiological stress/exercise in terms of postural control.

Summary of selected articles from this Research Topic

Overall, twelve articles (ten original research articles, one review article, one brief report) were published in this Research Topic between 2021 and 2023. Fifty-five researchers from different countries across the globe including Austria, Canada, China, France, Germany, Hungary, Iran, Italy, Slovakia, Slovenia, South Africa, the Netherlands, and the United Kingdom participated in this Research Topic. In January 2023, the Research Topic received 22,935 total views and 3,496 total downloads. In terms of demographics, the Research Topic received particular interest in the United States of America (2,661 views), Belgium (1,190 views), Germany (504 views), China (254 views), and finally Slovenia (22 views). The twelve articles touched clinical (limb asymmetries, spasticity) and sports-related topics (fatigue, blood flow restriction, compression garments) surrounding the overarching topic postural control. In the clinical setting, the examined study cohorts consisted of obese, post-stroke or knee osteoarthritis patients. In the sports-related setting, researchers examined (youth) athletes as well as the general youth population and healthy adult individuals.

Fadillioglu et al. (2022)

Although the effects of the stomatognathic motor system on postural balance have been established under static conditions, it has not yet been studied during dynamic steady-state balance. Fadillioglu et al. analyzed the effects of controlled stomatognathic motor activity on the control of the center of mass (COM) during dynamic balance under different stomatognathic motor conditions. The experimental conditions comprised jaw clenching, tongue pressure, and habitual stomatognathic behavior. Findings from this study indicate that deliberate jaw clenching or tongue pressing does not appear to affect dynamic steady-state balance. Due to balance task-specificity, further research is needed on the effects of stomatognathic motor activities on dynamic balance in different movement tasks.

Fu et al. (2021)

The aim of the study reported in the article of Fu et al. was to compare individuals with knee osteoarthritis and asymptomatic controls with regards to their postural balance and identify kinematic and lower extremity muscle activity characteristics during a stand-to-sit task. A comprehensive experimental study showed significantly larger COM displacements and peak instantaneous COM velocity in the anterior-posterior direction, reduced ankle dorsiflexion range of motion, greater anterior pelvic tilt range of motion, and lower quadriceps femoris and muscles activation level coupled with higher biceps femoris muscle activation level during the stand-to-sit task of the osteoarthritis patients. Findings from this study imply that rehabilitation programs targeting disease-related impairments could be beneficial for restoring the functional transfer in individuals with knee osteoarthritis.

Gebel et al. (2022)

In this original research article, Gebel et al. aimed to examine the effects of physical and mental fatigue on postural sway and cortical activity in healthy young adults. Before and after balance testing using a pressure sensitive mat and electroencephalography, participants performed in random order an all-out repeated sit-to-stand task to induce physical fatigue and a computer-based attention network test to provoke mental fatigue. The applied physical fatigue protocol resulted in increased postural sway accompanied by enhanced alpha-2 power in the parietal region of interest. Mental fatigue led to increased postural sway variability and alpha-2 power. The observed fatigue-related changes in cortical activity indicate impairments in sensory information processing related to movement planning and execution within the somatosensory cortex, resulting in balance declines.

Heil (2022)

Inter-limb asymmetries are associated with non-contact injuries, particularly in team sports (Dos'Santos et al., 2021). In this original research article, Heil investigated the influence of different physical fatigue protocols on inter-limb asymmetries during the performance of a dynamic postural control test in a rather large cohort of 128 young and physically active male and female adults. Following a between-subject design, participants were allocated to a fatigue protocol either executed on a bike ergometer (30 s all-out Wingate test) or a treadmill (10 s all-out test at maximal running velocity and a slope of 7.5%). Before and after the fatigue protocol, all participants performed the lower limbs Y-balance test in anterior direction. The analysis revealed that inter-limb asymmetries did not increase due to fatigue. No significant differences were found between the running and the cycling protocol. The author suggested that fatigue protocols of longer duration and acyclic character (e.g., change-of-direction tasks) should be examined in future studies.

Mahmoudzadeh et al. (2021)

The article of Mahmoudzadeh et al. evaluates the effects of spasticity of ankle plantar flexors on balance in post-stroke patients and determines the relationship between the spasticity with ankle proprioception, passive ankle dorsiflexion range of motion, and balance confidence. The results of the study show that joint proprioception was significantly better in the low spasticity group compared to the high spasticity group and that there were no significant relationships between the spasticity severity and passive ankle dorsiflexion range of motion, and balance confidence. The authors conclude that postural balance is significantly affected in post-stroke patients, regardless of the severity of the ankle plantar flexor spasticity.

Noé et al. (2022)

Based on the observation that inter-limb balance asymmetries increase the risk of injury in athletes, Noé et al. analyzed the effects of wearing compression garments (CG) to reduce these asymmetries. Their results show that inter-limb asymmetries were lower with CG in participants with high levels of asymmetries at baseline while they were higher in participants with low levels of asymmetries at baseline. Thus, in order to reduce inter-limb asymmetries and the associated injury risk in athletes, wearing CG has a beneficial effect in the presence of high levels of inter-limb balance asymmetries at baseline. Conversely, CG should be avoided in individuals with low baseline balance asymmetries as they likely produce confusion and overload in sensorimotor processing.

Russell et al. (2022)

Russell et al. conducted a study to investigate the effects of placebo and nocebo effects on postural stability. While the placebo effect has been shown to impact postural stability before (Villa-Sánchez et al., 2019), Russell et al. included both objective (measured) and subjective (perceived) measures of postural stability. The participants in the placebo/nocebo groups were given an inert capsule described as a potent supplement which would either positively or negatively influence their postural stability. As expected, an increase in body sway and reduced perceived stability were noted in the nocebo condition, and the opposite was true for the placebo group. Additional analyses also revealed that performance expectations heavily influenced the perception of postural instability. These results indicate that postural control (and its perception) is susceptible to expectation manipulation (both placebo and nocebo), which has important practical implications.

Sozzi and Schieppati (2022)

Sozzi and Schieppati investigated the adaptation of postural control (postural sway) on a compliant surface (i.e., a foam) across eight repetitions, and analyzed how visual sensory information and light touch influence postural control. Previous studies have examined balance adaptations in paradigms with external perturbations. However, balance adaptations during quiet unperturbed stance are unresolved. The authors confirmed that postural sway adaptations occur during standing on a compliant surface. This adaptation is reflected in a progressive increase in the amplitude of the lowest frequencies of the spectrum and a concurrent decrease in the high-frequency range. The authors concluded that the control of balance was shifted from the lower to the higher levels of the nervous system. In addition, the adaptation rate was modestly influenced by light touch.

Voglar et al. (2022)

In their original research article, Voglar et al. compared the effects of supported and unsupported intermittent trunk flexion on postural control during sitting in healthy adult males and females aged 23–24 years. Previous studies have indicated that prolonged spinal flexion (as in, for example, crane operator work) induces changes in trunk motor control and spinal stiffness (Voglar et al., 2016). However, this was one of the first studies to examine the changes in sitting postural control after prolonged flexion. The results indicated that prolonged intermittent flexion does not induce any changes in center of pressure motion during a seated balance task, regardless of the presence of a trunk support. The authors suggested that this was due to a successful compensation of decreased passive stiffness by increased reflex activity.

Wiesinger et al. (2022)

In their study, Wiesinger et al. contrasted measures of postural sway in young overweight/obese (YO) and young normal-weight (YN) children and adolescents. Their results indicated postural control deficits in YO compared to their YN peers, reflected in a more rigid postural control. The authors concluded that without targeted balance exercise, YO are susceptible to end up in a vicious circle of poor balance control and low physical activity. The researchers urge practitioners to implement postural balance training in the early years to avoid impairments later in life.

Willberg et al. (2021)

Muscle fatigue may have acute negative effects on static and/or dynamic postural control which could impair sports performance such as speed skating and alpine skiing. In a within subject design, Willberg et al. examined the acute effects of lower-body blood flow restriction using cuffs around the thighs on static and dynamic postural control in 20 physically active healthy males and females aged 26 years. Using block randomization, participants performed static and dynamic balance tests on a fixed and moveable platform (medio-lateral and antero-posterior direction) in a squatted bipedal position at a knee angle of 110° with inflated cuffs and without cuffs until exhaustion. Blood flow restriction resulted in significantly lower oxygenation of the m. quadriceps femoris and was associated with a significantly lower time to exhaustion compared to the non-restricted condition. Relative postural sway did not differ significantly between non-restricted and the blood flow restriction conditions. In conclusion, blood flow restriction resulted in deoxygenation and a reduced time to exhaustion in the squatted position. Postural control and the ability to regain stability after perturbation were unaffected.

Zemková and Kováčiková (2023)

Zemková and Kováčiková provide a scoping review article examining sport-specific training-induced adaptations of postural control and their relationship to measures of sport performance. The authors show that there is a relationship between measures of static and/or dynamic balance and sports performance in many sports. This can be attributed to an enhanced ability of athletes to make postural adjustments in highly demanding postural tasks. However, the extent to which sport-specific exercises contribute to their superior postural stability is unknown. The authors concluded that further research is needed to investigate the relative contributions of each of these balance exercises to improve sports-performance.

Conclusions

This Frontiers Research Topic contributes to our understanding of the acute and chronic changes in postural control due to different physiological states and external environmental conditions in different cohorts such as patients, healthy youth and adults. Information from this Research Topic can be used to design balance training programs for performance enhancement, rehabilitation, and injury prevention. More research is need on the underlying physiological mechanisms responsible for acute and chronic adaptations within the postural control system.

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

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ORIGINAL RESEARCH
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The Effect of Lower-Body Blood Flow Restriction on Static and Perturbated Stable Stand in Young, Healthy Adults

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Muscular fatigue can affect postural control processes by impacting on the neuromuscular and somatosensory system. It is assumed that this leads to an increased risk of injury, especially in sports such as alpine skiing that expose the body to strong and rapidly changing external forces. In this context, posture constraints and contraction-related muscular pressure may lead to muscular deoxygenation. This study investigates whether these constraints and pressure affect static and dynamic postural control. To simulate impaired blood flow in sports within a laboratory task, oxygen saturation was manipulated locally by using an inflatable cuff to induce blood flow restriction (BFR). Twenty-three subjects were asked to stand on a perturbatable platform used to assess postural-related movements. Using a 2 × 2 within-subject design, each participant performed postural control tasks both with and without BFR. BFR resulted in lower oxygenation of the m. quadriceps femoris (p = 0.024) and was associated with a significantly lower time to exhaustion (TTE) compared to the non-restricted condition [$F_{(1,19)} = 16.22$, $p < 0.001$, $\eta_p^2 = 0.46$]. Perturbation resulted in a significantly increased TTE [$F_{(1,19)} = 7.28$, $p = 0.014$, $\eta_p^2 = 0.277$]. There were no significant effects on static and dynamic postural control within the saturation conditions. The present data indicate that BFR conditions leads to deoxygenation and a reduced TTE. Postural control and the ability to regain stability after perturbation were not affected within this investigation.

Keywords: postural control, dynamic postural control, deoxygenation, muscular fatigue, BFR, perturbation

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INTRODUCTION

In many sport disciplines, athletes must respond quickly to changes in their environment and alter physical forces to maintain an optimal stance. In alpine skiing or speed skating, for example, there is a need to maintain an aerodynamic posture and to resist external disruptions of a stable upright postural state. The muscle activity thereby is mainly isometric and eccentric (Hofmann et al., 2001). Long-lasting, intensive isometric loading has been discussed as a potential cause of peripheral limitation of oxygen (O_2) transport (Degens et al., 1998; Oranchuk et al., 2018). The resulting disparity between energy demand and consumption leads to muscular fatigue (Sjøgaard et al., 1988; Sutton, 1992; Goodall et al., 2014; Grassi et al., 2015), which is defined as a decrease in muscle contractility and, consequently, a loss of performance (Szmedra et al., 2001; Allen et al., 2008; Hettinga et al., 2016). Moreover, joint angles, intramuscular forces, and duration of contraction may also affect muscular blood flow (Konings et al., 2015). If joint angles of the hip or knee decrease or load duration and external forces increase, blood flow is attenuated, and muscle deoxygenation increases (Foster et al., 1999). This has been shown in skating (Foster et al., 1999) and alpine skiing in which a decrease in oxygen saturation of the quadriceps femoris could be observed to values below 10% (Behringer et al., 2018). Previous data also indicate that muscular fatigue impacts on the proprioceptive system by altering the processing of sensory information to the central nervous system, resulting in impaired motor control (Gribble and Hertel, 2004; Barandun et al., 2009; Enoka et al., 2011). Because maintenance of postural control (PC) depends on adequate functioning of the sensory and the motor system (Shumway-Cook and Woollacott, 2017), it can be assumed that PC relates to muscular deoxygenation. PC refers to the ability to maintain, achieve or restore the line of gravity within the base of support (Pollock et al., 2000, p. 405). Therefore, it involves the use of both predictive strategies (i.e., maintaining balance) and compensatory strategies (i.e., regaining balance) (Maki and McIlroy, 1997). With increasing fatigue, it becomes more difficult to maintain or regain an upright posture. The time duration until this position can no longer be maintained is defined as time to exhaustion (TTE), and is used to quantify fatigue in muscle endurance tasks (Pageaux et al., 2013; Van Cutsem et al., 2017).

Papa et al. (2015) have shown that muscular fatigue significantly affects stability and the risk of falls in older adults. This also holds for young and healthy individuals (Ghamkhar and Kahlaee, 2019). In sports, PC is seen as a prerequisite for enhancing performance (Garcia et al., 2011; Andreeva et al., 2020) and is therefore often included in performance diagnostics. Thereby, in static PC assessments, center of pressure displacements on a stable base of support are analyzed using posturography or force plates (Lin et al., 2008; Chaubet et al., 2012; Ahmadi et al., 2017). Even though these tools deliver insight into balance-related processes, they are limited when it comes to drawing conclusion on dynamic and perturbed movements. The importance of those movements in sports is evident: In alpine skiing, for example, athletes need to remain in a stable upright position while being disrupted continuously by the environment, equipment, or even their own movements. Perturbed-dynamic PC tests aim to simulate these perturbations and analyze the ability to regain stable erect posture after disruptions. Some are administered in anticipatory settings. However, because this reflects predictive/anticipatory rather than compensatory abilities, non-anticipatory settings have been used to investigate reactive postural corrections (Piscitelli et al., 2017). Nonetheless, these studies were performed mainly with older subjects or people with cognitive impairment (Sibley et al., 2015) and not in a sport-specific context.

The goal of the present study was therefore to investigate the ability to maintain and regain postural stability after external disturbance and to test whether this ability is changed by deoxygenation of the working muscles. It aimed to investigate whether increased deoxygenation and associated muscular fatigue would influence static and dynamic (perturbed) PC. Blood flow restriction (BFR), was used to decrease O_2 saturation in the muscle. It was hypothesized that unperturbed and perturbed PC would deteriorate with increasing muscular deoxygenation.

MATERIALS AND METHODS

The study was approved in March 2018 by the local ethics committee of the Goethe University Frankfurt (Chair: Prof. Dr. Klein, 2018-34). All 23 subjects (13 female, 10 male; M_{age} = 26.2 female, 27.0 male) were informed about the procedure before the beginning of measurements and gave their written informed consent to participate. Three of them quit the project: two for personal reasons and one due to a lack of angle mobility hindering him in taking the required position. Because subjects were mainly sports students, the sample can be categorized as moderately to highly physically active according to WHO criteria (Pastuszak et al., 2014). None of the subjects was an elite athlete with practice time >6 h per day. All subjects were asked about their health status before the first appointment. Persons with hypotonic blood pressure, cardiovascular disease, or severe diseases that affect the quality of life were excluded. Those having had surgeries within the last 6 months, acute injuries, or the intake of perception-altering substances were also excluded. All subjects were advised not to change their activity-related habits during participation in this study. Body height and weight were measured on a scale (model 920, Seca GmbH & Co. KG, Hamburg, Germany) and subcutaneous fat *via* ultrasound (ACUSON X150, Siemens Medical Solutions United States, Mountain View, CA, United States). **Table 1** reports anthropometric data.

A 2 × 2 within subject design was used to investigate the effects of deoxygenation (with and without BFR) on static and dynamic postural control. Therefore, all 20 subjects participated in four sessions starting without perturbation (NP), followed by one measurement with perturbation (P) in both the BFR and NBFR conditions (**Figure 1**). To avoid sequence effects of the study design, subjects performed the measurements in two sequences of conditions distributed by block randomization. They were split into two equal sized groups, one starting with NBFR conditions, the other with BFR conditions. To eliminate carryover effects, washout periods of at least 48 h were used.

Before PC measurements, one session was used for anthropometric measurements and to define the subjects' occlusion pressure. In each PC condition, subjects were asked to stand bipedally on a platform allowing free movements in mediolateral as well as anterior–posterior directions. They had to take a defined position that was controlled by a

TABLE 1 | Descriptives of subjects after exclusion of three subjects.

	Minimum	Maximum	Mean	*SD*
Age (years)	23	34	26.55	2.56
Height (cm)	162	188	173.25	7.13
Weight (kg)	55	108	69.10	12.82
Thigh length (cm)	52	66	58.52	3.32
Subc. fat (mm)	0.6	5.4	2.21	1.0
Oc. pressure (mmHg)	115	225	148.25	22.96

N = 20, Women = 11, Men = 9.
Subc. fat = subcutaneous fat, SD = standard deviation, Oc. Pressure = occlusion pressure, mmHg = millimeters of mercury.

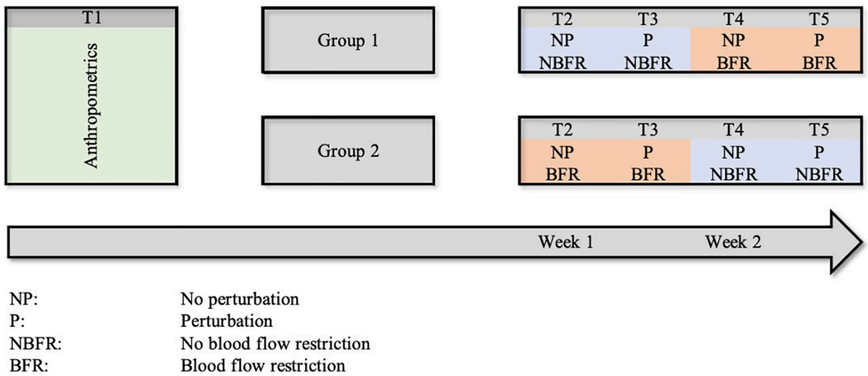


FIGURE 1 | Within subject design of the study.

goniometer The time subjects could remain in this position was measured and used as TTE. PC was determined by the extent of translational platform movement. Additionally, PC in response to external perturbations was operationalized by the time to regain stable stand.

Oxygen Saturation Monitoring and Occlusion Pressure

To assess O_2 saturation, near-infrared spectroscopy (NIRS) measurement was used as a manipulation check. Due to restricted instrument availability, saturation was measured on a random sample basis in 50% of subjects. Therefore, the NIRS group consisted of 10 persons (7 female, 3 male) with a mean weight of 65.99 kg (SD = 8.01) and 170 cm (SD = 7.24) height. The device (MOXY-3, MOXY, Hutchinson, MN, United States) was placed on the quadriceps femoris. To define individual NIRS placement, the distance from the proximal patella margin to the anterior superior iliac spine was measured. Measurements of NIRS and subcutaneous fat were then taken at one third of the length between these anatomic landmarks. Following the protocol of Laurentino et al. (2018), occlusion pressure was measured in a lying position on the artery femoralis using 10 cm cuffs (UT 1317-L, ulrich GmbH & Co. KG, Ulm, Germany). All data were recorded on the leg that subjects preferred to shoot a ball with. Within BFR conditions, subjects were asked to stand in an upright position when cuffs were inflated. Afterward, they stood still for 60 s to guarantee muscle deoxygenation before the beginning of PC measurement.

Measurement Setup

To monitor whether subjects remained in the defined body posture, a goniometer (BN-GON-110-XDCR, Biopac Systems Inc, Goleta, CA, United States) was placed on their knee (vastus lateralis of quadriceps femoris, distal to caput fibulae). *Via* biofeedback, subjects could control their position during measurement, because knee angle was projected (LV-7275, Canon, Tokyo, Japan) on to a screen in front of them (see **Figure 2**). The knee joint angle was defined as 110° (±2.5°) flexion angle between thigh and shank measured at the dorsal side of the respective leg. This angle results in an intermediate high position that is seen to provide a stable position and is therefore often adopted by athletes in, for example, speed skating (Foster et al., 1999; Müller and Schwameder, 2003). Subjects were placed on a posturometer (Zeptoring Deutschland GmbH, Berlin, Germany) that recorded displacements in the anterior–posterior (antpost) and medial–lateral (medlat) direction with a 1000 Hz sampling rate. This made it possible to register the course of the center of pressure on the support surface. In the dynamic PC conditions, the plate of the posturometer was pushed from medial to lateral pneumatically (4.6 bar) to disturb the subject's balance. The perturbations took place at 25, 50, 75, and 90% of the duration of the baseline measurement (NP) in the NBFR and BFR conditions of the corresponding participant. The timing of the perturbation was unknown for the subjects. All were told to maintain the position for as long as possible. Abortion criteria were either the loss of the target position three times within 10 s or a signal from the subjects that they could not hold the position any longer. Within the perturbation setting, same abortion criteria were used wherefore, in some cases (NBFR–75%: n = 2; 90%: n = 8; BFR – 90%: n = 5) subjects underwent less than 4 perturbations in their trial. If this was the case, only those perturbations which occurred before the measurement was stopped were considered for analysis. The duration from start to measurement abortion was measured and defined as TTE.

Data Processing

Data processing was carried out using MATLAB (R2018b, The MathWorks, Inc., Natick, MA, United States). As part of the data preprocessing, the raw data of the NIRS and the posturometer were first synchronized. Due to the different recording frequencies of the measuring instruments (MOXY 2 Hz, posturometer 1000 Hz), the NIRS data were subsequently linearly interpolated to 1000 Hz. Both, the antpost and the medlat data generated by the posturometer, were used to calculate the subjects' postural sway as the time-average of the instantaneous change in displacements in two directions. Therefore, the distance between two successive data points ($\text{antpost}_{T\,=\,i}$, $\text{antpost}_{t\,=\,i\,+\,1}$) in antpost and medlat direction was calculated and added over the entire course of the data. Since

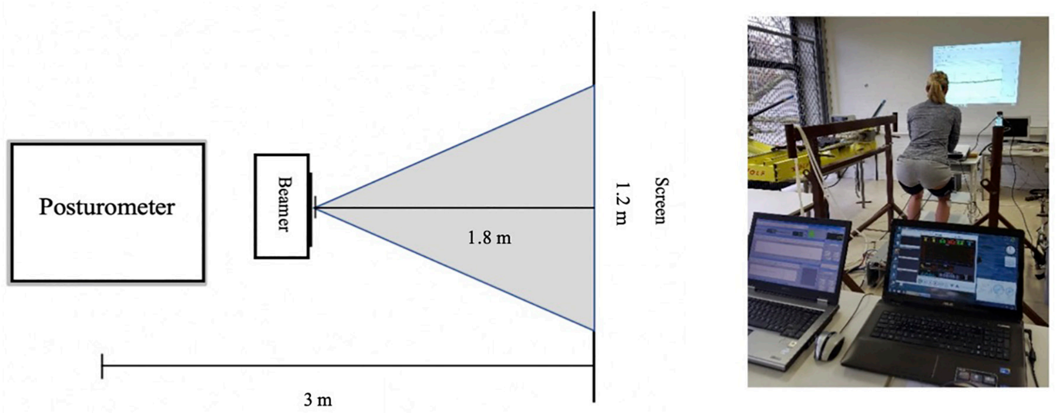


FIGURE 2 | Test setup. Left: bird's-eye view of the setup. Right: setup in the BFR condition.

the measurement duration varied, data had to be relativized over the test duration (n) using the formula below.

$$relative\ postural\ sway = \frac{\sum_{t=0}^{t=n} \sqrt{\left(antpost_{t=i+1} - antpost_{t=i}\right)^2 + \left(medlat_{t=i+1} - medlat_{t=i}\right)^2}}{n}$$

The restoration of postural control (referred to in the rest of the manuscript as re-stabilization) was defined as the time subjects needed from the start of perturbation until getting back to a "stable," bipedal stand. "Stable" stand was determined based on the subjects' time-average of postural sway in phase 2 of the NBFR NP condition because the plate displacement was the lowest in this phase. We determined the time, until the subjects get back to "stable" stand (phase 2 time-averaged postural sway $\pm$ 2*SD) as re-stabilization time.

Because one could assume that disturbance of the stable stand would have a short-term influence on O_2 saturation due to a shortly altered position, O_2 saturation was also calculated at the time of perturbation and shortly after (mean O_2 within 3 s after perturbation).

Near-infrared spectroscopy measurements under load are associated with a large dispersion of the measured values. To minimize the influence of any erroneous measurements, a moving average was calculated over 3,000 data points (3 s), whereby the mean value of an phase of 3,000 data points centered at the current time point was calculated continuously.

Statistics

SPSS (IBM SPSS Statistics, Version 24, Chicago, IL, United States) was used for the statistical analysis. Results were displayed using Jamovi (Jamovi project, Version 0.9, 2018) to illustrate mean values and 90% confidence intervals. Data were tested for normal distribution using the Shapiro–Wilk test and for violation of sphericity using Mauchly's test ($p > 0.05$). In line with the test design (within-subject) and hypotheses, variance analyses (multifactorial repeated-measures ANOVAs) were calculated to detect mean value differences between the conditions. First, relative postural sway was analyzed in general, using a 2 × 2 ANOVA with PC and O_2 saturation conditions being the factors. To further analyze relative postural sway during the course of the measurement a 2 × 5 ANOVA was used (PC conditions, measurement intervals). The segmentation into five intervals was made under consideration of the perturbation times and thus by the duration of the respective baseline conditions. Phase 1 comprised 0–20% of the duration of the NP setting; Phase 2, 20–40%; Phase 3, 40–60%; Phase 4, 60–80%; and Phase 5, 80–100%. This ensured that one perturbation was mapped in each of Phases 2–5. In a second step, re-stabilization time and O_2 saturation at the points of perturbation were analyzed using a 2 × 4 ANOVA (O_2 saturation conditions, points of perturbation). Also, oxygenation status before and directly after the point of perturbation was analyzed using a 2 × 2 × 4 ANOVA with O_2 saturation conditions, time point of NIRS measurement (pre, post perturbation) and points of perturbations. To analyze mean differences of TTE, a 2 × 2 ANOVA with PC and O_2 saturation conditions was conducted. Finally, as manipulation check, O_2 saturation was analyzed in PC and O_2 saturation conditions using a 2 × 2 ANOVA. As NIRS measurement was only conducted in $n = 10$ subjects, only those values are included into analysis where O_2 saturation was the outcome variable. Effect sizes of variance analyses are expressed as η_p^2. In case of significant interaction or main effects, *post hoc* testing was carried out with *t* tests using Bonferroni correction of the alpha level. Jamovi was used to create graphs and evaluate the *post hoc* tests in variance analyses with more than two factors. The significance level was set at $p < 0.05$ across all calculations. Results are all given in absolute numbers. Concerning O_2 saturation, the "%" marks only the unit of O_2 saturation, because this is usually displayed as a percentage. To avoid misinterpreting non-significant results, the absence of an effect was controlled with equivalence tests (Lakens, 2017). Paired-samples *t* tests were calculated as paired two one-sided tests (TOST-P) for the analysis of relative postural sway (Mara and Cribbie, 2012). The smallest effect size of interest was set at 0.46, which was the effect size of mediolateral sway reported by Barbieri et al. (2019) when investigating the effects of ankle muscle fatigue on postural sway.

RESULTS

Mauchly's test of sphericity indicated that the assumption of sphericity is met. There is no significant interaction between PC and O_2 saturation conditions [$F_{(1,19)} = 3.8$, $p = 0.07$, $\eta_p^2 = 0.17$]. Relative postural sway did not differ significantly between NBFR and BFR conditions [$F_{(1,19)} = 0.92$, $p = 0.35$, $\eta_p^2 = 0.05$]. However, the result is not equivalent to zero when calculating TOST-P with the defined effect size of interest [upper bound: $t(19) = -3.79$, $p < 0.01$, lower bound: $t(19) = 2.32$, $p = 0.38$; **Figure 3**]. As expected, relative postural sway was significantly lower in NP than in P conditions [$F_{(1,19)} = 41.1$, $p < 0.01$, $\eta_p^2 = 0.68$].

Concerning time effects in stable bipedal stand, the Greenhouse–Geisser adjustment was used to correct for violations of sphericity. There is a significant interaction [$F_{(4,76)} = 6.11$, $p = 0.06$, $\eta_p^2 = 0.24$] concerning PC conditions and measurement duration (**Figure 4**). While relative postural sway is lower in NBFR condition in the first intervals, differences in O_2 saturation increase during the course of the measurement with NBFR achieving higher values than BFR ($p_{bonf} = 0.01$, $dif_{mean} = -0.01$). There is a main effect of time, suggesting that the measurement duration influences relative postural sway values [$F_{(4,76)} = 33.19$, $p < 0.01$, $\eta_p^2 = 0.02$].

Analyzing the re-stabilization time, Mauchly's test of sphericity indicated that the assumption of sphericity is met. There is no interaction effect between the time of perturbation and the O_2 saturation conditions [$F_{(3,21)} = 0.52$, $p = 0.67$, $\eta_p^2 = 0.07$, **Figure 5**]. Also, there are no significant differences between NBFR and BFR conditions [$F_{(1,7)} = 0.04$, $p = 0.84$, $\eta_p^2 = 0.01$] and timepoints of perturbation [$F_{(3,21)} = 0.29$, $p = 0.83$, $\eta_p^2 = 0.04$]. Calculating TOST-P, the effects of perturbation one to three are equivalent to zero [p1: upper bound: $t(19) = -1.8$, $p = 0.04$, lower bound: $t(19) = 2.32$, $p = 0.02$, p2: upper bound: $t(19) = -2.03$, $p = 0.03$, lower bound: $t(19) = 2.09$, $p = 0.03$, p3: upper bound: $t(17) = -1.77$, $p = 0.05$, lower bound: $t(17) = 2.12$, $p = 0.02$]. Regarding perturbation four, results of TOST-P do not show equivalence [upper bound: $t(7) = -1.46$, $p = 0.09$, lower bound: $t(7) = 1.14$, $p = 0.15$]. The assumption that the effect is equivalent to zero can therefore not be proven for the last interval. Although the assumption of postural changes leading to short-time re-oxygenation cannot be confirmed [$F_{(1,6)} = 2.1$, $p = 0.2$, $\eta_p^2 = 0.26$], significant differences between the points of interruption were detected [$F_{(3,18)} = 5.02$, $p = 0.01$, $\eta_p^2 = 0.46$]. During the first perturbation O_2 saturation values are higher than during the last one ($p_{bonf} = 0.01$, $dif_{mean} = 9.33\%$).

Because O_2 saturation is assumed to be an important factor influencing muscular fatigue, TTE was analyzed. There is no violation of sphericity. As displayed in **Figure 6**, there are no interaction effects between PC and O_2 saturation conditions [$F_{(1,19)} = 0.24$, $p = 0.63$, $\eta_p^2 = 0.01$]. TTE is shorter in BFR than in NBFR conditions [$F_{(1,19)} = 16.22$, $p < 0.001$, $\eta_p^2 = 0.46$, $dif_{mean} = 73.7$ s]. Regarding PC conditions, subjects can maintain the position longer ($dif_{mean} = 31.4$ s) in the conditions where perturbation occur [$F_{(1,19)} = 7.28$, $p = 0.01$, $\eta_p^2 = 0.27$, dif_{mean} = of -31.4 s].

As a manipulation check, the saturation status of 10 subjects was analyzed. Mauchley's test of sphericity indicated that the assumption of sphericity is met. There is no significant interaction effect between PC and O_2 saturation conditions [$F_{(1,9)} = 2.95$, $p = 0.12$, $\eta_p^2 = 0.25$]. As expected, O_2 saturation is significantly higher in NBFR conditions ($dif_{mean} = 17.5\%$) than in BFR conditions [$F_{(1,9)} = 7.4$, $p = 0.02$, $\eta_p^2 = 0.45$]. PC conditions do not differ [$F_{(1,9)} = 0.52$, $p = 0.49$, $\eta_p^2 = 0.06$; **Supplementary Figure 1**].

DISCUSSION

This study aimed to investigate the ability to maintain and regain stability after external disturbance and to test whether this ability is changed by deoxygenation and the associated fatigue of the working muscle. As expected, the external pressure on the tissue caused by the application of pneumatic cuffs led to reduced O_2 saturation and premature fatigue defined as a reduced TTE. However, changes in the subjects' sway due to decreased O_2 saturation were not significant in static and perturbed-dynamic measurements.

A decreased TTE in the BFR conditions may be explained by the processes of muscular fatigue. Due to the lack of O_2, there is a failure in oxidative capacity, leading to a fast decline of force in type-I muscle fibers (Scott et al., 2014). The recruitment of fast-twitch fibers to maintain muscle function leads to an accelerated accumulation of metabolites (Loenneke et al., 2011), because the organism has to resort to other energy-producing metabolic processes to maintain the adenosine triphosphate (ATP) supply. Instead of mitochondrial ATP synthesis, anaerobic glycolysis becomes the main energy source. Due to this shift, phosphate accumulates that impairs the muscular work (for an overview, see Allen et al., 2008). Accordingly, the assumption that deoxygenation would lead to a decreased TTE could be confirmed in this study. Interestingly, TTE was significantly lower in the static condition than in the perturbed-dynamic condition.

There are different possible explanations for this observation: First, changes in position during the perturbation could have influenced O_2 saturation. Due to compensatory body movements, joint angles may have increased for a short time and thereby decreased contraction-induced pressure on the arterioles. Even though the main effect regarding O_2 saturation in PC conditions was not significant, differences between NBFR-NP and NBFR-P conditions could be seen, which supports this hypothesis. Second, it could be explained by the agonistic and antagonistic muscle activation. If the m. quadriceps femoris is the agonist for the isometric task, the antagonists (i.e., hamstrings) primarily need to work eccentrically to inhibit translation from medial to lateral (Blackburn et al., 2000). Within this mechanism, the quadriceps femoris does not have to perform extra work when

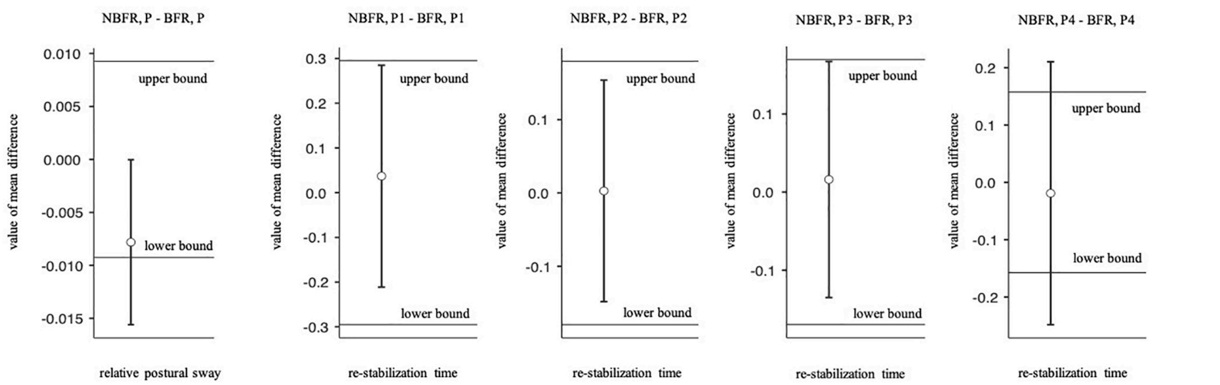
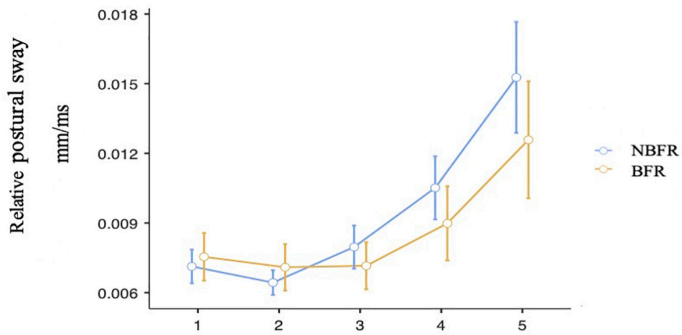


FIGURE 3 | Paired two one-sided *t* tests for equivalence (TOST-P). Boundaries and 90% CI of the relative postural sway and re-stabilization time. NBFR = No BFR, NP = No perturbation, P = Perturbation.



Mean (± *SD*) phase duration					
	1	2	3	4	5
NBFR, NP	43.91 ± 19.11	87.82 ± 38.21	131.73 ± 57.32	175.67 ± 76.42	219.55 ± 95.53
BFR, NP	28.12 ± 12.26	56.24 ± 24.52	84.36 ± 36.78	112.48 ± 49.04	140.59 ± 61.3

FIGURE 4 | Analysis of relative postural sway and measurement duration in the non-perturbated condition. Mean value comparison and 90% CI of the relative postural sway. NBFR = No BFR. *N* = 20 in all phases.

a perturbation occurs. Instead, there might be a short-term force release, resulting in a relaxation of the quadriceps femoris. In both explanations, the relaxation of the quadriceps would lead to a short-time blood and O_2 supply that might help to resist muscular fatigue. To investigate these assumptions, differences in O_2 saturation shortly before and after the perturbations were calculated. No changes concerning the NIRS measurement could be detected. This might be explained by the insufficient recording sensitivity and sampling frequency of the NIRS device used in this investigation. Additionally, it should be noted that only values from 10 subjects were included in the calculation, therefore the results should be regarded as an indication, not as proof. However, throughout the perturbations a deoxygenation could be seen, especially in the last phases. Whereas there were only small differences in O_2 saturation in the NBFR trials, the BFR conditions showed a trend toward decreasing saturation over time. This supports the hypothesis that re-oxygenation due to changed positions in the NBFR settings might have occurred. The fact that the perturbated NBFR conditions showed the highest O_2 saturation also argues in favor of this explanation. To obtain more robust answers regarding the underlying mechanisms, future investigations should enhance the control of the joint position. One option might be to include data from the goniometer. Additionally, the use of a NIRS device that allows for a higher sampling frequency is recommended.

A third explanation for the increasing TTE within the perturbation setting might be the mental distraction of the subjects due to the perturbations. The additional re-stabilization task might have led to an increase in motivation in those conditions. Additionally, a shift of focus might have occurred when subjects were trying not only to remain in one position but also to react quickly and regain stability after perturbation. Even though this was not controlled within this investigation, psychological or motivational effects could also be an explanation for the varying TTE between conditions.

As mentioned in the introduction, muscular fatigue is supposed to have a negative effect on PC due to the impairment of proprioception and kinesthesia. Previous data indicate that

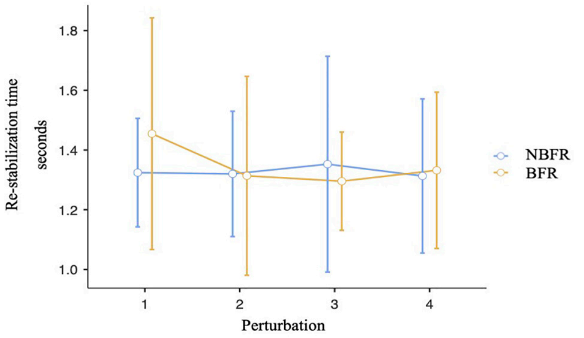


FIGURE 5 | Analysis of the re-stabilization time at the points of perturbation. Mean value comparison and 90% CI. NBFR = No BFR.

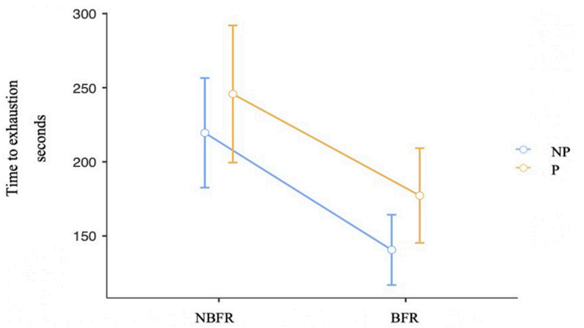


FIGURE 6 | Analysis of the TTE in PC and O_2 saturation conditions. Mean value comparison and 90% CI of the time to exhaustion in all conditions. NP = No perturbation, P = Perturbation, NBFR = No BFR.

fatigue alters the sensitivity of the muscle spindles, leads to a deterioration of afferent feedback, and reduces contraction force (Satas et al., 2020), resulting in a measurable decline of PC (Papa et al., 2015). In the present study, there was no significant change in relative postural sway between O_2 saturation conditions in the NP task. However, the results of the equivalence test show that the effect cannot be declared irrelevant. Reasons for these contradictory statements are that either the power of the study was too low to show existing effects with a 5% probability of error or the limits of the equivalence test were underestimated. Although the effect of fatigue on PC has already been investigated many times, effect sizes have rarely been reported, which makes it difficult to define boundaries.

Within each condition, relative postural sway increased during the course of the measurement. It should be noted here that the shortened TTE (e.g., in the BFR condition) must be taken into account when comparing the plate displacement over time. Interestingly, the relative postural sway was higher in the NBFR condition than in the BFR conditions. This phenomenon could be explained by different fatiguing processes. In the NBFR conditions an increased muscle tremor could be observed when muscular fatigue increased, which might be a reaction of the neuromuscular system to muscular work (Proske, 2019). However, the subjects were able to maintain the position despite the tremor for a long time in the NBFR. Although subjects reported that BFR was at first seen as less exhausting, the measurement was stopped in most of the cases before a tremor occurred. In this conditions fatigue and the failure to maintain posture started more suddenly and were sometimes accompanied by pain. This raises the question as to how far BFR influences somatosensory and pain perception due to the mechanical stimulation of, e.g., nerve bundles of effector afferences. Because peripheral nerves are considered to be particularly pressure-sensitive, further investigations could focus on the extent of the impairment by attaching BFR cuffs. It is conceivable that exceeding a somatosensory sensitivity threshold, which could result in an exacerbation of pain or fatigue perception, might be one explanation for an earlier and more sudden termination. As discussed before, the shift in focus due to cuff application and the resulting change in sensory feedback could be another reason for the late, but more rapid onset of fatigue. Because subjects were not accustomed to the application of BFR cuffs, their focus might have been more on the cuffs rather than on the biofeedback task. A longer familiarization within the test settings could help to control this parameter.

To analyze time effects, total times were subdivided into phases. However, because the measurement duration was lower in BFR conditions than in the corresponding NBFR conditions and subjects had more problems keeping their balance with increasing time, the comparability of the phases and the significance of the results is limited. Under BFR conditions, increased fatigue and the deterioration of PC are to be expected. In the context of this study, it is therefore comparable only directly up to the point at which the corresponding BFR conditions were aborted by the test subjects. However, in a situation without perturbations, deoxygenation does not seem to influence relative postural sway.

Because inferences from static to dynamic PC are of limited validity (Morrison and Sosnoff, 2010; Sibley et al., 2015), this study additionally investigated the response to external perturbation. A decline in the ability to restabilize was hypothesized in the BFR setting. However, significant differences between the O_2 saturation conditions weren't found. Additionally, exhaustion (operationalized by enhanced translatory plate movement) was not a factor influencing re-stabilization time. For the first three perturbations, this result could be confirmed by equivalence testing. Concerning the fourth perturbation, the equivalence test was non-significant. However, as some subjects could not remain the position until the last perturbation, the sample size of TOST-P is smaller concerning the last perturbation. There were only eight persons, who resisted all perturbations in both conditions. It is therefore important to emphasize that although the results can indicate tendencies, these should be examined further in order to be able to make distinctive statements. One reason for the absence of a decline in re-stabilization ability in this study might be the different activation patterns of the musculature within the static stand and the re-stabilization task. As described above, the disruption of PC changes the joint angles of the limbs. Therefore, muscle activation patterns are altering in the trunk and lower extremities, resulting in increased coactivation of muscles in these areas (Horak, 2006). Also, more motor units are recruited, and the innervation rate increases (Pollock et al., 2000). Whereas the quadriceps femoris primarily provides isometric strength in the stance phases, more muscle groups might be activated as a result of disruption, and this would relieve the quadriceps femoris. The muscular work applied to maintain a stable position against disruptions is primarily eccentric (Hofmann et al., 2001). In comparison to a stable position, PC disruption thus makes different requirements on the musculature. This might explain the non-significant change of re-stabilization time in fatiguing tasks. Moreover, it is conceivable that no additional O_2 consumption results from the perturbation, because the stabilizing muscles contract only briefly in the perturbation situation (average re-stabilization time < 1.4 s) and thus rely primarily on anaerobic metabolism for energy production—which acts independently from the O_2 supply. Thus, there would be no additional O_2 consumption due to the re-stabilization task. Another explanation could be the lack of muscular fatigue of the stabilizing musculature in the design of the study. Whereas an athlete's deep musculature has to work permanently for stabilization during a downhill run in alpine skiing, these muscles were activated less in this test setup because there were only four perturbations. The state of fatigue of the stabilizing musculature can therefore be considered as relatively low in this study. To intensify the impact, shortening the time phases between the perturbations could be one option, whereby it would have to be ensured that the test persons reach a stable state before further perturbation is triggered.

This work does reveal some limitations regarding the measurement setup and the data processing. Concerning the predefined position, it was notable that the hip angle varied depending on the flexibility of the ankle joint, thereby resulting in different hip angles. Thus, by using biofeedback, it was possible to standardize the knee angle but not the body position across subjects and settings. This might have influenced the manipulation of the subjects' blood flow. A further problem was the low acquisition rate of the NIRS that resulted in high variances. Although the reliability of the measurement technique seems to be given under such conditions, how far external factors such as muscle contractions influence the device is unclear. It would also be useful to determine the latency of the NIRS device and to include appropriate correction factors in the corresponding data sections (e.g., to investigate the effects of PC disruption more closely). Again, it should be noted, that within this study NIRS measurement was conducted on ten participants. An interpretation of the NIRS data should therefore be made cautiously. However, this study confirmed that BFR is a valid method to investigate the effects of muscle deoxygenation. Relative postural sway is not impaired by reduced O_2 saturation in the working muscle neither in static nor in perturbed-dynamic conditions.

CONCLUSION

Muscular deoxygenation during prolonged isometric muscle activity and the associated muscular fatigue influenced TTE, whereas static and dynamic postural control were not impaired directly by cuff application. In the dynamic PC conditions, TTE was not affected by reduced O_2 saturation in the BFR settings. Furthermore, due to the perturbation of the stable stand, TTE could be delayed in the BFR and NBFR test settings. This may be explained by a short-time load change as well as by a possible relief and reoxygenation through the change of body position. Regarding the PC, manipulation of O_2 saturation did not differ significantly between the static and dynamic settings. An increase in postural sway was observed during ongoing muscular fatigue in all conditions. If static PC is considered, this investigation confirms the assumption that stability deteriorates with decreased O_2 saturation, which is associated with muscular fatigue. This effect is more pronounced in BFR conditions than in NBFR conditions. Considering the re-stabilization task, even when the duration of the stable stand increased and muscular fatigue was greater, no change in the re-stabilization time could be shown. This leads to the assumption that muscular deoxygenation does not affect performance within a non-anticipated re-stabilization task.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Local Ethics Committee of the Goethe University Frankfurt (Chair: Klein, Grant Number: 2018-34). The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

All authors prepared the set up together. CW carried out the measurements and data analyses, and also wrote the manuscript. MB and KZ helped interpreting the data and checked the manuscript drafts several times.

SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fnhum.2021.756230/full#supplementary-material

Supplementary Figure 1 | Analysis of the O_2 saturation in PC and O_2 saturation conditions. Mean value comparison and 90% CI. NP = No perturbation, P = Perturbation, NBFR = No BFR.

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Postural Balance in Individuals With Knee Osteoarthritis During Stand-to-Sit Task

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Objective: Stand-to-sit task is an important daily function, but there is a lack of research evidence on whether knee osteoarthritis (knee OA) affects the postural balance during the task. This study aimed to compare individuals with knee OA and asymptomatic controls in postural balance and identify kinematic and lower extremity muscle activity characteristics in individuals with knee OA during the stand-to-sit task.

Methods: In total, 30 individuals with knee OA and 30 age-matched asymptomatic controls performed the 30-s Chair Stand Test (30sCST) at self-selected speeds. Motion analysis data and surface electromyography (sEMG) were collected while participants performed the 30sCST. To quantify postural balance, the displacement of the center of mass (CoM) and the peak instantaneous velocity of the CoM were calculated. The kinematic data included forward lean angles of the trunk and pelvic, range of motion (RoM) of the hip, knee, and ankle joints in the sagittal plane. The averaged activation levels of gluteus maximus, vastus lateralis, vastus medialis, rectus femoris, biceps femoris (BF), tibialis anterior (TA), and medial head of gastrocnemius muscles were indicated by the normalized root mean square amplitudes.

Results: Compared with the asymptomatic control group, the knee OA group prolonged the duration of the stand-to-sit task, demonstrated significantly larger CoM displacement and peak instantaneous CoM velocity in the anterior-posterior direction, reduced ankle dorsiflexion RoM, greater anterior pelvic tilt RoM, and lower quadriceps femoris and muscles activation level coupled with higher BF muscle activation level during the stand-to-sit task.

Conclusion: This study indicates that individuals with knee OA adopt greater pelvic forward lean RoM and higher BF muscle activation level during the stand-to-sit task. However, these individuals exist greater CoM excursion in the anterior-posterior direction and take more time to complete the task. This daily functional activity should be added to the rehabilitation goals for individuals with knee OA. The knee OA group performs reduced ankle dorsiflexion RoM, quadriceps femoris, and TA activation deficit. In the future, the rehabilitation programs targeting these impairments could be beneficial for restoring the functional transfer in individuals with knee OA.

Keywords: knee osteoarthritis, stand-to-sit, postural balance, motion analysis, electromyography

INTRODUCTION

Knee osteoarthritis (knee OA) is the most common degenerative joint disease, affecting an estimated 18% population in China (Wang et al., 2018). The disease is associated with pain, joint stiffness, quadriceps weakness, instability, and functional disability (Hunter and Bierma-Zeinstra, 2019).

In daily life, walking function is the basic activity, while sit-to-stand and stand-to-sit tasks can be the prerequisite and termination of gait, respectively. As people get older, sit and stand transition becomes a more demanding functional daily task (Galan-Mercant and Cuesta-Vargas, 2013). In fact, sit-to-stand or stand-to-sit motion variability has been proved to be significantly correlated with the risk of falling (Ghahramani et al., 2020). People suffering from knee OA have a higher prevalence of falls compared to non-OA subjects (Deng et al., 2021). Therefore, analyzing sit and stand transition and developing targeted rehabilitation plans can help individuals with knee OA perform the abovementioned tasks and reduce fall risk.

Sit-to-stand and stand-to-sit are sagittal planes dominant tasks. At present, the biomechanical characteristics of the sit-to-stand task such as kinetics, kinematics, and electromyography in the sagittal plane have been studied extensively in individuals with knee OA (Sonoo et al., 2019). The meta-analysis showed that individuals with knee OA tend to stand up with a lower knee extension moment during the sit-to-stand task (Sonoo et al., 2019). Moreover, individuals with knee OA demonstrated larger trunk flexion angle and forward center of mass (CoM) displacement (Naili et al., 2018; Sonoo et al., 2019). In neuromuscular activation level, individuals with knee OA activate more type II fibers of rectus femoris (RF) or increase the muscle activity of hamstrings (Bouchouras et al., 2015; Anan et al., 2016). These abovementioned biomechanical alterations indicate individuals with knee OA cannot perform the sit-to-stand task efficiently.

However, fewer studies have considered the stand-to-sit task. Unlike sit-to-stand, stand-to-sit is directly linked to the opposite movement and different muscle activation patterns (Ashford and De Souza, 2000). The stand-to-sit task requires almost simultaneous control of the anterior-posterior and vertical displacement of body CoM against gravity (Kerr et al., 1997). Trunk anteflexion and ankle dorsiflexion (Nakagawa and Petersen, 2018) play important roles during the stand-to-sit task to control the CoM backward and downward. Previous research reported women with knee OA showed smaller ankle dorsiflexion angles during the stand-to-sit task (Wu et al., 2015). Moreover, the eccentric contraction of the knee and hip extensors is essential in slowing down the movement velocity to enable a stable and safe landing to the seat (Ferrante et al., 2005). Another study that found women with knee OA exists weaker vastus lateralis (VL) activation combined with reduced knee flexion range of motion (RoM) during the stand-to-sit task (Bouchouras et al., 2020). These abovementioned motor changes and muscle activation alternations may bring challenges to individuals with knee OA with the increased risk to fall back to the seat.

Proper balance is essential to perform the stand-to-sit task and prevent high impact forces during seat contact that would lead to increased impact to the spine (Chen et al., 2010; Sibley et al., 2013). Thus, the importance of achieving movement control during the stand-to-sit task should not be underestimated for individuals with knee OA. In addition, there is still a lack of evidence to explore trunk motion and hip or ankle muscle activity in individuals with knee OA during the stand-to-sit task. Illustrating the postural balance and kinematic and muscle activity characteristics during the stand-to-sit task in individuals with knee OA is essential for customizing rehabilitation goals and restoring functional transfer.

The 30-s Chair Stand Test (30sCST) is one of the five physical function tests recommended for people with knee OA by The Osteoarthritis Research Society International (OARSI; Dobson et al., 2013). Compared to the five-repetition stand-to-sit test, performing as many stand-to-sit repetitions as possible during 30sCST is easier to fully capture the impaired postural balance and biomechanical alterations.

Therefore, the purpose of this study was to investigate the impact of knee OA on postural balance and identify kinematic and lower extremity muscle activity characteristics in individuals with knee OA during the descending phase of 30sCST. The primary outcome of this study was CoM displacement and velocity. The secondary outcomes included the duration of the stand-to-sit task, segment RoM, and lower extremity muscle activation level. We hypothesized that the individual with knee OA would display larger CoM displacement or velocity, longer task duration together with different movement strategies, and lower extremity muscle activity alternations during the task (Wu et al., 2015; Bouchouras et al., 2020).

MATERIALS AND METHODS

Participants

Individuals with unilateral or bilateral mild-moderate (II or III Kellgren/Lawrence (K/L) grade) knee OA were the focus group in this study. To identify the performance variation, a control group with age-matched asymptomatic individuals was included in this study. By referring to similar research (Bouchouras et al., 2020), we used the power of 0.8, the effect size (ES) of 0.75, and two-sided $\alpha = 0.05$ to calculate the sample size. G × Power software (version 3.1.9.2, Franz Faul, University of Kiel) showed that a minimum number of 29 participants per group should be obtained. Finally, 30 participants for each group were recruited from the neighboring communities of the Fujian University of Traditional Chinese Medicine (FJTCM) *via* advertisements in print/radio/social media.

The following inclusion criteria were set for the individuals with knee OA: fulfilled with the clinical diagnosis of 2018 Diagnosis and Treatment of Osteoarthritis (Osteoporosis Group of Chinese Orthopedic Association, 2018) and had II or III K/L grade. The K/L grade of individuals with knee OA was defined by anterior X-ray images of identified osteophytes and narrowing of the joint space in a standing position (Ribeiro et al., 2020). The inclusion criteria of the control group were age-matched people without knee OA-related symptoms and any other conditions that would affect walking and postural balance. Participants of both groups were able to accomplish sit and stand transitions without assistive devices. Participants were excluded if they had other lower extremity joint pain, severe back pain, rheumatoid arthritis, fractures, neurological system pathology, or obesity (body mass index (BMI) > 28 kg/m^2) (Li et al., 2020).

The experiment protocol was approved by the Ethics Committee of the Affiliated Rehabilitation Hospital of FJTCM (#2018KY-006-1) and registered on the Chinese Clinical Trial Registry website (identifier number ChiCTR1800018028)[1]. All participants were informed about the study protocol as well as potential benefits and risks and provided written and oral consent prior to the experiment.

Data Collection

Lower extremity muscle activities were measured with a wireless surface electromyography (sEMG) system (Trigno Wireless EMG System, Delsys Inc., Natick, MA, United States) at a sampling frequency of 2,000 Hz and a band-pass filter of 20–450 Hz. Skin preparation and location of the electrodes followed the recommendations of sEMG for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000). NuPrep skin preparation gel is beneficial for use where motion artifacts can affect readings, and when a reduction of skin impedance would enhance a test result. Adhesive pre-gelled Ag/AgCl electrodes (Trigno Avanti Sensor, Delsys, United States) were placed bilaterally on the gluteus maximus (GMAX), VL, vastus medialis (VM), RF, biceps femoris (BF), tibialis anterior (TA), and medial head of gastrocnemius (MG) muscles.

After the placement of electrodes, three amplitude normalization tests were performed for each investigated muscle separately to direct quantitative comparison of sEMG data between participants. Before the normalization tests, each participant performed the initial warming up sequence (stretching, 5 min). During each muscle normalization test, participants followed visual (looking at the real-time sEMG curve on the screen) and verbal stimulation, slowly started increasing the force, reached the maximum effort, and held it for 3 s, and promptly relaxed (Anan et al., 2016). Each muscle repeated the normalization test three times with a pausing period of 30–60 s in between. For VL/VM/RF/BF/TA, the normalization tests adopted the gold standard that is the maximum voluntary isometric contraction (MVC) test. The MVC tests were measured at the muscle strength test system (Myonline Professional, DIERS International GmbH, Germany). The starting position was standardized with the participants seated on the device with the pelvis as close as possible to the backrest. The lower legs were set between the two leg extension/flexion pads, and the participant was secured firmly using the pelvic/hip strap and the thigh strap. Then, each participant performed maximal knee extension, knee flexion, and ankle dorsiflexion against the rear pad around the ankle joint successively (**Figure 1A**). When one leg was measured, the other leg was supported in a relaxed position. Due to the fact that the Myonline equipment could not complete the GMAX and MG MVC tests, the GMAX and MG normalization tests were replaced by the isometric contraction against gravity in a standing position. The alternative normalization test also has good reliability (Burden, 2010). For GMAX, the participants slowly extended one hip joint to the highest height as possible with the upper body upright (**Figure 1B**). For MG, the participants performed as follows: one lower limb was tiptoe while the other side was off the ground (**Figure 1C**).

After the normalization tests, 75 retroreflective markers were placed on the anatomical landmarks and top of each segment as tracking markers (**Table 1** and **Figure 2**). The placement of the marker was according to the calibrated anatomical systems technique protocol (Cappozzo et al., 1995) to form a 15-segment whole-body model. The kinematic data were collected by a 3D motion capture system equipped with a 10-camera setup (Oqus 7+, Qualisys AB, Sweden) at a sampling rate of 100 Hz. After the placement of the reflective markers, a static standing trial was recorded to create a model of the participant in Visual 3D.

Participants were instructed to perform 30sCST while wearing sports shoes (112027711-4/122025523R-2, Anta Co. Ltd., China) (**Figure 3**). Each participant performed the 30sCST under the following instructions: (1) started from the seated position, the feet were allowed to be placed flat on the floor and shoulder-width apart, arms crossed on chest, stood completely up, and then sat completely back down and (2) rising at the natural speed of the participant and as fast as possible during 30 s. An armless chair with a standardized seat height of approximately 43 cm (17-inch) was used according to the OARSI (Dobson et al., 2013). The seat was placed on an anti-slip surface. This process was performed with each leg on one force plate. Ground reaction forces (GRFs) were simultaneously measured using force plates with a sampling rate of 2,000 Hz (9260AA, Kistler Ltd., Switzerland). Some practice trials were performed prior to the test by all participants to familiarize themselves with the 30sCST.

Data Analysis

All data processing and outcome calculations were in Visual3D (V6, C-motion Inc., Germantown, MD, United States). The marker data were filtered using a fourth-order low-pass Butterworth filter with a cutoff frequency of 6 Hz (Robertson and Dowling, 2003). The GRF raw data were filtered with a

[1]http://www.chictr.org.cn

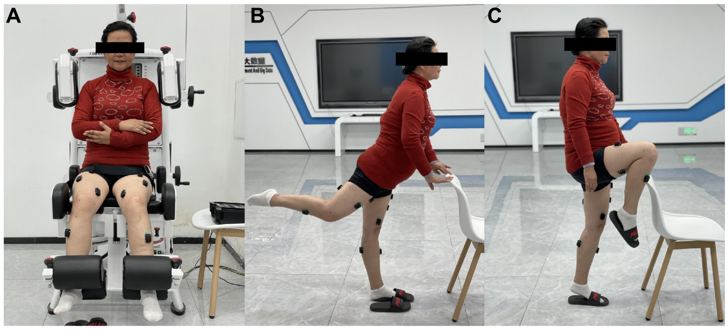

FIGURE 1 | Surface electromyography (sEMG) amplitude normalization tests: **(A)** for vastus lateralis/vastus medialis/rectus femoris/biceps femoris/tibialis anterior (VL/VM/RF/BF/TA), **(B)** for gluteus maximus (GMAX), and **(C)** for medial head of gastrocnemius (MG).

fourth-order low-pass Butterworth filter with a cutoff frequency of 20 Hz (Piano et al., 2020). Both primary and secondary outcomes, i.e., the CoM displacement, peak velocity and the stand-to-sit task time, segment RoM, and lower extremity muscle activation level, were calculated based on the filtered data.

TABLE 1 | Seventy-five retroreflective markers placement.

Marker name	Marker location
Upper body	
L/R_HEAD	Just above the ear
SGL	Glabella
CLAV	Clavicular notch
STRN	Sternum
CV7	7th Cervical Vertebrae
TV10	10th Thoracic Vertebrae
L/R_SIA	Scapula-Inferior Angle
L/R_SAE	Scapula-Acromial Edge
L/R_ASH	Anterior shoulder
L/R_PSH	Posterior shoulder
L/R_1-3 Cluster	Cluster of three markers placed on the lateral surface of the upper arm
L/R_HLE	Humerus – Lateral Epicondyle
L/R_HME	Humerus – Medial Epicondyle
L/R_1-3 Cluster	Cluster of three markers placed on the lateral surface of the forearm
L/R_RSP	Radius – Styloid Process
L/R_USP	Ulna – Styloid Process
L/R_HM2	Basis of Forefinger
Lower body	
L/R_IAS	Anterior superior iliac spine
L/R_IPS	Posterior superior iliac spine
L/R_TH1-4 Cluster	Cluster of four markers placed on the lateral surface of the thigh
L/R_FLE	Lateral epicondyle
L/R_FME	Medial epicondyle
L/R_TT	Tuberositas tibiae
L/R_SK1-4 Cluster	Cluster of four markers placed on the lateral surface of the shank
L/R_FAL	Lateral prominence of the lateral malleolus
L/R_TAM	Medial prominence of the medial malleolus
L/R_FCC	Aspect of the Achilles tendon insertion on the calcaneus
L/R_FM1	Dorsal margin of the first metatarsal head
L/R_FM2	Dorsal aspect of the second metatarsal head
L/R_FM5	Dorsal margin of the fifth metatarsal head

Segment coordination systems of the trunk, pelvis, both thighs, shanks, and feet were defined based on the anatomical markers (Robertson et al., 2013). Hip, knee, and ankle joint angles were defined as the angle between proximal and distal segments. Trunk segment angle and pelvic segment angle were determined with respect to the laboratory coordinate system (Jeon et al., 2021). Joint angles were calculated with a Cardan x–y–z (mediolateral, anteroposterior, and transverse) rotation sequence (Cole et al., 1993). The forward lean RoM of the trunk and pelvic and the RoM of the hip, knee, and ankle joints in the sagittal plane were calculated using Visual3D.

The CoM was calculated using the weighted average of all the segments of the body according to the study by Robertson et al. (2013). The peak-to-peak displacement of CoM and the peak instantaneous velocity of the CoM in anterior-posterior and vertical directions were used to quantify the body oscillation during dynamic functional tasks (Hsue and Su, 2014). An increased value for either variable suggests a decreased ability to maintain balance (Hsue and Su, 2014).

Motion initiation was defined as the first transition from negative to positive trunk angular velocity after the occurrence of the maximum knee extension angle (Anan et al., 2015). Motion termination was defined as the instant vertical vector of the GRF less than 10 N (Anan et al., 2015). The duration of the stand-to-sit task was the time interval between the movement initiation and termination.

The sEMG signals of the normalization tests and the stand-to-sit task were full-wave rectified and enveloped with a root mean square (RMS) algorithm with a 50-ms window (Anan et al., 2016). Each sEMG signal during the stand-to-sit task was normalized to the corresponding peak value of three normalization tests. The RMS value of each normalized sEMG signal was calculated during the stand-to-sit task to quantify the magnitude of the muscle excitation.

Each index was represented using the mean value of many stand-to-sit transitions during the stable period (10–25 s) of 30sCST. For lower extremity muscle activity and joint RoM, we compared the more affected leg of the knee OA group and the dominant side leg of the control group. The dominant side

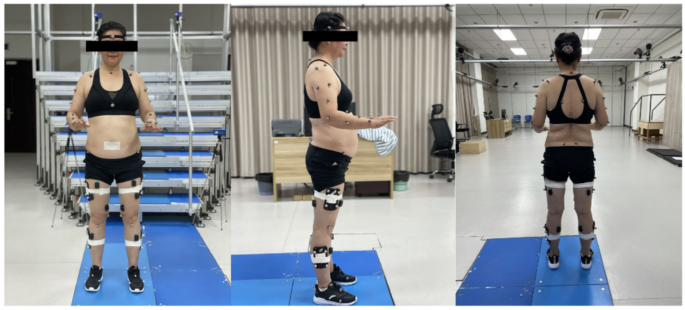

FIGURE 2 | The placement of the markers.

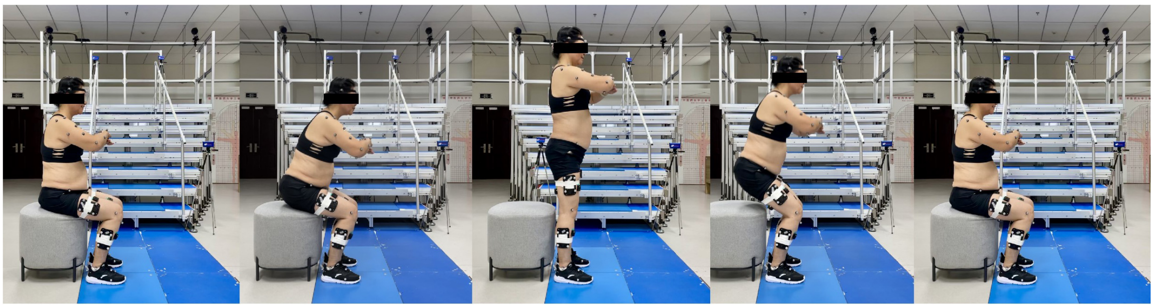

FIGURE 3 | The 30-s Chair Stand Test (30sCST).

leg is defined as the preferred limb when kicking a ball (van Melick et al., 2017). If bilateral symptomatic individuals with knee OA have similar knee pain on both sides, we would choose the dominant limb.

Statistics

All values are presented as mean ± SD. Prior to all analyses, the normality of the quantitative data was assessed using the Shapiro–Wilk test. The two independent samples *t*-test was used to compare continuous normally distributed variables, i.e., age, height, body mass, BMI, CoM parameters, segment RoM, and BF muscle activation level. The Wilcoxon Mann–Whitney *U* test was used to compare the non-normal variables, i.e., stand-to-sit time and muscle activation level except for BF. A chi-square test was used to compare the qualitative data, i.e., gender. IBM SPSS version 25.0 (SPSS Inc., Chicago, IL, United States) was used for all statistical analyses. The significance level was set at less than 0.05. To determine the magnitude of difference between the two groups, ES calculations (Cohen's *d* for quantitative data and Cramer's φ for qualitative data) were reported for all measures. An ES from 0.1 to 0.3 was regarded as a small effect, 0.3–0.5 as intermediate, >0.5 as a strong effect (Cohen, 2013).

RESULTS

Participants

In total, 60 participants completed the study (knee OA group: $n = 30$; control group: $n = 30$). There was no significant difference between the groups for age (knee OA group: 58.63 ± 5.67 vs. control group: 59.33 ± 5.14 years, $P = 0.618$, ES = 0.129), height (1.60 ± 0.56 vs. 1.60 ± 0.67 m, $P = 0.900$, ES = 0), body mass (59.13 ± 7.56 vs. 59.00 ± 9.63 kg, $P = 0.952$, ES = 0.015), BMI (23.08 ± 2.54 vs. 22.85 ± 2.41 kg/m^2, $P = 0.721$, ES = 0.096), and the male/female ratio (4/26 vs. 8/22, $P = 0.197$, ES = 0.167; **Table 2**).

Stand-to-Sit Task Time

The stand-to-sit task time in the knee OA group and control group was 0.95 ± 0.15 and 0.81 ± 0.20 s, respectively. The knee OA group showed a statistically significant longer task time ($P < 0.001$, ES = 0.849; **Table 3**).

TABLE 2 | Characteristics of the knee osteoarthritis (OA) and control groups.

	knee OA group (n = 30)	Control group (n = 30)	*P*-value	Effect size
Age	58.63 ± 5.67	59.33 ± 5.14	0.618	0.129
Height (m)	1.60 ± 0.56	1.60 ± 0.67	0.900	0
Body mass (kg)	59.13 ± 7.56	59.00 ± 9.63	0.952	0.015
BMI (kg/m^2)	23.08 ± 2.54	22.85 ± 2.41	0.721	0.096
Gender (male/female)	4/26	8/22	0.197	0.167
K/L grade (II/III)	23/7	/	/	/
Course of disease (month)	93.10 ± 86.74	/	/	/

Mean ± SD.
BMI, body mass index; K/L, Kellgren/Lawrence.

Center of Mass Parameters

In the anterior-posterior direction, the results from the study demonstrated that the CoM displacement of the knee OA group was 0.03 m larger than that of the control group (0.19 ± 0.04 vs. 0.16 ± 0.04 m, $P = 0.002$, ES = 0.750) during the stand-to-sit task. Also, the peak instantaneous velocity of CoM in the anterior-posterior direction of the knee OA group was 0.04 m/s higher than that of the control group (0.38 ± 0.07 vs. 0.34 ± 0.07 m/s, $P = 0.029$, ES = 0.571). Whereas in the vertical direction, there was no statistically significant difference of CoM displacement and peak instantaneous velocity between the two groups ($P > 0.05$; **Table 3**).

Segment Range of Motion

The pelvic forward lean RoM in the knee OA group was significantly larger than that in the control group (21.99° ± 4.75° vs. 18.32° ± 5.1°, $P = 0.005$, ES = 0.744). In addition, the knee OA group presented smaller ankle dorsiflexion RoM (11.80° ± 5.35° vs. 15.10° ± 4.75°, $P = 0.014$, ES = −0.652). The trunk forward lean RoM and hip and knee flexion RoM showed no statistically significant difference between the two groups ($P > 0.05$; **Table 3**).

Muscle Activation Level

There were smaller RMS values of VL, VM, RF, and TA muscles in the knee OA group compared to the control group (VL: 13.11 ± 5.13 vs. 17.78 ± 8.09%, $P = 0.032$, ES = −0.689; VM: 14.96 ± 5.68 vs. 20.04 ± 9.57%, $P = 0.035$, ES = −0.646; RF: 8.93 ± 4.69 vs. 12.87 ± 7.15%, $P = 0.022$, ES = −0.652; TA: 7.59 ± 4.67 vs. 10.18 ± 4.71%, $P = 0.017$, ES = −0.552). Meanwhile, BF muscle in the knee OA group showed larger RMS value than that in the control group (5.01 ± 2.50 vs. 3.81 ± 1.69%, $P = 0.034$, ES = 0.562). There was no statistically significant difference in the RMS values of GMAX and MG between the two groups ($P > 0.05$; **Table 3**).

TABLE 3 | Data between the knee OA and control groups.

	knee OA group (n = 30)	Control group (n = 30)	P-value	Effect size
Stand-to-sit time (s)	0.96 ± 0.15	0.81 ± 0.20	**<0.001**	**0.849**
CoM parameters				
$d_{CoM,AP}$ (m)	0.19 ± 0.04	0.16 ± 0.04	**0.002**	**0.750**
$d_{CoM,vertical}$ (m)	0.24 ± 0.03	0.24 ± 0.03	0.800	0
$V_{CoM,AP}$ (m/s)	0.38 ± 0.07	0.34 ± 0.07	**0.029**	**0.571**
$V_{CoM,vertical}$ (m/s)	0.54 ± 0.10	0.58 ± 0.11	0.172	−0.381
RoM (°) in sagittal plane				
Trunk	28.49 ± 7.64	25.05 ± 5.72	0.053	0.510
Pelvic	21.99 ± 4.75	18.32 ± 5.11	**0.005**	**0.744**
Hip	74.56 ± 11.54	69.67 ± 7.35	0.056	0.505
Knee	80.57 ± 10.12	80.29 ± 8.86	0.910	0.029
Ankle	11.80 ± 5.35	15.10 ± 4.75	**0.014**	**−0.652**
RMS (%) of muscle				
GMAX	4.03 ± 2.64	2.89 ± 1.34	0.065	0.545
VL	13.11 ± 5.13	17.78 ± 8.09	**0.032**	**−0.689**
VM	14.96 ± 5.68	20.04 ± 9.57	**0.035**	**−0.646**
RF	8.93 ± 4.69	12.87 ± 7.15	**0.022**	**−0.652**
BF	5.01 ± 2.50	3.81 ± 1.69	**0.034**	**0.562**
TA	7.59 ± 4.67	10.18 ± 4.71	**0.017**	**−0.552**
MG	3.56 ± 2.25	3.29 ± 2.87	0.329	0.105

Mean ± SD.
CoM, center of mas; $d_{CoM,AP}$, displacement of CoM in anterior-posterior direction; $d_{CoM,vertical}$, displacement of CoM in vertical direction; $V_{CoM,AP}$, velocity of CoM in anterior and posterior direction; $V_{CoM,vertical}$, velocity of CoM in vertical direction; RMS, root mean square; GMAX, gluteus maximus; VL, vastus lateralis; VM, vastus medialis; RF, rectus femoris; BF, biceps femoris; TA, tibialis anterior; MG, medial head of gastrocnemius. The bolded values mean the P-value is smaller than 0.05.

DISCUSSION

This study aimed to investigate the influence of knee OA on postural balance and investigate the differences in the measures of the trunk, pelvic, lower extremity kinematics, and lower extremity muscle activity between the knee OA group and the control group during the stand-to-sit task. We found that individuals with knee OA showed greater postural sway and prolonged duration of the stand-to-sit task, reduced ankle dorsiflexion RoM, quadriceps femoris, and TA activation level during the stand-to-sit task in comparison with the control group. At the same time, individuals with knee OA may increase pelvic anterior tilt RoM and BF muscle activity to functional compensation than the control group during the task.

The meta-analysis showed that individuals with knee OA had significantly longer sit-to-stand times (Sonoo et al., 2019). Longer task time is associated with limited physical function (Segal et al., 2013). However, few studies reported the duration of the stand-to-sit task in individuals with knee OA. It was previously reported that there was no statistically significant difference in task duration between women with knee OA and healthy subjects during three sittings (Bouchouras et al., 2020). In our study, results showed that individuals with knee OA took more time to accomplish the stand-to-sit task. The ability to perform the stand-to-sit task is influenced by knee OA disease. 30sCST seemed to be challenging enough to capture the impaired function in individuals with knee OA compared with the three-repetition stand-to-sit task.

With regard to postural stability, the results demonstrated that the knee OA group had greater CoM displacement and peak instantaneous velocity in the anterior-posterior direction, which could be an indication that individuals with knee OA would have a greater risk to fall backward. Some researchers have reported that individuals with knee OA showed impaired balance in other daily activities such as standing (Truszczyńska-Baszak et al., 2020), walking (Graber et al., 2021), and stair descending (Koyama et al., 2015). Poor balance is related to muscle weakness in individuals with knee OA (Bennell et al., 2011). The stand-to-sit task is performed with an eccentric contraction of the knee and hip extensors to slow down the movement velocity (Ferrante et al., 2005). However, due to disuse atrophy and reflex inhibition caused possibly by pain, knee OA would result in deficits in the voluntary activation of the quadriceps femoris (Kittelson et al., 2014). Our results demonstrated VL, VM, and RF activation deficit in individuals with knee OA during the stand-to-sit task, which conformed to the findings from previous research (Bouchouras et al., 2020). Quadriceps eccentric contraction exercise may need to be addressed for individuals with knee OA to improve physical balance.

Moreover, the results from this study showed that the knee OA group demonstrated reduced ankle dorsiflexion RoM and lower muscle activity of TA. The result of reduced ankle dorsiflexion RoM in our study was similar to the result of a previous study (knee OA: 13.2° ± 6.3° vs. control: 15.8° ± 5.2°) (Wu et al., 2015). The dynamic balance could be influenced by the change in the ankle movement during weight-bearing activities. It was reported that the reduced ankle dorsiflexion RoM was correlated with instability along the anterior-posterior direction and would affect the ability to lower the CoM of the body (Nakagawa and Petersen, 2018). In addition, limitations in ankle dorsiflexion showed that it could result in knee abnormal alignment and increase the risk for knee joint pathology (Basnett et al., 2013; Lima et al., 2018). TA muscle is the active muscle that produces ankle dorsiflexion, which had been rarely studied in individuals with knee OA before this study. A previous study concluded that knee OA would lead to a decrease in TA muscle contractile tissue (Taniguchi et al., 2015), which may influence TA muscle activation. Our study demonstrated TA muscle activation deficit in individuals with knee OA. Future studies should be performed to determine whether interventions directed at improving ankle dorsiflexion RoM and TA muscle activation would have an effect on postural balance.

The human body generally takes compensation strategies to maintain equilibrium when postural sway happens during daily activity. Trunk and pelvic anteflexion would control the backward movement of the CoM (Takeda, 2012; Darwish et al., 2019). During the sit-to-stand task, individuals with knee OA were found to have an increased trunk flexion angle to move CoM forward (Sonoo et al., 2019). In our study, we found that individuals with knee OA adopted another strategy, which was reflected as a greater pelvic anterior tilt angle during the stand-to-sit task. Greater pelvic anterior tilt could keep the CoM within the base of support, and this enables the CoM to retain longer in the support area throughout the task to reduce the risk of falling back to the seat (Darwish et al., 2019). In contrast, this strategy characterizes a method to reduce the quadriceps demand (Goncalves et al., 2017).

Increasing the BF muscle activity is a common appearance in individuals with knee OA during daily activities (Mills et al., 2013). The BF muscle activation provided the additional force to balance and stabilize the knee joint (Mills et al., 2013). Higher BF muscle activation level in our study could be the strategy to compensate quadriceps activation deficit, but this strategy would result in higher energetic costs or joint load (Hortobágyi and DeVita, 2000; Patsika et al., 2011). BF muscle is a two-joint muscle, originated from ischial tuberosity to the lateral aspect of the fibular head. From the perspective of the BF muscle anatomy, the other alternative explanation is that the greater pelvic anterior tilt results in the lengthening of the BF that leads to higher muscle activity.

Limitation

First, the height of the chair was not adjusted to the lower leg length of the participants. We used a chair with a standardized seat height of approximately 17-inch according to the OARSI that has the tremendous advantage to reflect the real-life situation for elderly people. However, a previous study reported that chair seat height in relation to the lower leg length should be considered when interpreting 30sCST performance (Kuo, 2013). Second, we ignored unilateral/bilateral symptoms and the movements that may occur in the frontal plane, which may further explain impairments in postural balance. Moreover, there was unequal men/women representation, and the results of our study may not be applicable to men. Whether there are different performances between men and women remain to be studied. Finally, this study was a cross-sectional design, it may be hard to conclude the cause-and-effect relationships. Further research should evaluate the influence of rehabilitation on improving ankle dorsiflexion kinematic and lower extremity muscle activity in relation to postural balance.

CONCLUSION

In our study, individuals with knee OA adopt greater pelvic forward lean RoM and higher BF muscle activation levels during the stand-to-sit task. However, these individuals still demonstrated greater CoM excursion in the anterior-posterior direction and took more time to complete the task. Knee OA leads to postural instability and functional disability during the stand-to-sit task. This daily functional activity should be added to the rehabilitation goals for individuals with knee OA. The healthcare professional should recommend that individuals with knee OA use an armrest or handrail to reduce the risk of falls during the stand-to-sit task. Our findings demonstrated that individuals with knee OA performed reduced ankle dorsiflexion RoM, quadriceps femoris, and TA activation deficit. The rehabilitation programs targeting these impairments could be beneficial for restoring the functional transfer in individuals with knee OA.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Ethics Committee of the Affiliated Rehabilitation Hospital of Fujian University of Traditional Chinese Medicine, Fuzhou, Fujian, China (#2018KY-006-1). The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

XW and LC: contributed to conceive, design, and obtain funding for the study. SF, TD, and MH: contributed to experimental design, data collection and analysis, and manuscript writing. FY, YaC, YoC, and BL: contributed to data collection and analysis. AL: contributed to the establishment of research questions, the discussion focus, and manuscript revision. YM: contributed to manuscript revision. All the authors were involved in the revision and final approval of the manuscript.

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Role of Spasticity Severity in the Balance of Post-stroke Patients

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Background: Lower limb spasticity after stroke is common that can affect the balance, increase the risk of falling, and reduces the quality of life.

Objective: First, evaluate the effects of spasticity severity of ankle plantar flexors on balance of patients after stroke. Second, to determine the relationship between the spasticity severity with ankle proprioception, passive ankle dorsiflexion range of motion (ROM), and balance confidence.

Methods: Twenty-eight patients with stroke based on the Modified Modified Ashworth Scale (MMAS) were divided into two groups: High Spasticity Group (HSG) (MMAS > 2) ($n = 14$) or a Low Spasticity Group (LSG) (MMAS ≤ 2) ($n = 14$). The MMAS scores, Activities-Specific Balance Confidence Questionnaire, postural sway of both affected and non-affected limbs under the eyes open and eyes closed conditions, timed up and go (TUG) test, passive ankle dorsiflexion ROM, and ankle joint proprioception were measured.

Results: The ankle joint proprioception was significantly better in the LSG compared to the HSG ($p = 0.01$). No significant differences were found between the LSG and HSG on all other outcome measures. There were no significant relationships between the spasticity severity and passive ankle dorsiflexion ROM, and balance confidence.

Conclusion: The severity of ankle plantar flexor spasticity had no effects on balance of patients with stroke. However, the ankle joint proprioception was better in patients with low spasticity. Our findings suggest that the balance is affected regardless of the severity of the ankle plantar flexor spasticity in this group of participants with stroke.

Keywords: stroke, dynamic balance, postural sway, spasticity severity, balance confidence

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INTRODUCTION

Stroke is a common cause of disability and residual physical impairments following a stroke and can pose a significant threat to quality of life (World Health Organization, 2018). Every year, 3.7 million individuals globally suffer a hemorrhagic stroke, while 7.3 million suffer from an ischemic stroke (Feigin et al., 2017). Specifically, the sensorimotor and cognitive impairments following a stroke can have serious impacts on independence and activities of daily living (ADL) (Geurts et al., 2005).

Of these stroke complications, impaired balance is critical for safe mobility, and any deficiencies in balance negatively affect gait, limit ADLs, and/or increases the risk of individuals falling (Paillex and So, 2005; Kollen et al., 2006; Van de Port et al., 2006).

Spasticity is one of common complications of stroke that negatively affects balance. Spasticity is a common sensorimotor disorder defined neurophysiologically as a velocity-dependent increase in muscle tone and stretch reflex hypersensitivity (Lance, 1980). It has been reported as many as 50% of patients with stroke have muscle spasticity (Dorňák et al., 2019). Spasticity contributes in balance dysfunction through various mechanisms (Sinkjær, 1996; Carpenter et al., 2004; Bensoussan et al., 2007; Trumbower et al., 2010). After stroke, lower limb spasticity decreases the range of motion (ROM) and increases the stiffness of the muscles and fascia around the joints (Gao et al., 2011). Further, balance control is further affected when inappropriate muscle and joint afferents and subsequent movement responses occur with inappropriate ankle strategies (Lee et al., 2010).

While lower limb spasticity affects balance, gait, and falling in post-stroke patients (Soyuer and Ozturk, 2007; Sommerfeld et al., 2012), the effects may vary according to the intensity of the muscle spasticity after stroke (Nardone et al., 2001; Cakar et al., 2010; Phadke et al., 2014). To these authors' knowledge, only one study has investigated the relationship between the spasticity severity and the balance in patients with stroke (Rahimzadeh Khiabani et al., 2017). However, authors assessed only static balance using one force plate, and did not evaluate the ankle proprioception and ROM in patients with stroke. Further, they measured the severity of spasticity based on the Modified Ashworth Scale that its reliability and validity is questioned (Ansari et al., 2006) and caution had been expressed in using it for spasticity assessment (Fleuren et al., 2010). Thus, this study aimed to evaluate the effects of ankle plantar flexor spasticity severity on balance and to determine the relationship between the spasticity severity with ankle proprioception, passive ROM, and balance confidence in post-stroke patients. We hypothesized that stroke subjects with high spasticity would have greater balance impairment compared with stroke subjects with low spasticity.

MATERIALS AND METHODS

Study Design

The protocol of this cross-sectional study has been previously reported (Mahmoudzadeh et al., 2020) with the exception of one modification made on posturography. The present study utilized two force plates to assess the both affected and less affected limbs separately. Modified Modified Ashworth Scale (MMAS) scores, Activities-Specific Balance Confidence Questionnaire, postural sway in the open and closed eyes conditions, timed up and go (TUG) test, ankle dorsiflexion passive ROM, and ankle joint proprioception were measured in two post-stroke patient groups based on the level of ankle plantar flexor spasticity [i.e., High Spasticity Group (HSG) (MMAS $>$ 2) and a Low Spasticity Group (LSG) (MMAS $\leq$ 2)].

Setting

The measurements were taken at the Javad Movafaghian Research Center, Tehran, Iran.

Approval of Study Protocol

The study protocol was approved by the Review Board and the Ethical Committee of the Tehran University of Medical Sciences (IR.TUMS.FNM.REC.1397.012) in compliance with the Helsinki declaration. All participants provided their written informed consent prior to the assessments.

Informed Consent

All eligible participants provided a written formal consent after receiving information about the research procedure. Study details, risks, and outcome measures were explained to participants prior to giving the written informed consent and taking the measurements.

Participants

Patients with stroke were included from those who were referred to neurology and physiotherapy clinics in Tehran, Iran. The patients were included if they had the following criteria: (1) unilateral, first-ever Hemorrhagic/Ischemic stroke, (2) ankle plantar flexor spasticity $\geq$ 1 based on the MMAS, (3) walking ability, (4) no fixed contracture in the ankle, (5) independent standing with eyes open/closed, (6) ability to understand and follow the commands, and (7) no pain in the lower limbs. The exclusion criteria were: (1) vision problems, (2) depression, and (3) taking antispastic medications.

Sample Size

Considering a previous study and $\beta = Z_\beta = 0.842$, $\alpha = 0.05$, $\alpha = Z_\alpha = 1.96$ (Rahimzadeh Khiabani et al., 2017; Mahmoudzadeh et al., 2020), the sample size was calculated at 28 (n = 14 in each group).

Procedures

The study procedures and measurements utilized in this study have been published previously (Mahmoudzadeh et al., 2020). Demographic data of the all patients were collected prior to the initiation of assessments. All tests were performed by an experienced physiotherapist. Spasticity severity of ankle plantar flexor muscle was evaluated using the MMAS (Ghotbi et al., 2011; Nakhostin Ansari et al., 2012). Patients were classified as High (MMAS $\geq$ 2) (HSG n = 14) and Low (MMAS $<$ 2) (LSG n = 14) spasticity. The Activities-Specific Balance Confidence (ABC) questionnaire was used to assess the balance confidence (Salbach et al., 2006; Azad et al., 2016) and includes16 questions asking subjects to score their confidence in performing their activities in daily living from 0% (no confidence) to 100% (complete confidence). Posturography was used to assess the static balance (Sawacha et al., 2013; Lendraitienë et al., 2017) using two force plates which were placed together without spaces between to measure the postural sway of affected and less affected limbs independently. The examiner asked each patient to stand on the force plate with bare feet, heel spacing to be 9 cm, the angle between the two feet being 30 degrees, and upper limbs alongside the body. The patient was asked to look at a point on the wall at a distance of 3 m during the test with an open eye and closed eye. The open or closed eye conditions were randomly applied and a 2-min rest was considered between these two conditions. Each condition was repeated for three times (with intervals of 20 s) and the duration of each repetition was 20 s (Rahimzadeh Khiabani et al., 2017). The dynamic balance of patients was measured using the TUG test (Ng and Hui-Chan, 2005). Ankle passive ROM was measured using a standard goniometer (Radinmehr et al., 2019). The ankle joint proprioception was measured using electrogonometer as reconstruction errors. To assess the proprioception, the average of three repetitions of reconstruction angles measured at the angles of 5° and 15° plantar flexion as well as 15° dorsi flexion were calculated as the angles of reconstruction errors.

Outcome Measures

The primary outcome measures were the MMAS scores, ABC questionnaire, posturography measures in open- and closed-eyes conditions, and TUG test. The secondary outcome measures were the ankle passive ROM and ankle joint proprioception. BioWare software (Bioware 5.3.2.9-2.0, Kistler Bioware.msi, Kistler Instrument Group) was used for transforming the force plate data numerical mode. Medio-lateral (ML) and antrio-posterior (AP) displacement, average and instant velocity of the center of pressure (COP) and area were calculated using Excel (Excel, 2010, Ink) and MATLAB (MATLAB, R2018b, Ink) softwares.

Statistical Analysis

Normal distribution of the data was assessed by the Kolmogrov-Smirnov test. *T*-tests were used to compare the clinical data between two groups. A mixed model Repeated Measures ANOVA of $2 \times 2 \times 2$ was performed to analyze the "Group effect" (High spasticity vs. Low spasticity), "Limb effect" (affected and unaffected limbs), "Eye effect" (eyes open and closed conditions), and interactions between variables. The relationships between the severity of spasticity and outcome variables were analyzed using the Spearman's correlation test. SPSS software (SPSS, version 22, SPSS Inc., Chicago, IL, United States) was used for the data analysis. Statistical significance was defined at $\alpha = 0.05$.

RESULTS

Participant Demographics

Twenty-eight post-stroke patients were included in the current study. High and Low spasticity groups were similar in terms of height, weight, age, the time since the stroke onset, etiology (i.e., ischemic or hemorrhage) and affected side ($P > 0.05$). Demographic characteristics of the two study groups are illustrated in **Table 1**.

Clinical Measures

No significant difference was found between the two groups for TUG, ABC scores, and ankle passive dorsiflexion ROM ($p > 0.05$). However, ankle joint proprioception was found to be significantly different between the groups ($p = 0.01$) (**Table 2**).

The distribution of spasticity severity based on MMAS in low and high spasticity groups is presented in **Table 3**. In the low spasticity group, spasticity severity of five patients were MMAS = 1 and spasticity severity of nine patients were MMAS = 2. In the high spasticity group, spasticity severity of nine patient were MMAS = 3 and spasticity severity of other five patients were MMAS = 4.

Posturography

Posturography data for the groups are presented in **Table 4**. There were no significant differences in the medio-lateral (ML) and anterio-posterior (AP) displacements, velocity, and area between open and close eyes conditions, between the affected and unaffected limbs within groups and between groups. The interactions between the groups and the limbs or eyes conditions were not significant ($p > 0.05$).

Correlations

There was a significant correlation between the ABS scores of balance confidence and the TUG test in the HSG ($r = -0.55$, $p = 0.04$) (**Table 5**). There were no other significant correlations between the variables.

DISCUSSION

To these authors' knowledge, this is the first study that evaluated the effects of ankle plantar flexor spasticity severity on balance and determined the relationship between the spasticity severity with ankle proprioception, passive ROM, and balance confidence in post-stroke patients. We found no differences between the LSG and HSG groups in terms of balance confidence, dynamic balance, and ankle dorsiflexion ROM. In addition, postural sway in the open and closed eye conditions was not different in both the LSG and HSG groups for both the less affected and affected limbs. However, ankle joint proprioception in terms of repositioning error angle was better in the LSG compared to the HSG. A relationship was found between TUG scores and balance confidence in the HSG.

TABLE 1 | Demographic characteristics of the High (MMAS* $\geq$ 2) (HSG n = 14) and Low (MMAS < 2) (LSG n = 14) spasticity study groups.

		Low spasticity group (n = 14)	High spasticity group (n = 14)	p-value
Age (year)		59.14 ± 13.8	54.36 ± 10.02	0.3
Height (centimeters)		164 ± 10.62	168 ± 6.3	0.24
Time since stroke (months)		40.07 ± 26.2	54.92 ± 38.63	0.24
Sex (female) (n)		8	6	0.46
Etiology (ischemic) (n)		11	9	0.41
Affected side	Left (n)	8	6	0.71
	Right (n)	6	8	

*Data are presented as means ± SD, *MMAS, Modified Modified Ashworth scale.*

TABLE 2 | Clinical characteristics of the High (MMAS* ≥ 2) (HSG *n* = 14) and Low (MMAS < 2) (LSG *n* = 14) spasticity study groups.

	Low spasticity group (*n* = 14)	High spasticity group (*n* = 14)	*p*-value
Timed Up and Go test (seconds)	17.8 ± 9.9	23.4 ± 8.09	0.11
Balance confidence	1051.4 ± 279.9	996.07 ± 287.5	0.61
Reconstruction error angle (degree)	1.14 ± 0.9	2.03 ± 0.9	0.01*
Ankle passive dorsiflextion range of motion (°)	12.2 ± 1.7	11.6 ± 2.6	0.45

*Data are presented as means ± SD; *statistically significant at p ≤ 0.05, *MMAS, Modified Modified Ashworth scale.*

TABLE 3 | Distribution of spasticity of patients based on MMAS* in two groups (*n* = 28).

Group	Subjects	Affected side	MMAS
Low spasticity group (MMAS < 2)	1	Left	1
	2	Left	1
	3	Right	2
	4	Left	2
	5	Right	2
	6	Right	2
	7	Left	2
	8	Right	1
	9	Left	1
	10	Right	2
	11	Left	1
	12	Right	2
	13	Right	2
	14	Left	2
High spasticity group (MMAS ≥ 2)	1	Right	3
	2	Left	4
	3	Left	4
	4	Left	4
	5	Right	3
	6	Right	3
	7	Right	4
	8	Left	3
	9	Right	4
	10	Right	3
	11	Left	3
	12	Right	3
	13	Right	3
	14	Left	3

**MMAS, Modified Modified Ashworth scale.*

In the standing position, the central nervous system keeps an individual's center of pressure within the base of support (Shumway-Cook and Woollacott, 2007). Therefore, the amount of sway of the pressure center is considered as an indicator of the balance control such that less sways indicate more stability or better balance control (Inness et al., 2015). The findings of the present study showed that there was no difference between ML and AP displacements between the low and high spasticity groups. This finding is similar to that of Rahimzadeh Khiabani et al. (2017) that revealed no differences in ML and AP displacements between the two groups of patients with low and high ankle plantar flexor spasticity. However, Rahimzadeh Khiabani et al. (2017), used only a single force plate and the amount of displacement was the sum of the displacements of both affected and less affected legs. A study used two force plates in patients post-stroke and found no differences in ML and AP displacements of posturography (Singer et al., 2013).

Various investigations have examined the balance control of patients post-stroke in the frontal plane (Geurts et al., 2005; Marigold and Eng, 2006). A review of standing balance in patients with stroke found increased ML sway and impaired balance with asymmetric weight bearing toward the less affected limb (Geurts et al., 2005). In the present study, the absence of a difference in ML displacement between the low and high spasticity groups might have been due to the lack of a difference in the severity of hip and knee spasticity between groups. It may be that ML sway occurs primarily due to the activity of hip adductors and abductors (Winter et al., 1996).

In this study, in line with previous studies (Sosnoff et al., 2010; Rahimzadeh Khiabani et al., 2017), there was no difference in the AP displacement of affected and less affected limbs between the low and high spasticity groups. This may be explained by the fact that the hip and knee movement strategies could be utilized to minimize the ankle movements (Sosnoff et al., 2010). In addition, the compensatory activity of the less affected limb may have played a role in limiting AP displacements of both affected limb and less affected limb observed in this study (Rahimzadeh Khiabani et al., 2017). The differences reported in the AP displacements of patients post-stroke and healthy individuals point to the role of spasticity, regardless of its level, adapted strategies to maintain the balance, and posture stabilization through minimizing the displacements in the frontal and sagittal planes (Marigold et al., 2004; Genthon et al., 2008).

Interestingly, no difference was found in the postural sway between the low and high spasticity groups and between the affected and less affected limbs after the removal of the vision. This was unexpected in that patients post-stroke typically rely on the vision for maintaining their balance (Bonan et al., 2004; Marigold and Eng, 2006). A possible reason for this finding could be that both the low and high spasticity groups used a stiffening strategy by maintaining the knees in extension to increase the stability and balance, regardless of their eyes being open or closed (Mansfield et al., 2013). Further, both groups could also have shifted more weight onto the affected limb in both the eyes open and closed conditions as a further strategy to minimize the AP displacements and thus improving the postural balance (Mansfield et al., 2013).

This study demonstrated that the high spasticity group displayed worse ankle proprioception when compared to the low spasticity group. This finding is in agreement with a previous study that showed proprioception impairment in patients in post-stroke (Carey et al., 1993). This finding could be expected as the high spasticity has been shown to impair the accuracy of deep position sense input (Lee et al., 2010; Gao et al., 2011). While a decreased ankle proprioception in post-stroke patients with spasticity may be postulated that can impair the balance function (Horak et al., 1997; Niam et al., 1999; Tyson et al., 2006), this unexpectedly was not found in the present study. This could be due to the use of an ankle strategy depending on the type of somatosensory input for balance control.

In the present study, in line with a previous study (Rahimzadeh Khiabani et al., 2017), no difference was found in the ABC scores between the low and high spasticity groups. We expected with increasing the severity of muscle spasticity, balance confidence would decrease in post-stroke patients. Nevertheless, a negative relationship between balance

TABLE 4 | Posturography data of the High (MMAS* ≥ 2) (HSG** n = 14) and Low (MMAS < 2) (LSG*** n = 14) spasticity study groups.

			Affected lower limb			Less affected lower limb		
			Mean ± SD	**Max**	**Min**	**Mean ± SD**	**Max**	**Min**
Low spasticity group (n = 14)	Closed eyes	Anterio-posterior displacement (mm)	0.55 ± 0.32	0.21	1.13	0.50 ± 0.26	0.19	1.2
		Medio-lateral displacement (mm)	0.90 ± 0.32	0.28	1.62	0.84 ± 0.15	0.64	1.14
		Velocity (m.sec^{-1})	1.36 ± 1.30	0.51	5.66	0.88 ± 0.23	0.48	1.41
		Area (m^2)	2.90 ± 0.42	2.32	3.75	2.83 ± 0.34	3.42	2.33
	Open eyes	Anterio-posterior displacement (mm)	0.69 ± 0.63	0.15	2.54	0.57 ± 0.33	0.22	1.26
		Medio-lateral displacement (mm)	1.10 ± 0.66	0.55	3.25	0.84 ± 0.16	0.63	1.19
		Velocity (m.sec^{-1})	1.26 ± 0.96	0.50	4.37	0.9 ± 0.23	0.69	1.51
		Area (m^2)	2.90 ± 0.42	2.30	3.83	2.8 ± 0.33	2.34	3.37
High spasticity group (n = 14)	Closed eyes	Anterio-posterior displacement (mm)	0.64 ± 0.45	0.21	1.74	0.43 ± 0.26	0.12	0.92
		Medio-lateral displacement (mm)	1.01 ± 0.30	0.41	1.76	0.78 ± 0.27	0.46	1.44
		Velocity (m.sec^{-1})	1.16 ± 0.39	0.69	2.09	0.75 ± 0.29	0.37	1.21
		Area (m^2)	2.90 ± 0.42	2.30	3.84	2.9 ± 0.33	2.38	3.48
	Open eyes	Anterio-posterior displacement (mm)	0.54 ± 0.42	0.12	1.64	0.52 ± 0.3	0.01	0.97
		Medio-lateral displacement (mm)	1.00 ± 0.32	0.41	1.72	0.73 ± 0.26	0.39	1.27
		Velocity (m.sec^{-1})	1.14 ± 0.37	0.66	2.02	0.76 ± 0.29	0.37	1.22
		Area (m^2)	2.80 ± 0.45	2.26	3.72	2.9 ± 0.38	2.29	3.53

*mm, millimeters; m.sec^{-1}, meters per second; m^2, square meters; *MMAS, Modified Modified Ashworth scale; **HSG, High spasticity group; ***LSG, Low spasticity group.*

TABLE 5 | Correlation between variables in the High (MMAS* ≥ 2) (HSG** n = 14) and Low (MMAS* < 2) (LSG*** n = 14) spasticity study groups low and high spasticity groups.

		TUG	**Balance confidence**	**Proprioception**	**Passive dorsi-flexion ROM**
Low spasticity group (n = 14)	TUG		r = −0.39 p = 0.16	r = -0.2 p = 0.49	r = −0.11 p = 0.69
	Balance confidence			r = −0.03 P = 0.93	r = 0.09 p = 0.74
	Proprioception				r = 0.49 p = 0.07
	Ankle plantarflexor MMAS	r= −0.05 p = 0.85	r = −0.46 p = 0.09	r = −0.47 P = 0.09	r = −0.36 p = 0.21
High spasticity group (n = 14)	TUG test		r = −0.55* p = 0.04	r = −0.31 p = 0.27	r = −0.27 p = 0.37
	Balance confidence			r = 0.8 p = 0.76	r = 0.26 p = 0.36
	Proprioception				r = 0.36 p = 0.2
	Ankle plantarflexor MMAS	r = 0.16 p = 0.85	r = 0.04 p = 0.09	r = 0.09 p = 0.09	r = 0.32 p = 0.26

*TUG, Timed up and go test; *MMAS, Modified Modified Ashworth scale; **HSG, High spasticity group; ***LSG, Low spasticity group; ROM, Range of Motion.*

confidence and static balance as well as gait in patients post-stroke has been demonstrated (Schinkel-Ivy et al., 2016, 2017). In addition, a higher spasticity level has been associate with recurrent falling in patients post-stroke (Wei et al., 2017).

In this study, dynamic balance, as measured by the TUG, was not different between the low and high spasticity groups. This indicated that regardless of spasticity level, the dynamic balance was impaired in this sample of patients with stroke. However, the scores on the TUG test were worse in the High spasticity group than those in the Low spasticity group (23.4 vs. 17.8). A difference of $\sim$6 s between the two groups was clinically relevant that indicated that with increasing spasticity level severity the dynamic balance would worsen as demonstrated in previous studies with stroke (Lin et al., 2006; Soyuer and Ozturk, 2007; Sommerfeld et al., 2012). Replicating the study with more patients in the groups will clarify it.

Passive ankle ROM was not different between low and high spasticity groups. A previous study has reported that a higher spasticity was associated with higher limitations in the passive ROM (Li, 2020). In this study, both the low and high spasticity group had significant restricted ankle passive ROM. This suggests that the ankle passive ROM is influenced by ankle muscle spasticity regardless of spasticity intensity. Spasticity, weakness of ankle muscles, and muscle contracture may explain the limited ankle passive ROM in both groups (Mecagni et al., 2000; Li, 2020).

There was a significant negative correlation between the balance confidence and TUG test in the High spasticity group. This finding indicates that the time for the TUG test will be lower with more confidence on balance. However, there was no significant correlation between spasticity and proprioception and passive ROM of the ankle, balance confidence, TUG test, and postural sway in each group. The sample size is very important to examine the correlation in cross-sectional studies. As the number of variables increases, more samples are needed (Schönbrodt and Perugini, 2013). The non-significant correlations obtained for the variables might be due to the small sample size. A study with larger sample of stroke patients of different level of spasticity is required to clarify the size of correlations.

Limitations

Although the sample size was determined, the number of samples may have been too small to investigate the relationship between clinical data and static and dynamic balance (Schönbrodt and Perugini, 2013). In the present study, both groups of stroke patients had spasticity and there was also no control group. A control group consisting of neurologically healthy people or stroke patients without spasticity will help to more closely examine the effect of spasticity on balance.

CONCLUSION

These results show that although stroke patients with spasticity have impaired static and dynamic balance, the severity of spasticity has no effect on the exacerbation of balance control. Therefore, the spasticity post stroke must be considered for management regardless of its severity.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Ethics Committee of Tehran University of Medical Sciences (IR.TUMS.FNM.REC.1397.012). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

NN, SN, and AM presented the study conception and idea. NN, SN, AM, OM, EG, BS, and IS designed the study protocol. AM drafted the first version of this manuscript that was reviewed and revised critically for intellectual content by NN. AM, EG, and OM collected the data. NN, SN, BS, and IS revised the manuscript. All authors read, commented, and approved the final manuscript for submission.

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Can Compression Garments Reduce Inter-Limb Balance Asymmetries?

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Sensory cues provided by compression garments (CG) can improve movement accuracy and potentially reduce inter-limb balance asymmetries and the associated risk of injury. The aim of this study was to analyze the effects of CG wearing on inter-limb balance asymmetries. The hypothesis was that CG would reduce inter-limb balance asymmetries, especially in subjects with high level of asymmetries. Twenty-five sportsmen were recruited. They had to stand as motionless as possible in a one-leg stance in two postural tasks (stable and unstable), while wearing CG or not. Asymmetry indexes were calculated from center of foot pressure parameters. The effects of CG wearing were analyzed according to participants' baseline level of asymmetry (i.e., without wearing CG) with correlation analyses. A qualitative analysis was also performed after a dichotomization procedure to check for a specific influence of CG on the dominant and non-dominant leg. Inter-limb balance asymmetries were reduced with CG in participants with high levels of asymmetries at baseline. However, asymmetries were increased with CG in participants with low levels of asymmetries at baseline. The dominant leg was more affected by this negative effect. CG wearing could reduce inter-limb balance asymmetries and the related injury risk in subjects with high levels of inter-limb balance asymmetries at baseline. Nevertheless, CG should not be used in individuals with low baseline balance asymmetries since it can increase asymmetries in these subjects, likely by confusing and overloading the sensorimotor processing on the dominant leg.

Keywords: laterality, symmetry, posture, balance control, compression garments

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INTRODUCTION

Asymmetries between two limbs are prevalent in human movement. They are characterized by a difference in performance between the right and left limbs and/or the preferential use of one side of the body (i.e., left or right) to perform a motor action (Schneiders et al., 2010; Bishop et al., 2018). The etiologies of inter-limb asymmetries are related to lateralization in sensorimotor control originating from cerebral hemispheric asymmetry (Promsri et al., 2018, 2019; Marcori et al., 2020; Liu et al., 2021) and biomechanical factors (e.g., bilateral asymmetry in the bones length, imbalance of muscle strength between the left and right limb) (Auerbach and Ruff, 2006; Bishop et al., 2018). Inter-limb asymmetries can be modulated by numerous factors. Injuries and pathologies such as anterior cruciate ligament injuries (Mohammadi et al., 2012) or scoliosis (Yang et al., 2013) increase inter-limb asymmetries. Subjects affected by neurologic conditions (e.g., stroke and multiple sclerosis) and individuals with experience in asymmetric sports (e.g., soccer, volley-ball, and hand-ball) also demonstrate pronounced inter-limb asymmetries (Chisholm et al., 2011; Paillard and Noé, 2020; Pau et al., 2021).

Inter-limb asymmetries can have a negative impact on sports performance and are associated with a high incidence of lower-limb injury, especially if asymmetries concern balance control (Stiffler et al., 2017; Bishop et al., 2018; Promsri et al., 2019). Actually, inter-limb balance asymmetries may impair the ability to shift the body weight onto one leg thus negatively impacting sports-related actions such as sidestepping or changes of direction and predisposing sportspeople to non-contact lower-limb injury (Bishop et al., 2018; Dos'Santos et al., 2019; Promsri et al., 2019). Therefore, training strategies such as balance training, resistance training or warm-up programs, can be implemented to minimize inter-limb balance asymmetries, improve balance control and reduce the relative risk of sports-related non-contact injury (Bishop et al., 2018; Pardos-Mainer et al., 2019; Madruga-Parera et al., 2020). Affordable external devices that interact with cutaneous receptors such as compression garments (CG) can also acutely improve balance control (Kuster et al., 1999; Michael et al., 2014; Woo et al., 2017; Baige et al., 2020). The constriction provided by CG acts as a mechanically supportive framework that can activate interacting cutaneous mechanoreceptors that individually would not have been activated (Baige et al., 2020), thus improving movement accuracy during tasks that include a large somatosensory component (Hasan et al., 2016; Ghai et al., 2018; Broatch et al., 2021) and offering a potential benefit in reducing inter-limb balance asymmetries. Nevertheless, to our knowledge, no study has specifically investigated the effects of CG wearing on inter-limb balance asymmetries.

Hence, this study was undertaken to investigate the effects of CG on inter-limb balance asymmetries in individuals with extensive experience in an asymmetric sport such as handball players. Handball is a sport involving asymmetric motor actions (lower and upper limbs) that can accentuate inter-limb asymmetries, which might negatively affect athletic performance and increase the risk of lower extremity injuries (Steib et al., 2016; Paillard, 2017; Barrera-Domínguez et al., 2021). Thus, a passive intervention such as CG wearing is hypothesized to reduce inter-limb balance asymmetries, especially in subjects with a high level of asymmetry, and could represent a cost effective sport-related injury prevention strategy.

MATERIALS AND METHODS

Participants

Twenty-five young handball players (age: 27.1 ± 6.9 years old, height: 184.8 ± 8.1 cm; body mass: 87.2 ± 15.3 kg; mean ± SD) who reported no neuromuscular problems in the past 2 years volunteered for the current study. Participants were recruited from two regional semi-professional teams. They were asked to avoid strenuous activity and the ingestion of alcohol or/and excitatory substances 24 h before the experimental session. A written informed consent was obtained from all participants before starting the experiment, which was in accordance with the Declaration of Helsinki. All procedures were approved by and performed in accordance with the relevant guidelines and regulations of the University of Pau and Pays de l'Adour Ethics Committee.

Apparatus and Procedure

Participants were asked to sway and move as little as possible when standing barefoot in a one-leg stance for 25 s on a force platform (Stabilotest® Techno Concept™, Mane, France) which sampled the displacements of the center of foot pressure (position of the point of application of the vertical ground reaction force) at 40 Hz. Two postural tasks were conducted: a stable task where participants stood directly on the force platform (on stable ground) with their eyes closed (while keeping their gaze straight ahead) and an unstable task where they stood with their eyes open (while looking at a fixed level target at a distance of 2 m) on a wobble board with a diameter of 40 cm and a height of 8 cm (Balance-board, Sissel® GmbH, Bad Dürkheim, Germany) which was placed on the force platform to generate instability. For accurate and similar feet positioning between all participants, the platform and the wobble board were marked with a central horizontal line, and participants were required to align the middle of the foot with the mark (the middle of each foot was delimited beforehand). The center of the wobble board was also aligned with the center of the force platform.

Postural tasks were performed with or without wearing compression garments (CG and REF condition, respectively). In the CG condition, calf compression sleeves (Booster, BV sport®, Saint Etienne, France) made of 79% Polyamide and 21% Elastane, were worn by the participants (**Figure 1**). According to the manufacturer's specifications, these garments provide a pressure that increases gradually from 13 mmHg at the lower part of the calf to 20–25 mmHg at the gastrocnemius belly (20 mmHg at the lateral part and 25 mmHg at the medial part). The size of the compression sleeves was individually fitted according to guidelines of the manufacturers, based on participants' height and calf circumference. Four sizes were used: M+, L+, XL+, and XXL+, sized for 34–38/>175, 38–43/>175, 38–43/>192, 43–48/192 (calf circumference/height, in cm), respectively, so as to accommodate all participants' body shape. Participants performed a set of three trials in each postural task (stable and unstable) and condition (REF and CG) with each set completed in a randomized order to avoid any learning effect. The first two trials were not recorded and served as familiarization trials. Since a stable and reliable balance score is achieved at the third trial in a one one-leg stance (Cug and Wikstrom, 2014), only the third trial was recorded and considered for statistical analysis for each postural task and in each experimental condition. Subjects were sitting quietly on a chair for 2 min between balance assessments.

Analysis of Data

The following parameters of the center of pressure were initially calculated: mean velocity (sum of the cumulated COP displacement divided by the trial time) along the medio-lateral (VX) and antero-posterior (VY) axes and surface area (S: 90% confidence ellipse) (Paillard and Noé, 2015). Based on the method

FIGURE 1 | Compression sleeves worn by participants in the compression garments (CG) condition.

proposed by Read et al. (2018) and Fort-Vanmeerhaeghe et al. (2020), an asymmetry index (ASY) between both legs was then calculated for each center of pressure parameter (ASY_VX, ASY_VY, ASY_S) as follows:

$$\text{ASY }(\%) = \text{ ABS}\left(\frac{(\text{Highest performing limb} - \text{Lowest performing limb})}{\text{Highest performing limb}}\right) \times 100,$$

with ABS: absolute value.

When considering that with participants who were asked to sway as little as possible when standing upright, the lower the center of pressure parameters (i.e., VX, VY, S), the more efficient the balance control (Paillard and Noé, 2015), the highest performing limb was then defined as the side with the lowest center of pressure value.

Statistical Analysis

The two postural tasks (stable and unstable) were analyzed independently. Normality was tested using the Shapiro–Wilk test. As the dependent variables (ASY_VX, ASY_VY, and ASY_S) did not meet the assumption of normal distribution, non-parametric Wilcoxon sign rank tests were applied to determine differences between the CG and REF conditions. The difference of each asymmetry index between the CG and REF conditions (difference = CG - REF) was also calculated to easily differentiate participants who benefit from CG wearing to reduce inter-limb balance asymmetries (negative difference) and those who do not (positive difference). Then, Spearman's rank order correlations were undertaken between asymmetry indexes at baseline (i.e., in the REF condition) and differences in asymmetry between the REF and CG conditions to assess whether the beneficial effects of CG in reducing inter-limb balance asymmetries was related to the participant's inter-limb balance asymmetries at baseline.

A qualitative analysis was also carried out to check for a specific influence of compression garments on balance control of the dominant and non-dominant leg, determined by the preferred kicking leg (Paillard and Noé, 2020). A dichotomization method was performed on the center of pressure parameters to characterize three levels of CG influence on balance control: a negative influence, a negligible influence and a positive influence. The following are the thresholds used in the dichotomization procedure: a negative influence was considered when the value in the CG condition exceeded that in the REF condition by more than 10%; a negligible influence was considered when the difference between either conditions did not exceed 10%; a positive influence was considered when the value in the REF condition exceeded that in the CG condition by more than 10%. χ^2-conformity tests were independently performed on DL and NDL data to test the null hypothesis (similar distribution of individuals for whom CG have a negative, negligible, or positive influence on balance control). This qualitative analysis was performed on the whole sample and on subgroups of individuals in order to check whether individuals who benefited from CG to decrease inter-limb balance asymmetries and those who do not were differently affected by the wearing of CG on the dominant and non-dominant leg. Sub-groups were formed by distinguishing subjects in whom CG wearing had no substantial influence on asymmetry indexes ("unchanged asymmetry with CG" subgroup, i.e., with an increase/decrease of the asymmetry index between REF and CG conditions of less than 10%), subjects in whom CG wearing markedly increased the level of asymmetry ("increase in asymmetry with CG" subgroup, i.e., with an increase of the asymmetry index of more than 10% in the CG condition) and subjects in whom CG wearing markedly decreased the level of asymmetry ("decrease in asymmetry with CG" subgroup, i.e., with a decrease of the asymmetry index of more than 10% in the CG condition). Based on previous studies about reliability of center of pressure-based postural sway measures during one-leg stance in young healthy participants, the 10% threshold was chosen to distinguish between clear and unclear changes due to measurement noise (Lin et al., 2008; Muehlbauer et al., 2011; da Silva et al., 2013). The relationships between subgroups of individuals and specific influence of CG on the dominant and non-dominant leg were then tested with the Fisher's exact test after having constructed contingency tables. Statistical analyses were performed with R statistical software. The significance level was set at $p < 0.05$.

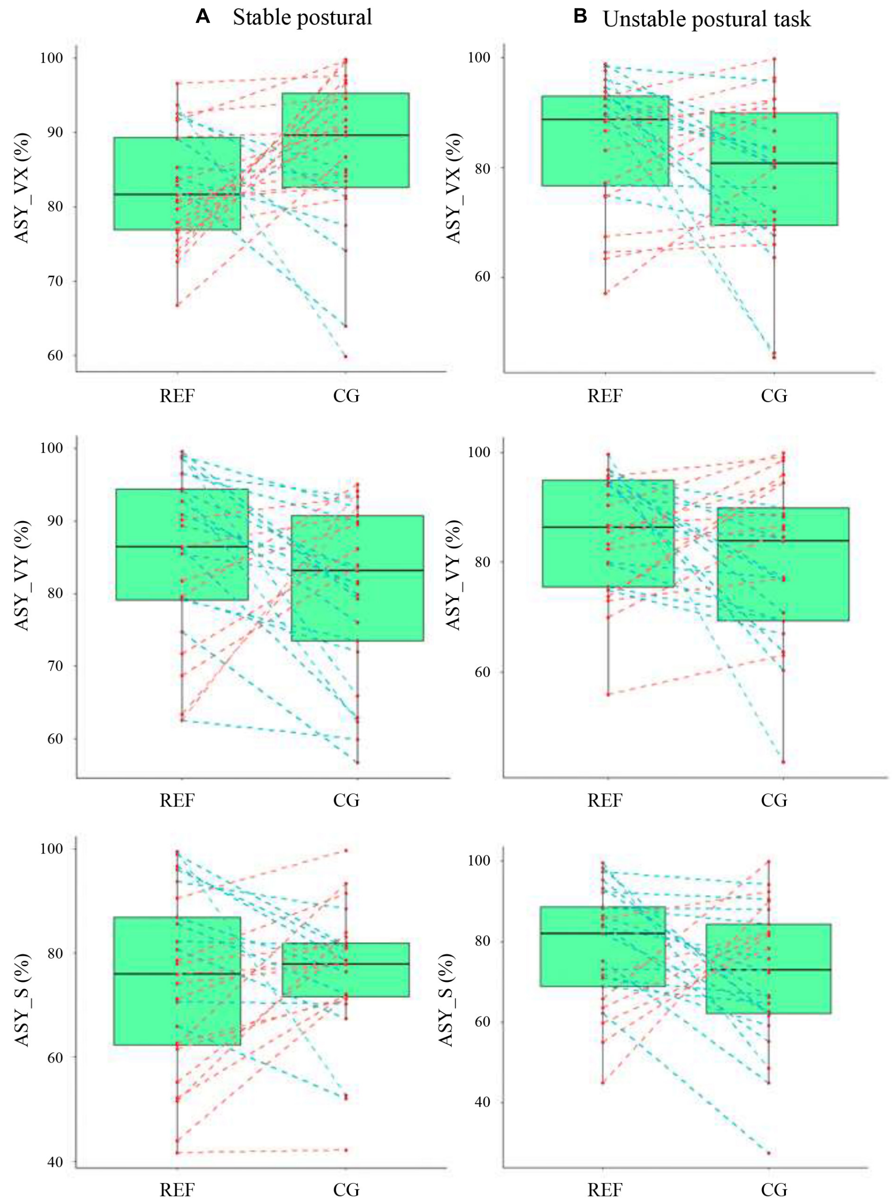

FIGURE 2 | Boxplot representation with individual data points of center of pressure asymmetry indexes in the reference (REF) and compression garments (CG) conditions in **(A)** the stable postural task and **(B)** the unstable postural task. The red dotted lines illustrate individuals whose asymmetry indices increase with CG while the blue dotted lines show individuals whose asymmetry indices decrease with CG.

RESULTS

Figure 2 illustrates boxplots with individual data points of asymmetry indexes in the REF and CG conditions. No significant differences were observed between REF and CG conditions in both the stable and unstable postural tasks. When focusing on the individual data points, one could notice that the asymmetry indexes did not evolve similarly among participants between the REF and CG conditions: asymmetry indexes increased with CG in some participants (red dotted lines on **Figure 2**), while asymmetry indexes decreased with CG in others (blue dotted lines on **Figure 2**).

Scatterplots illustrating correlations between ASY indexes in the REF condition and the differences between the CG and REF conditions are shown in **Figure 3**. In both stable and unstable postural tasks and for all asymmetry indexes, there were significant and large relationships between asymmetry indexes in the REF condition and asymmetry differences between CG and REF conditions. This result indicates that high levels of inter-limb balance asymmetry at baseline [approximate (40–120%) range of asymmetry] was associated with negative differences between

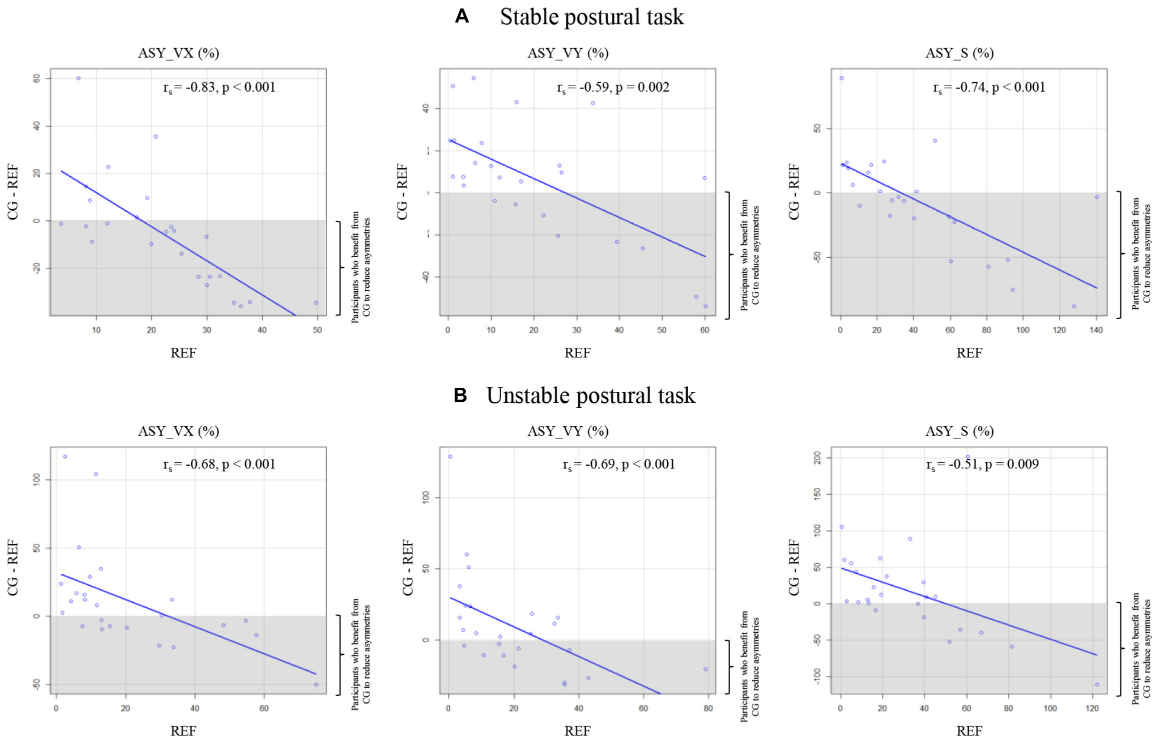

FIGURE 3 | Scatterplots representing the correlations between asymmetry indexes (ASY_VX, ASY_VY, and ASY_S) in the REF condition and the differences in asymmetry between REF and compression garments (CG) conditions (CG–REF) in **(A)** the stable postural task and **(B)** the unstable postural task. Gray areas correspond to participants who benefit from CG wearing to reduce inter-limb balance asymmetries (negative difference). Blank areas correspond to participants who do not benefit from CG wearing to reduce inter-limb balance asymmetries (positive difference). r_S: Spearman's rank correlation coefficient.

CG and REF conditions (i.e., a beneficial effect of CG wearing in reducing inter-limb balance asymmetries). On the contrary, low levels of inter-limb balance asymmetry at baseline [approximate (0–40%) range of asymmetry] were rather associated with positive differences between CG and REF conditions (i.e., an increase of inter-limb balance asymmetries when wearing CG).

Table 1 shows the distribution of participants according to the influence of CG wearing (negative, negligible, or positive influence) on balance control of the dominant and non-dominant leg in the whole sample and in subgroups of individuals. In the stable postural task, results from χ^2 conformity tests performed in the whole sample were not significant, thus illustrating a homogeneous distribution among participants with a negative, negligible or positive influence of CG wearing on balance control of the dominant or the non-dominant leg. In this postural task, results from the Fisher's exact test were also not significant. In the unstable postural task, results from the χ^2 conformity test applied on participants' distribution based on S data on the dominant-leg showed a significant difference from an equal distribution ($\chi^2 = 6.08$; $p < 0.05$), with a an under-representation of individuals for whom CG wearing had negligible influence on the dominant-leg. When the χ^2 test was applied on participants' distribution based on VY data on the non-dominant leg, a significant difference from an equal distribution ($\chi^2 = 11.12$; $p < 0.004$), with an over-representation of individuals for whom CG wearing had negligible influence on balance control of the non-dominant leg. Results from the Fisher's exact test showed a significant association between subgroups of participants and CG influence on balance control on the dominant leg when participants' distribution were based on VX ($p < 0.003$) and VY data ($p < 0.005$). Individuals in whom CG wearing markedly increased the level of asymmetry were negatively influenced by CG wearing on the dominant leg. Results from the Fisher's exact test did not show any significant association between subgroups of participants and CG influence on balance control of the non-dominant leg.

DISCUSSION

The aim of this study was to analyze the effects of CG wearing on inter-limb balance asymmetries. The hypothesis was that CG would reduce inter-limb balance asymmetries, especially in subjects with high level of asymmetries. When using a standard statistical approach with pairwise comparisons between group levels, our results showed that the wearing of CG was not associated with a significant reduction of inter-limb balance asymmetries. However, when analyzing the effect of CG wearing according to participants' initial level of asymmetry, results from correlation analyses indicated that beneficial effects of CG in reducing inter-limb balance asymmetries were related to participant's asymmetry levels at baseline. Only participants

TABLE 1 | Distribution of participants according to the influence of compression garments (CG) wearing (negative, negligible, or positive) on the dominant and non-dominant leg in the whole sample and in subgroups of individuals.

				Stable task			Unstable task		
				Negative	**Negligible**	**Positive**	**Negative**	**Negligible**	**Positive**
VX		Whole sample	Dominant leg	9	10	6	10	9	6
			Non-dominant leg	5	12	8	8	11	6
	Subgroups	Decrease in asymmetry with CG	Dominant leg	5	2	3	4	0	3
			Non-dominant leg	2	4	4	1	3	3
		Unchanged asymmetry with CG	Dominant leg	1	6	2	0	6	0
			Non-dominant leg	1	6	2	1	5	0
		Increase in asymmetry with CG	Dominant leg	3	2	1	6	3	3
			Non-dominant leg	2	2	2	6	3	3
VY		Whole sample	Dominant leg	10	8	7	10	7	8
			Non-dominant leg	6	10	9	3	16	6
	Subgroups	Decrease in asymmetry with CG	Dominant leg	3	2	3	3	3	3
			Non-dominant leg	1	4	3	2	5	2
		Unchanged asymmetry with CG	Dominant leg	1	1	0	0	4	0
			Non-dominant leg	1	1	0	0	4	0
		Increase in asymmetry with CG	Dominant leg	6	5	4	7	0	5
			Non-dominant leg	4	5	6	1	7	4
S		Whole sample	Dominant leg	11	3	11	13	3	9
			Non-dominant leg	9	7	9	8	8	9
	Subgroups	Decrease in asymmetry with CG	Dominant leg	3	2	7	4	0	3
			Non-dominant leg	6	3	3	2	3	2
		Unchanged asymmetry with CG	Dominant leg	1	1	2	1	2	1
			Non-dominant leg	0	1	3	0	3	1
		Increase in asymmetry with CG	Dominant leg	7	0	2	8	1	5
			Non-dominant leg	3	3	3	6	2	6

Subgroups of individuals were formed by distinguishing subjects in whom CG wearing had no substantial influence on asymmetry indexes ("unchanged asymmetry with CG" subgroup), subjects in whom CG wearing markedly increased the level of asymmetry ("increase in asymmetry with CG" subgroup) and subjects in whom CG wearing markedly decreased the level of asymmetry ("decrease in asymmetry with CG" subgroup).
VX, mean velocity of the center of pressure along the medio-lateral axis; VY, mean velocity of the center of pressure along the antero-posterior axis; S, center of pressure surface area (90% confidence ellipse).

with high levels of inter-limb balance asymmetries at baseline benefited from CG to decrease inter-limb balance asymmetries.

Inter-limb balance asymmetries are mainly related to differences in somatosensory processing between both legs originating from cerebral hemispheric asymmetry (Promsri et al., 2018, 2019; Marcori et al., 2020; Liu et al., 2021). Within the perspective of the dynamic dominance model of laterality (Sainburg, 2005, 2014), one hemisphere (the left hemisphere for right-handed individuals) would rather be specialized in limb trajectory control, whereas the other hemisphere would rather be specialized for impedance control—i.e., control of limb position and maintenance of a stable posture (Sainburg, 2005, 2014; Marcori et al., 2020). Lateralization of impedance control is based on hemisphere specialization for the utilization of somatosensory cues (Goble et al., 2006; Sainburg, 2014; Liu et al., 2021), which is characterized by the superiority of one hemisphere in processing somatosensory information (the right hemisphere for right-handed individuals), thus resulting in better proprioception of one limb (the left limb for right-handed individuals) compared to the other (Carnahan and Elliott, 1987; Goble et al., 2006). Participants' injury history and/or experience in asymmetric motor practices can modulate lateralization of impedance control and increase the difference in somatosensory processing between the two legs, thus inducing large inter-individual differences of balance inter-limb asymmetry (Marcori et al., 2019; Paillard and Noé, 2020).

When wearing CG, frictional forces that activate both slow and fast-adapting cutaneous mechanoreceptors are applied to the skin, which can improve joint position sense and balance control (Kuster et al., 1999; You et al., 2004; Cameron et al., 2008; Michael et al., 2014; Woo et al., 2017; Baige et al., 2020; Broatch et al., 2021). In the present study, there was a great heterogeneity in the ability of participants to benefit from CG wearing to reduce inter-limb balance asymmetries. Previous studies about the effects of CG on functional somatosensory abilities have produced concordant findings while reporting a strong inter-individual variability in responses to the wearing of CG in young healthy participants (You et al., 2004; Cameron et al., 2008; Broatch et al., 2021). After ranking participants according to their score in a movement discrimination task, these studies showed that only participants with poor lower limb somatosensation benefited from the wearing of CG to improve joint position sense, thus illustrating that the magnitude of the beneficial effects of CG wearing was inversely related to the participant's somatosensation at baseline. Similar findings were observed with other stimulation strategies such as ankle taping (Long et al., 2017), textured insoles (Steinberg et al., 2016), and vibration stimulation (Liu et al., 2021).

In the present study, participants with high levels of inter-limb balance asymmetries at baseline benefited from CG wearing to reduce inter-limb balance asymmetries. However, in both the stable and unstable postural tasks, the data from the qualitative analysis about the specific influence of CG on both legs (**Table 1**) did not illustrate a stronger influence of CG on the dominant leg or the non-dominant leg in individuals who benefited from CG wearing to markedly decrease inter-limb asymmetries. In participants with low levels of inter-limb balance asymmetries at baseline, CG were unhelpful and could also increase inter-limb-balance asymmetries in some participants. Further studies reported analogous findings while showing that proprioceptive acuity decreased in participants with higher levels of proprioceptive ability when they wore external devices that stimulate cutaneous receptors (Cameron et al., 2008; Steinberg et al., 2016; Long et al., 2017; Liu et al., 2021). These authors postulated that in subjects with a high proprioceptive acuity at baseline, the application of an external stimulation device confused and overloaded the sensorimotor system by delivering redundant information, thus resulting in an impaired proprioceptive ability. Liu et al. (2021) noticed that proprioception deteriorated only in the dominant leg in individuals with high proprioceptive acuity following stimulation of the calf muscles with a vibrating foam roller. Our results from the qualitative analysis also suggest that, in the most challenging postural task (i.e., the unstable postural task) CG wearing would have a greater impact on the dominant leg and, more specifically, that CG negatively influenced balance control only on the dominant leg in participants in whom CG wearing increased inter-limb asymmetries. Our results therefore seem to confirm the hypothesis formulated by Liu et al. (2021) about different effects of the application of external stimuli on different hemispheres and show that the non-dominant hemisphere would to be less sensitive to somatosensory stimulation.

A limitation of the current research is the absence of measure of the pressure exerted by the calf compression sleeves. Although compression sleeves were individually fitted by following the manufacturer's specifications and choosing proper sizing according to individuals' calf circumference, commercially manufactured compression garments do not provide exactly the same level of compression in all subjects. Potential differences in the pressures exerted by the CG might modulate the effects of CG on inter-limb balance asymmetries. Nevertheless, studies that have tested the effects of CG providing various compression levels on balance control did not report any significant differences attributed to the level of compression (e.g., Hijmans et al., 2009; Jaakkola et al., 2017; Woo et al., 2018). It should be noted that the compression sleeves used in this study did not cover the ankle joint, whereas somatosensory information from the ankle joint plays an essential role in balance control (Han et al., 2015). Hence, one can hypothesize that the use of CG covering the ankle joints such as compression socks could potentially provide larger benefits in reducing inter-limb balance asymmetries. Further experiments are needed to explore the influence of various designs of CG (e.g., full leg CG, compression socks, calf and knee compression sleeves) on inter-limb balance asymmetries.

The present study provides new insights about the influence of CG on inter-limb balance asymmetries. From a practical point of view, these results suggest that CG wearing could reduce inter-limb balance asymmetries and the related injury risk in participants with high levels of inter-limb balance asymmetries at baseline. In return, since it rather increases asymmetries in participants having low baseline balance asymmetries, CG wearing should be avoided in these participants.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the University of Pau and Adour Countries Ethics Committee. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

FN, KB, and TP designed the study. KB acquired the data. FN analyzed the data and wrote the manuscript, while KB and TP revised it. All authors signed the final approval for publication.

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Load-Induced Changes of Inter-Limb Asymmetries in Dynamic Postural Control in Healthy Subjects

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Inter-limb asymmetries are associated with a higher potential risk for non-contact injuries. Differences in function or performance between the limbs might lead to imbalances and promote instability, increasing the potential risk for injuries. Consequently, an investigation of inter-limb asymmetries should be included in injury risk assessment. Furthermore, since non-contact injuries mainly occur under loaded conditions, an investigation of load-induced changes of inter-limb asymmetries can provide additional information on the athlete's potential injury risk. Therefore, the current study aimed to investigate the influence of physical load on inter-limb asymmetries in dynamic postural control, which is essential in situations with a high risk for non-contact injuries such as landing, cutting, or stopping. In total, dynamic postural control of 128 active and healthy subjects (64 males and 64 females, age: 23.64 ± 2.44, height: 176.54 ± 8.96 cm, weight: 68.85 ± 10.98 kg) was examined. Dynamic postural control was tested with the Y-Balance Test (YBT) before and after a loading protocol on a bicycle ergometer or a treadmill. The results showed no significant increase of the inter-limb asymmetries in anterior direction [$F_{(1,\ 126)} = 4.44$, $p = 0.04$, $\eta^2_p = 0.03$]. Moreover, there is high variation between the subjects regarding the magnitude and the direction of the asymmetries and the changes due to load. Therefore, a more individual analysis considering the magnitude and the direction of the asymmetries is required. Thereby, considering different modifying factors, e.g., sex, injury history, and baseline level of asymmetries, can be helpful. Moreover, an analysis of the changes during load might provide further insights, reveal possible differences, and help detect the reasons and mechanisms underlying inter-limb asymmetries and asymmetrical loading.

Keywords: physical load, running, cycling, side-differences, injury risk, Y-Balance Test

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INTRODUCTION

Many athletes develop a difference in function or performance between their limbs, i.e., *inter-limb asymmetries* (Bishop et al., 2017). It might occur differences in strength, physical capacity, or balance (Bishop et al., 2018b; Dos'Santos et al., 2021; Helme et al., 2021). Such differences appear in different sports and might be a consequence of the sporting activity (Parrington and Ball, 2016). Many sports are mainly characterized by asymmetric (or unilateral) execution of movements with a preferred limb, such as kicking in soccer or throwing in handball (Bromley et al., 2021). Such unilateral movements possibly evoke inter-limb asymmetries (Parrington and Ball, 2016; Bishop et al., 2018b). However, inter-limb asymmetries occur not only in asymmetric but also in symmetric sports with mainly cyclic or alternating movement patterns, e.g., running, cycling, or swimming (Hart et al., 2016; Parrington and Ball, 2016; Maloney, 2019). These inter-limb asymmetries might be caused by the predominant use of one preferred limb, leading to differences in strength development, neural development, or uneven flexibility and range of motion in favor of the preferred limb (Parrington and Ball, 2016).

Inter-limb asymmetries are associated with a higher potential risk for non-contact injuries (Dos'Santos et al., 2021). They might lead to unequal force absorption or a loss of frontal plane stability which are essential to bear the impacting forces in situations with high unilateral loading and a higher risk for non-contact injuries, such as landing, cutting, or stopping (Paterno et al., 2010). Several studies investigated the relationship between certain side differences and occurring injuries (Helme et al., 2021). For example, asymmetries in movement competencies measured with a functional movement screen (FMS) (Chalmers et al., 2017; Attwood et al., 2019), Y-Balance Test (YBT) (Kiesel et al., 2014; Gonell et al., 2015), or Star Excursion Balance Test (SEBT) (Plisky et al., 2006) showed an association with non-contact injuries. However, the association of inter-limb asymmetries in dynamic force production tests, e.g., single-leg hop (SLH) (Brumitt et al., 2020) or in isolated muscle actions measured with isokinetic tests (Dauty et al., 2016), and injuries is not that clear due to an inconsistency of results (Helme et al., 2021; Guan et al., 2022). Nevertheless, additional consideration of side differences between the limbs can obtain further information about an athlete's potential injury risk and should be implemented in injury risk assessment.

In this context, *dynamic postural control* plays an important role. Non-contact injuries mainly occur during dynamic actions, e.g., landing, cutting, or stopping. In these situations, the athletes must, among others, maintain stability in situations of high (unilateral) loading (Güler et al., 2020). Maintaining balance while the body is in motion is denoted as an athlete's dynamic postural control (Johnston et al., 2018). Regarding the high number of dynamic actions in sports, the importance and necessity of good dynamic postural control in sports is not debatable (Whyte et al., 2015; Johnston et al., 2018). Moreover, poor dynamic postural control is associated with instability and reactive or compensatory movements. Instability and compensatory movements possibly increase the impacting load on the muscles, tendons, and ligaments and therewith the potential risk for lower limb injuries (Plisky et al., 2006; Wright et al., 2013; Whyte et al., 2015; Güler et al., 2020). Additionally, inter-limb asymmetries in dynamic postural control are also associated with a higher potential risk for lower limb injury (Helme et al., 2021). Relative and absolute side-differences measured with the YBT or SEBT, especially in the anterior reach direction, proved to be good precursors for sports injuries (Plisky et al., 2006; Stiffler et al., 2017; Helme et al., 2021). Therefore, it might help to assess inter-limb asymmetries in dynamic postural control to provide more insights into an athlete's potential injury risk.

Furthermore, injuries typically occur during matches, competitions, or training when the athlete is physically stressed (Ekstrand et al., 2021). This might be ascribed to load-induced alterations of physiological processes, possibly leading to altered muscle patterns, reduced muscle activation, delayed muscle contraction, or decreased muscle-torque generation. These alterations possibly result in changes in risk factors, such as knee valgus, ground reaction forces, or dynamic postural control (Santamaria and Webster, 2010; Whyte et al., 2015; Barber-Westin and Noyes, 2017). Therefore, an analysis of the potential injury risk under loaded conditions and not only when the athlete is recovered is advisable, and analysis of inter-limb asymmetries under loaded conditions might gather additional insights into an athlete's potential injury risk (Heil et al., 2020a; Verschueren et al., 2020).

Nevertheless, only a few studies have already examined the influence of physical load on inter-limb asymmetries (Heil et al., 2020a). Bromley et al. (2021), e.g., found a large effect of a soccer match on inter-limb asymmetries in eccentric impulses and peak forces during single-leg countermovement jumps, but only small and moderate effects in other parameters, e.g., peak landing force or peak landing impulse. Bell et al. (2016) showed no changes of inter-limb asymmetries in vertical ground reaction force after a standardized exercise protocol. Moreover, Bishop et al. (2021b) found a large effect of a repeated sprint protocol on jump height asymmetries. In contrast, Bromley et al. (2021) stated no significant changes in jump height asymmetries after a soccer match. About the influence of physical load on inter-limb asymmetries in dynamic postural control is not much known yet. Konstantopoulos et al. (2021) found an increase of inter-limb asymmetries in dynamic postural control measured with the YBT. Nevertheless, the protocol used in this study intended to induce local muscle stress on only one side of the body. Hence, still, nothing is known about more global protocols better reflecting the demands of sports and their influence on inter-limb asymmetries in dynamic postural control.

Overall, these findings indicate a possible change of inter-limb asymmetries due to load. Moreover, these findings also show a dependency of the results and changes on the measured parameter, the load type, the loading protocol, and the task used to measure a certain parameter. Consequently, to obtain reliable insights and deduce aspects for injury prevention practice, future studies need methods, i.e., load types, loading protocols, and tasks, reflecting the demands of the sporting context they should investigate. An athlete's potential injury risk should be assessed under approximately real sporting conditions (Benjaminse et al., 2019; Bolt et al., 2021). In this context, since most sports require running, especially those with a higher risk for non-contact injuries, e.g., soccer, handball, or track and field sports, running protocols should be used. Nevertheless, many studies still use cycling protocols because they seem easier to apply in a laboratory setting (Johnston et al., 2018; Verschueren et al., 2021).

Therefore, the current study aimed to investigate load-induced changes of inter-limb asymmetries in dynamic postural in more representative conditions. Dynamic postural control was analyzed before and after two global loading protocols: a commonly used cycling protocol and a comparable running protocol. It was hypothesized that asymmetries in dynamic postural control would increase after physical load.

TABLE 1 | Subject characteristics.

	Total	Group 1 (Cycling)	Group 2 (Running)
N	128 (64 m, 64 f)	64 (32 m, 32 f)	64 (32 m, 32 f)
Age (years) (*M* ± *SD*)	23.64 ± 2.44	24.11 ± 2.42	23.17 ± 2.37
Height (cm) (*M* ± *SD*)	176.54 ± 8.96	175.53 ± 8.17	177.56 ± 9.65
Weight (kg) (*M* ± *SD*)	68.85 ± 10.98	67.16 ± 10.08	70.51 ± 11.67
Leg length kicking leg (cm) (*M* ± *SD*)	96.09 ± 6.49	94.94 ± 6.54	97.24 ± 6.29
Leg length standing leg (cm) (*M* ± *SD*)	96.18 ± 6.53	94.94 ± 6.59	97.42 ± 6.28

F, female; m, male; M, Mean, SD, standard deviation.

MATERIALS AND METHODS

The study was part of a bigger project considering the influence of physical load on dynamic postural control. The protocol was based on a study by Johnston et al. (2018) that was adapted and systematically replicated. The recorded data was analyzed in different studies considering different aspects (Heil et al., 2020b; Heil and Büsch, 2022). The whole project was conducted in accordance with the Declaration of Helsinki, and the local Ethics committee approved the protocol.

Subjects

For the whole project *a priori* a sample size of n = 126 was determined with a power estimation (F-Test: η^2_p = 0.20, α = 0.01, 1-β = 0.99) for a multivariate three-way mixed analysis of variance (MANOVA) using G*power software (Vers. 3.1.9.7) (Faul et al., 2007). In total, 128 physically active and healthy people mainly normally trained (systematic training for 1–5 years) sport students (64 males and 64 females, age: 23.64 ± 2.44, height: 176.54 ± 8.96 cm, weight: 68.85 ± 10.98 kg) participated in the study (**Table 1**). The subjects were divided into two examination groups. One group completed the loading protocol on a cycle ergometer, and the other group completed the protocol on a treadmill.

To participate in the study, subjects had to fulfill different criteria: (1) No injuries in the lower limbs in the past 6 months. (2) Be able to perform the loading protocol. Ability was checked with the Physical Activity Readiness Questionnaire (PAR-Q) (Warburton et al., 2011). (3) No balance disorders or medication for balance disorders, no cardiovascular disease, no previous reports of chest pain, no neurological diseases, no vestibular or visual impairment, and no chronic ankle instability.

Procedures

All subjects were tested in one 90-min session in a laboratory setting. At first, subjects were informed about the procedures and provided written informed consent to the experiment. Then, in a questionnaire, personal data, sporting background, injury history, laterality (kicking and standing leg), and the questions of the PAR-Q (Warburton et al., 2011) were prompted. Subjects not fulfilling eligibility criteria were excluded from the study.

Before testing, several anthropometric measurements (weight, height, leg length) were conducted. Weight was recorded using the InBody270 (InBody Co., Seoul, Korea), height was measured with a stadiometer (Seca GmbH & Co., KG, Hamburg, Germany), and measuring tape was used to measure leg length of both legs [distance between the subject's anterior-superior iliac spine and the most distal part of the medial malleolus (Gribble and Hertel, 2003)]. Moreover, each subject performed four practice rounds of the YBT to get familiar with the testing procedures.

After a short resting period, testing procedures started. As pre-load measurement, three YBT rounds with 10 min rest in between were performed (20 min pre-load [pre01], 10 min pre-load [pre02], and immediately pre-load [pre03]). Then the subjects completed one of the two loading protocols. After the protocol, the subjects directly went back on the YBT for post-load measurement and had to perform one round of the YBT again.

Instruments

Y-Balance Test

Dynamic postural control was measured with the YBT (Danville, VA, United States).[1] For testing, the subject stands barefoot on a platform with one leg while sliding a block as far as possible in each direction of the YBT (anterior, posteromedial, posterolateral) with the other leg (**Figure 1**). During execution, the subject has the hands on the hips and tries to maintain balance. The reach distance (cm) in each direction is recorded. Dynamic postural control is investigated for the leg with whom the subject is standing on the platform. Trials were controlled according to the formerly published criteria by Plisky et al. (2009). If a trial was considered invalid, the subject must start over again with the current trial.

In the current study, one round of testing consisted of one trial on each leg. Meaning, after sliding the block in each direction on one leg, subjects returned to bilateral stance, switched sides, and conducted the YBT standing on the other leg. The starting leg varied between the subjects. In each group, one half of the subjects started the measurements with their preferred kicking leg, while the other half started with their preferred standing leg.

Loading Protocols

Cycling

As cycling protocol, a modified version of the Wingate Anaerobic-Test (Carey and Richardson, 2003) on a bicycle ergometer (Cyclus 2, RBM elektronik-automation GmbH, Leipzig, Germany) was conducted. Before the protocol started, each subject completed a 5-min warm-up (male: 90 RPM, female: 60 RPM). After a transition phase of 30 s (50–60 RPM), the protocol started, and subjects were instructed to accelerate and maintain their maximal effort for 60 s. Based on a former study from Johnston et al. (2018), the ergometer's resistance was set at 7.5% of the subject's weight. Heart rate was measured during the protocol with a Polar® sensor.

Running

The running protocol was based on a protocol of Schnabel and Kindermann (1983) and performed on a treadmill (PPS 55med-I,

[1] functionalmovement.com

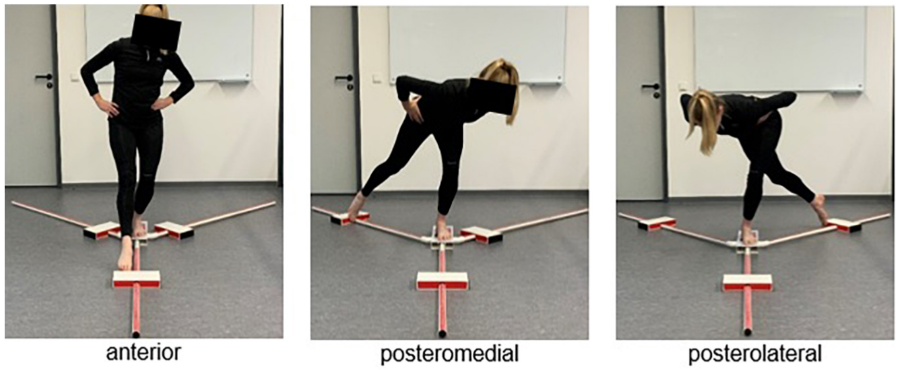

FIGURE 1 | Y-Balance Test.

WOODWAY GmbH, Weil am Rhein, Germany) with a slope of 7.5%. Before the protocol started, each subject completed a 5-min warm-up (8 km/h). Afterward, to determine the individual maximum velocity for the loading protocol, each subject had to complete an incremental test. The incremental protocol started with 8 km/h. Then the velocity was increased by 2 km/h every 20 s until the subject could not run at a certain speed. The final feasible speed was set as the subjects' maximum velocity.

The running protocol started after a 5-min resting period. The treadmill was accelerated to the subject's prior determined individual maximum velocity within 10 s and had a slope of 7.5%. The subjects had to run at this velocity until volitional exhaustion. This led to an average duration of about 60 s, which is comparable to the performed modified Wingate-Anaerobic-Test in cycling. Heart rate was measured with a Polar® sensor during running.

Inter-Limb Asymmetries

Inter-limb asymmetries were calculated with the following three steps:

1. The measured reach distances of the YBT were normalized to leg length using the following equation (Plisky et al., 2009):

$$\text{Normalized reach distance (NRD) (\%)} = \frac{\text{reach distance (cm)}}{\text{leg length (cm)}} \times 100 \quad (1)$$

2. The three pre-load values of the kicking and standing leg were averaged:

$$\text{Mean normalized reach distance (\%)} = \frac{\text{NRD pre01} + \text{NRD pre02} + \text{NRD pre03}}{3} \quad (2)$$

3. The percentage difference between the mean normalized reach distances was calculated according to Bishop et al. (2018a).

$$\text{Percentage difference (PD)} = \frac{100}{\text{max value}} \times \text{min value} \times (-1) + 100 \quad (3)$$

Statistical Analysis

The data were analyzed using SPSS (version 28.0, IBM Corporation, Armonk, NY, United States). At first, the data were checked for normal distribution using the Shapiro-Wilk test. To determine the reliability of the normalized reach distances, the ICC (3,1) with an absolute agreement (Shrout and Fleiss, 1979) was used. ICC values were calculated across the three baseline measurements of the YBT and interpreted according to Koo and Li (2016) as > 0.9 = excellent, 0.75–0.9 = good, 0.5–0.75 = moderate, and < 0.5 = poor. Moreover, within-session reliability was assessed using the CV calculated as CV = [*SD* (trial pre01-pre03)/Mean (trial pre01-pre03)] × 100. CV values < 10% were acceptable according to Cormack et al. (2008). Additionally, to assess the degree of variation between the repeated measures for each leg, the standard error of measurement (SEM) was calculated as SD × $\sqrt{1\text{-ICC}}$.

With the given normalized reach distances, PD was calculated with the maximum and minimum values. Then, two-way mixed ANOVA (load type × time) was conducted to compare the PD between the two points of time (pre vs. post) and between the two load types (cycling vs. running). Significance level was set at $p < 0.01$.

Moreover, the direction of the asymmetries (kicking or standing leg) was determined with an IF Function in Microsoft Excel: *IF (kicking leg > standing leg, 1,2) (Bishop et al., 2021a). Kappa coefficients were calculated to determine the direction of the asymmetries and to analyze how consistently inter-limb asymmetries favored the same leg (kicking or standing leg) before and after the loading protocol. Kappa values were interpreted according to Viera and Garrett (2005), as 0.01–0.20 = slight, 0.21–0.40 = fair, 0.41–0.60 = moderate, 0.61–0.80 = substantial and 0.81–0.99 = almost perfect. Additionally, the subjects were divided into four direction types according to the direction of the asymmetries pre- and post-load: Group 1 (positive/positive), Group 2 (negative/negative), Group 3 (positive/negative), and Group 4 (negative/positive).

To include the direction of the asymmetries directly into the analysis, to display possible changes of the direction and to detect possible asymmetric loading between the legs, PD was also calculated between the kicking leg (kl) and the standing leg (sl).

$$\text{Percentage difference (PD)} = \frac{100}{\text{value kl}} \times \text{value sl} \times (-1) + 100 \quad (4)$$

Then, two-way mixed ANOVA (time × direction type) was conducted with the given PD values (kicking vs. standing leg) to compare the values between the four direction types pre- and post-load.

For all mixed ANOVAs, η^2_p is stated as effect size. Additionally, 90% CIs for the η^2_p were calculated using a SPSS syntax by Wuensch (2016). Moreover, effect sizes for repeated measures (Cohen's d_z) and 95% CIs for the adjacent points of time were also calculated using SPSS.

RESULTS

Regarding an athlete's potential injury risk, only inter-limb asymmetries in the anterior (ANT) reach direction of the YBT are associated with a higher potential injury risk (Plisky et al., 2006; Stiffler et al., 2017; Helme et al., 2021). Therefore, only inter-limb asymmetries in ANT direction will be analyzed.

Statistical Assumptions

The three baseline measurements in ANT direction showed excellent reliability and acceptable variability for the kicking (ICC = 0.96; CV = 2.85%; SEM = 0.34) and the standing leg (ICC = 0.96; CV = 2.66%, SEM = 0.34). According to West et al. (1995), all data were normally distributed due to a given skewness < 2 and kurtosis < 7. With the normalized reach distances, the PD was calculated and analyzed afterward. PD values are provided in **Table 2**. The raw data of the NRD and PD are provided in the **Supplementary Material**.

Physical Load and Load Type

A two-way mixed analysis of variance (time × load type) (**Figure 2**) showed no significant main effect of time [$F_{(1,\ 126)}$ = 4.44, p = 0.04, η^2_p = 0.03] and no interaction between time and load type [$F_{(1,\ 126)}$ = 3.31, p = 0.13, η^2_p = 0.03, 1-β = 0.33]. Between the subjects a significant difference was found between the two load types [$F_{(1,\ 126)}$ = 9.41, p < 0.01, η^2_p = 0.07, 90% CI (0.02, 0.15)].

TABLE 2 | Percentage difference anterior (max vs. min).

		Group	Pre	Post
PD ANT (%)	Total	*M* ± *SD*	4.29 ± 3.47	5.14 ± 4.73
		d [95% CI]		−0.19 [−0.36; −0.10]
		Average change		0.85 ± 4.57
		Change (%)		19.75
	Cycling	*M* ± *SD*	4.89 ± 4.15	6.35 ± 3.92
		d [95% CI]		−0.30 [−0.55; −0.05]
		Average change		1.46 ± 4.86
		Change (%)		29.79
	Running	*M* ± *SD*	3.69 ± 2.53	3.92 ± 4.30
		D [95% CI]		−0.06 [−0.30; 0.19]
		Average change		0.24 ± 4.21
		Change (%)		6.42

ANT, anterior; CI, confidence interval; M, mean value; PD, percentage difference, SD, standard deviation.

Direction of Asymmetries

Regarding the direction of the asymmetries, the results show a moderate level of agreement for all subjects (Kappa = 0.41), moderate agreement for the cycling group (Kappa = 0.47), and a fair level of agreement for the running group (Kappa = 0.36). 90 of 128 subjects favored the same limb pre- and post-load (cycling = 47, running = 43), 38 did not. **Figure 3** illustrates the PD values (kicking vs. standing leg) for the individual subjects pre- and post-load and was prepared using a SPSS syntax provided by Loffing (2022).

According to the direction of the asymmetries pre- and post-load the subjects were divided into four groups: Group 1 (positive/positive) n = 43, Group 2 (negative/negative) n = 47, Group 3 (positive/negative) n = 14 and Group 4 (negative/positive) n = 24. **Table 3** shows the mean values for the four groups for the PD (kicking vs. standing leg) and the difference between pre and post.

A comparison of the PD values (kicking vs. standing leg) pre- and post-load between the different direction types (**Figure 4**) with a two-way mixed ANOVA (time × direction type) showed no significant main effect of time [$F_{(1,\ 124)}$ = 1.51, p = 0.22, η^2_p = 0.01, 1-β = 0.23] but a significant interaction between time and direction type [$F_{(3,\ 124)}$ = 29.99, p < 0.001, η^2_p = 0.42, 90% CI (0.30, 0.50)]. Between the subjects a significant difference was found between the different direction types [$F_{(3,\ 124)}$ = 71.28, p < 0.001, η^2_p = 0.63, 90% CI (0.54, 0.69)].

DISCUSSION

The current study aimed to investigate how inter-limb asymmetries in dynamic postural control change due to physical load. Dynamic postural control was chosen because it is essential in many sporting situations, e.g., cutting, landing, sprinting, or stopping (Güler et al., 2020). Moreover, deficits and inter-limb asymmetries in dynamic postural control are associated with a higher potential risk for lower limb injuries (Plisky et al., 2006; Stiffler et al., 2017; Helme et al., 2021). Therefore, an investigation of (inter-limb asymmetries in) dynamic postural control should be inevitable in injury prevention. Additionally, an investigation of risk factors and inter-limb asymmetries under loaded conditions can provide further information on an athlete's potential injury risk since injuries mainly occur in loaded situations, e.g., during matches, competition, or training (Bourne et al., 2019; Verschueren et al., 2020). Recently, several studies investigated the influence of load on dynamic postural control (Wright et al., 2013; Johnston et al., 2018; Heil et al., 2020b). Nevertheless, up to now, only one study investigated the influence of a physical loading protocol on inter-limb asymmetries in dynamic postural control. Konstantopoulos et al. (2021) found increased inter-limb asymmetries after unilateral jumping. However, nothing

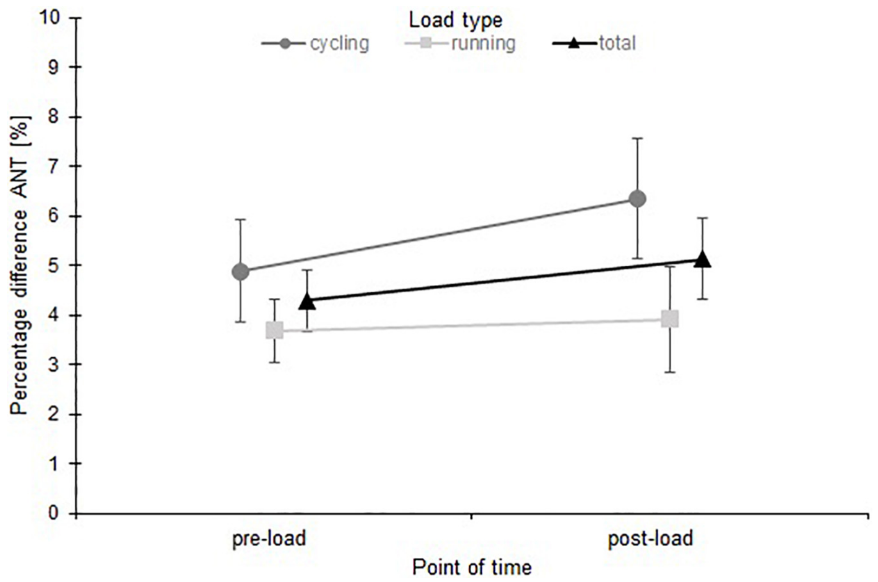

FIGURE 2 | Mean values and 95% confidence intervals of the percentage difference anterior (max vs. min) for both load types (cycling and running) and for the total subject group.

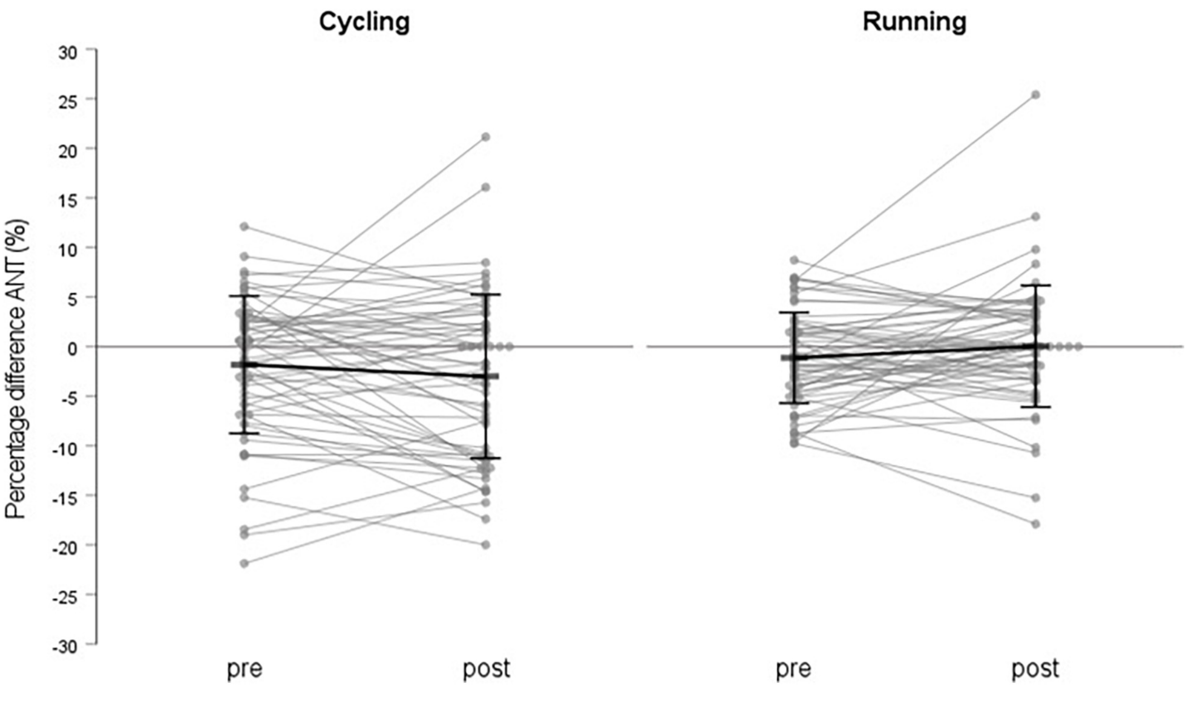

FIGURE 3 | Mean values of percentage difference anterior (kicking vs. standing leg) pre- and post-load for the individual subjects separated between the two load types (cycling and running) and the mean values and standard deviation for the whole groups.

is known about more global protocols better mimicking the real demands of sporting activities. Therefore, in the current study, inter-limb asymmetries of dynamic postural control were regarded and compared before and after physical load. It was hypothesized that inter-limb asymmetries would increase due to physical load.

This hypothesis could not be confirmed, and inter-limb asymmetries did not increase due to physical load. Moreover, two different types of load were compared because several studies regarding the influence of load on dynamic postural control and inter-limb asymmetries have shown that the changes seem to depend on the implemented load type and protocol (Wright et al.,

TABLE 3 | Mean values and standard deviation of the percentage difference anterior (kicking vs. standing leg) for the different direction types.

Direction type		PD (kl vs. sl)	PD (kl vs. sl)	Difference	Difference absolute
Pre	Post	Mean ± *SD* pre	Mean ± *SD* post	(post–pre)	(|post| −|pre|)
Positive*	Positive	3.79 ± 2.91	4.25 ± 5.07	0.46 ± 5.21	0.46 ± 5.21
Negative*	Negative	−6.37 ± 5.07	−7.82 ± 5.42	−1.45 ± 4.71	1.45 ± 4.71
Positive	Negative	2.36 ± 1.61	−5.56 ± 3.62	−7.92 ± 4.10	3.20 ± 3.82
Negative	Positive	−3.56 ± 2.54	3.02 ± 3.86	6.58 ± 4.25	−0.54 ± 4.97

**Positive, NRD kicking leg > NRD standing leg; Negative, NRD kicking leg < NRD standing leg.*
kl, kicking leg; NRD, normalized reach distance; PD, percentage difference; SD, standard deviation; sl, standing leg.

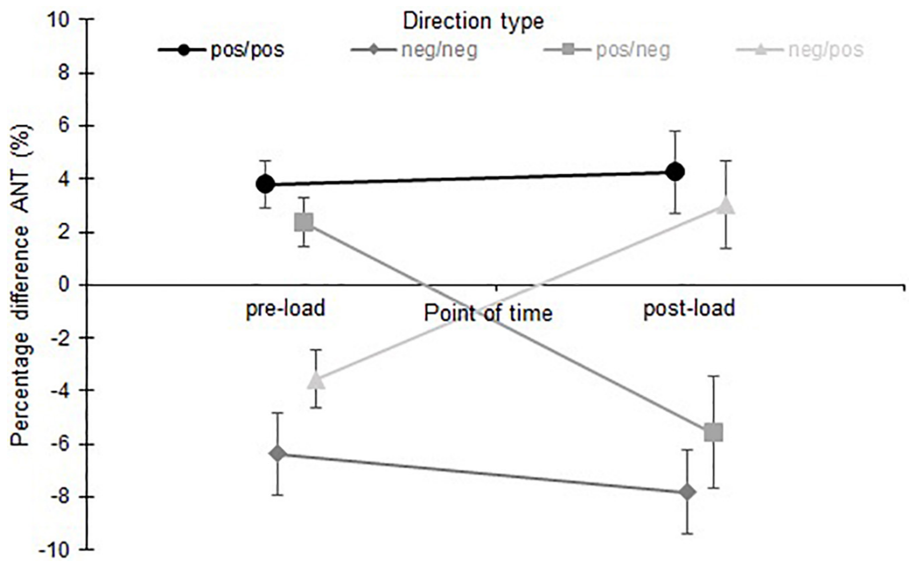

FIGURE 4 | Mean values and 95% confidence intervals of percentage difference anterior (kicking leg vs. standing leg) pre- and post-load for the different direction types (positive/positive, negative/negative, positive/negative, negative/positive).

2013; Heil et al., 2020a; Verschueren et al., 2020). Furthermore, studies often use cycling protocols (Johnston et al., 2018; Heil et al., 2020b; Verschueren et al., 2021), although running protocols seem to be the better choice because they are closer to the real sporting conditions. Therefore, a commonly used cycling protocol was compared to a comparable running protocol. Regarding the results, no differences were found between running and cycling. These findings indicate that the legs were possibly stressed equally due to both types of physical load, which seems not surprising concerning the symmetric/cyclic nature of the chosen loading protocols. However, it must be considered that the normalized reach distances were only decreasing after cycling but not after running (Heil and Büsch, 2022). Therefore, future studies should also take potentially upcoming differences in the internal load between different load types into account. In this context, protocols with longer durations, including changes in intensity and direction, should be used to better mimic the demands of sports, e.g., soccer or handball.

Although no increase in inter-limb asymmetries was found, nevertheless, there is a high variance (**Figure 3**). Moreover, an additional analysis of the direction of the asymmetries using kappa values showed only moderate agreement of the favored limb pre- and post-load. In 38 of the 128 subjects, the direction of the asymmetries changed after the loading protocol (**Table 3**). These results imply that the legs were maybe not equally stressed, although the magnitude of the inter-limb asymmetries was not significantly increasing. Therefore, a more individualized analysis of inter-limb asymmetries concerning the magnitude and direction and their changes due to physical load is indicated. Otherwise, different reactions of the legs might not be revealed and remain undetected only looking at the changes in the magnitude of inter-limb asymmetries.

In this context, it might be helpful to calculate the PD not only between the maximum and minimum value because this method cannot to show changes in the direction between different measurements and different points of time (Parkinson et al., 2021). It might be better to use one leg as a reference leg to evaluate changes in asymmetries and their direction due to training or physical load. Thereby, it is suggested to use a task-specific distinction or to use an inventory of questions or diagnostic tests to determine the preferred leg of an athlete (Dos'Santos et al., 2019; Virgile and Bishop, 2021). Moreover, it might also be beneficial to control/analyze the internal load of the legs during the loading protocol to detect possible differences and to detect asymmetrical loading and the mechanisms behind it. In this context, it could also help to mind possible modifying factors such as injury history, sporting background, or baseline level of asymmetries. Moreover, dynamic postural control is determined and influenced by other factors, e.g., anthropometric characteristics, sex, strength, or mobility (Fusco et al., 2020) and a consideration of these factors might help to detect the reasons for the changes of dynamic postural control and upcoming inter-limb asymmetries or asymmetrical loading.

Considering the given results in terms of injury prevention, the potential injury risk of the athletes is not increased due to higher inter-limb asymmetries. Nevertheless, a more detailed analysis and a consideration of the direction of the asymmetries showed that some subjects, however, possibly stressed their legs differently. Therefore, during injury risk assessment, not only a consideration of the changes in the magnitude of inter-limb asymmetries but also an observation of the changes in the direction of the asymmetries is indicated to reveal inter-limb asymmetries, respectively, asymmetric loading. Thereby, not only changes due to symmetric/cycling loading protocols should be concerned. Therefore, it might be helpful to use protocols with a longer duration, including changes of intensity and/or changes of direction, to create a design and a diagnostic that is more representative for the demands of sports with a higher risk for non-contact injuries (Whyte et al., 2015; Bolt et al., 2021).

Moreover, the current study had some more limitations that must be considered. (1) The current study used the YBT to measure dynamic postural control. Nevertheless, predicting the likelihood of injuries using the YBT has recently been doubted (Luedke et al., 2020). Moreover, the YBT preliminarily measures anticipative dynamic postural control, and in sports, mainly non-anticipative movements are present (Heil et al., 2020b). Therefore, it might be advantageous to use other balance tests besides the YBT to regard an athlete's dynamic postural control. Additionally, a consideration of other elements, such as landing or cutting, could depict more aspects and risk factors occurring under real sporting conditions. (2) Moreover, it might be helpful to take different modifying factors into account. Thereby, a closer look might be taken at an athlete's injury history. For example, it could help to record and concern the side of an athlete's former injuries to distinguish between a healthy and an "injured" leg and to show if the legs still react differently to physical load after a certain period of rehabilitation. (3) Considering an athlete's sporting background might help to detect if participating in a certain sport causes asymmetries and possible asymmetrical loading. Thereby it could be beneficial to differentiate between "symmetric" and "asymmetric" sports. Moreover, testing subjects from only one sport might reduce the number of potential sports conditional factors.

Altogether, no significant load-induced changes of inter-limb asymmetries in dynamic postural control were shown. Therefore, these findings indicate no increase in the potential injury risk due to higher inter-limb asymmetries. Nevertheless, assessing risk factors under physically loaded conditions seems advisable from a theoretical and practical perspective, especially when the high variation between the subjects is regarded. In this context, the current study shows several aspects that should be concerned during injury risk assessment. A consideration of possible modifying factors, such as injury history or sporting background, might provide further information. Moreover, further studies should regard changes in the magnitude and changes in the direction of inter-limb asymmetries to detect possible asymmetric loading. Furthermore, further studies using other protocols and tasks reflecting other aspects of sports are needed to confirm the current findings and to create a more comprehensive picture of an athlete's potential injury risk and possible upcoming inter-limb asymmetries.

CONCLUSION

The current study showed no significant changes of inter-limb asymmetries in dynamic postural control. This indicates that the limbs were possibly equally stressed due to physical load. However, an analysis of the direction of inter-limb asymmetries revealed possible asymmetric loading for some of the subjects. Therefore, it is suggested that changes in the magnitude and the direction of inter-limb asymmetries should be considered to obtain better insights into an athlete's potential injury risk and avoid overlooking asymmetrical loading. Thereby a more individualized analysis of asymmetries and the consideration of possible modifying factors are indicated.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author/s.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Ethics Committee of the Carl von Ossietzky University of Oldenburg (EK/2020/035-02, 24 June 2020). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

The author confirms being the sole contributor of this work and has approved it for publication.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fnhum.2022.824730/full#supplementary-material

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Influence of Controlled Stomatognathic Motor Activity on Sway, Control and Stability of the Center of Mass During Dynamic Steady-State Balance—An Uncontrolled Manifold Analysis

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Multiple sensory signals from visual, somatosensory and vestibular systems are used for human postural control. To maintain postural stability, the central nervous system keeps the center of mass (CoM) within the base of support. The influence of the stomatognathic motor system on postural control has been established under static conditions, but it has not yet been investigated during dynamic steady-state balance. The purpose of the study was to investigate the effects of controlled stomatognathic motor activity on the control and stability of the CoM during dynamic steady-state balance. A total of 48 physically active and healthy adults were assigned to three groups with different stomatognathic motor conditions: jaw clenching, tongue pressing and habitual stomatognathic behavior. Dynamic steady-state balance was assessed using an oscillating platform and the kinematic data were collected with a 3D motion capturing system. The path length (PL) of the 3D CoM trajectory was used for quantifying CoM sway. Temporal dynamics of the CoM movement was assessed with a detrended fluctuation analysis (DFA). An uncontrolled manifold (UCM) analysis was applied to assess the stability and control of the CoM with a subject-specific anthropometric 3D model. The statistical analysis revealed that the groups did not differ significantly in PL, DFA scaling exponents or UCM parameters. The results indicated that deliberate jaw clenching or tongue pressing did not seem to affect the sway, control or stability of the CoM on an oscillating platform significantly. Because of the task-specificity of balance, further research investigating the effects of stomatognathic motor activities on dynamic steady-state balance with different movement tasks are needed. Additionally, further analysis by use of muscle synergies or co-contractions may reveal effects on the level of muscles, which were not visible on the level of kinematics. This study can contribute to the understanding of postural control mechanisms, particularly in relation to stomatognathic motor activities and under dynamic conditions.

Keywords: jaw clenching, tongue pressing, masseter, Posturomed, postural control, UCM, covariation, detrended fluctuation analysis

INTRODUCTION

Balance maintenance and proper body orientation in space are essential for human life. They require a good, reliable and flexible postural control system which is capable of processing multiple sensory feedback inputs from the visual, somatosensory and vestibular systems in the spinal and supraspinal structures of the central nervous system (CNS) in a task dependent manner (Takakusaki, 2017). The control of posture involves control of the body position in space for stability and orientation. Stability is defined as the control of the center of mass (CoM) in relation to the base of support, whereas orientation refers to the ability to maintain an appropriate relationship between the body segments as well as between the body and the environment (Shumway-Cook and Woollacott, 2017). A healthy motor control system modulates the postural movements continuously as a function of the changing tasks. The inability to modulate postural sway, but also environmental or individual constraints may lead to poor performance, instability and falls (Haddad et al., 2013). Furthermore, it has been shown that improved postural control is associated with a decreased risk of falls (Horak, 2006; Rubenstein, 2006) as well as a decreased risk of injury (Hrysomallis, 2007).

Attentional processing is required during postural tasks; therefore, they may reduce the performance of a secondary task when performed simultaneously. On the other hand, a secondary task may improve the postural control by an improved automaticity, an increased arousal or through the utilization of reduced sway for the sake of a better supra-postural task performance (Shumway-Cook and Woollacott, 2017). Previous studies showed that postural control may be influenced by several factors, including motor activity in the stomatognathic motor system (Julià-Sánchez et al., 2020). A frequently cited explanation for this is based on the stimulation of periodontal mechano-receptors that are centrally integrated along with other sensory input and, therefore, facilitates the excitability of the human motor system (Boroojerdi et al., 2000) in a manner similar to the Jendrassik maneuver (Jendrassik, 1885), which in turn increases the neural drive to the distal muscles (Ebben, 2006; Ebben et al., 2008). A variety of studies indicated that stomatognathic motor activity in the form of chewing, tongue activity or different clenching conditions affects human balance and posture under static conditions (Gangloff et al., 2000; Sakaguchi et al., 2007; Hellmann et al., 2011, 2015; Alghadir et al., 2015; Ringhof et al., 2015a,b). Among others, a reduced body sway in the anterior-posterior direction (Hellmann et al., 2011), a reduced variability of muscular co-contraction patterns of posture-relevant muscles of the lower extremities (Hellmann et al., 2015), and reduced trunk and head sway under the influence of controlled biting activities were reported during upright standing (Ringhof et al., 2015a). Furthermore, the review by de Souza et al. (2021) reported that jaw clenching during activities that involve the lower and upper limbs enhance neuromotor stimulation in terms of increased H-reflexes (Miyahara et al., 1996) and stimulate a larger area of the brain. Specifically, a large amount of activity was observed over the frontal, parietal, and temporal cortices and cerebellum during hand grip combined with jaw clenching compared to without jaw clenching (Kawakubo et al., 2014). The authors suggested that the stomatognathic motor system may have effects on the function of remote muscles via cortical activations. Furthermore, a higher excitability of the human motor system during voluntary jaw clenching has also been shown (Boroojerdi et al., 2000). From an evolutionary perspective, it was hypothesized that jaw clenching increases the blood flow to anterior temporal lobe structures during acute activation of the limbic fear circuits (Bracha et al., 2005a). Jaw clenching may increase the blood flow to temporal lobe structures by pumping blood through the temporal bone emissary veins, thus conferring a possible survival advantage during activation of the limbic fear-circuits in expectation of situations requiring the freeze, flight, fight, fright acute fear response (Bracha et al., 2005b). Stomatognathic motor activity also seems to be part of a common physiological repertoire used to improve motor performance during balance recovery tasks (Ringhof et al., 2016). Besides all these facts, it should be mentioned that it stomatognathic motor activity might be of clinical relevance for the prevention of falls. In elderly people there is evidence for an increased risk of falling resulting from an insufficient dental or prosthetic status (Okubo et al., 2010; Mochida et al., 2018).

In contrast to balance under static conditions (e.g., sitting or standing), the influence of stomatognathic motor activity under dynamic conditions (e.g., standing on a balance board or on an oscillating platform) has not yet been investigated in detail (Ringhof et al., 2016). Since the effects found during one balance task may not necessarily be transferable to another balance task (Giboin et al., 2015; Kümmel et al., 2016; Ringhof and Stein, 2018), the question arises whether the effects of stomatognathic motor activity found during static balance tasks would also be observable during dynamic ones. Accordingly, we started to investigate the effects of stomatognathic motor activity in the form of jaw clenching and tongue pressing on dynamic reactive balance performance (Fadillioglu et al., 2022). This was realized by use of an oscillating platform perturbed randomly in one of four horizontal directions. In our previous study, the focus was on the first reactive part of the task. We showed that jaw clenching improved dynamic reactive balance in a task-specific (i.e., direction-dependent) way. The performance improvements found for jaw clenching were associated with lower mean speeds of distinct anatomical regions compared to both the tongue pressing and habitual groups. Subsequent to these findings, the question arises as to whether this performance increase is associated with a changed sway, stability or control of the CoM in the steady-state phase of the task.

The CoM is suggested to be the controlled variable in postural studies (Winter et al., 1998; Kilby et al., 2015; Nicolai and Audiffren, 2019; Richmond et al., 2021), although experimental verification is difficult (Shumway-Cook and Woollacott, 2017). Scholz et al. (2007) used an uncontrolled manifold (UCM) approach to determine if the CoM is the variable which is primarily controlled by the CNS during postural control. They showed that during recovery from a loss of balance, the participants tend to re-establish the position of the CoM rather than those of the joint configurations (Shumway-Cook and Woollacott, 2017), and therefore suggested that the CoM is the key variable controlled by the CNS. In postural control studies, CoM sway is an important parameter (Richmond et al., 2021) and its spatial dynamics can be quantified among others by the total distance covered (Prieto et al., 1996; Richmond et al., 2021). Another important aspect is the temporal dynamics of the sway, since variations in supra-postural activities may lead not only to spatial but also to temporal changes (Chen and Stoffregen, 2012). It was suggested that a detrended fluctuation analysis (DFA) can reveal the temporal dynamics of postural data, specifically to quantify the long-range correlations (or fractality) of the data (Duarte and Sternad, 2008; Stergiou, 2016).

When controlling the body during balance tasks, the CNS has to coordinate a redundant musculoskeletal system (Bernstein, 1967) possessing more degrees of freedom than necessary to achieve the given task (Latash et al., 2002). Different approaches have been suggested to analyze how the CNS treats this redundancy, such as motor programs (Schmidt et al., 2018), optimal control (Todorov and Jordan, 2002) or synergies (d'Avella et al., 2003; Latash et al., 2007; Stetter et al., 2020). Latash et al. (2007) define "synergy" as a neural organization consisting of a multi-element system that organizes sharing of a task among a set of elemental variables (EVs), and ensures the stabilization of a performance variable (PV) through the co-variation of EVs. The fact that different combinations of EVs may result in the same PV indicates that the co-varied behavior provides flexibility for the system. In this context, redundancy is considered not as a problem but as an advantage for the motor control system. According to the motor abundance principle (Gelfand and Latash, 1998), redundancy in the motor control system can be considered positive since the co-variation at the level of the EVs may provide robustness against perturbations (Gera et al., 2010).

The UCM approach (Scholz and Schöner, 1999) is one possibility to quantify the amount of equivalent movement solutions and the degree of stability of the PV. The UCM approach requires a model that relates the changes in EVs to changes in the PV; and ultimately the effects of changes in EVs on the PV are analyzed (Scholz and Schöner, 2014). Both the EVs and PV are chosen on a physiological basis with task-specific considerations. The variability in EVs that results in a changed PV is quantified by the $UCM_{\perp}$ component, whereas it is associated with the $UCM_{\parallel}$ component if the PV remains the same even if the EVs vary over repetitions (Scholz and Schöner, 1999; Latash et al., 2007). The UCM approach has been applied to analyze various motor tasks; for example, reaching and pointing (Tseng et al., 2002; Domkin et al., 2005), pistol shooting (Scholz et al., 2000), sit-to-stand (Scholz et al., 2001; Reisman et al., 2002), parkour jumps (Maldonado et al., 2018), treadmill walking (Verrel et al., 2010; Qu, 2012) and running (Möhler et al., 2019). Kinematic or kinetic data were commonly used as EVs to investigate their effects on the PVs. There is also a number of studies that apply UCM to postural tasks (Krishnamoorthy et al., 2005; Freitas et al., 2006; Hsu et al., 2007, 2013; Hagio et al., 2020). When analyzing postural tasks, the CoM is typically chosen as the PV and joint angles as EVs (Krishnamoorthy et al., 2005; Freitas et al., 2006; Hsu et al., 2007, 2013; Scholz et al., 2007; Hagio et al., 2020). By means of UCM analysis, changes in the variability of coordinated joint movements in association with the stability and control of the CoM have been investigated for various setups with different research questions. In line with the previous studies, a UCM analysis was conducted in this study with stomatognathic motor conditions as the independent variable.

The aim of this study was to investigate the influence of different stomatognathic motor activities (jaw clenching and tongue pressing) on the sway, stability and control of the CoM during a dynamic steady-state balance task (one-legged standing on an oscillatory platform after perturbation). The path length (PL) of the 3D CoM was used to quantify the possible effects of different stomatognathic motor activities on the spatial dynamics of CoM sway, whereas its temporal dynamics was assessed with a DFA. A UCM approach was applied to investigate if and how the co-variation of the joint movements led to the stabilization and control of the CoM, which were quantified by UCM_{Ratio} and $UCM_{\perp}$, respectively. Following the results of our above-mentioned study on the influence of jaw clenching and tongue pressing on dynamic reactive balance performance (Fadillioglu et al., 2022), it was hypothesized that these activities decrease the sway and increase the control and stability of the CoM. Therefore, a decreased PL of the CoM trajectory, an increased alpha of DFA, an increased UCM_{Ratio} and a decreased $UCM_{\perp}$ for the jaw clenching group (JAW) and the tongue pressing group (TON) compared to the group with habitual stomatognathic behavior (HAB) were expected. The findings of this study may contribute to the understanding of postural control, particularly in relation to stomatognathic motor activities and under dynamic conditions.

METHODS

This study comprised a follow-on analysis of the original data set used in Fadillioglu et al. (2022). In the previous study, the reactive phase of the task was analyzed, whereas in the present one, the following steady-state phase is investigated. An a priori power analysis was performed based on the findings of the study (Ringhof et al., 2015a) which analyzed the effects of submaximum jaw clenching on postural stability and on the kinematics of the trunk and head. The analysis revealed that 16 participants per group would be enough to reach the sufficient power (>0.8).

Participants

Forty-eight healthy adults (25 female, 23 male; age: 23.8 ± 2.5 years; height: 1.73 ± 0.09 m; body mass: 69.2 ± 11.4 kg) voluntarily participated in the study after giving written informed consent. All participants completed a questionnaire, confirmed that they were physically active (physical activity 4.6 ± 1.5 days/week and 436 ± 247 min/week) and naive to the tasks on an oscillating platform; had no muscular or neurological diseases; no signs and symptoms of temporomandibular disorders (assessed by means of the RDC/TMD criteria, Dworkin and LeResche, 1992). They presented in good oral health with full dentition (except for 3rd molars) in neutral occlusion. The study was approved by the Ethics Committee of the Karlsruhe Institute of Technology.

Experimental Procedure

Balance Tasks

Dynamic steady-state balance was assessed by means of a Posturomed oscillating platform (Haider-Bioswing, Weiden, Germany), which is a widely-used commercial device to analyze or improve dynamic balance in scientific studies as well as in physiotherapy (Kiss, 2011a; Freyler et al., 2015). It consists of a rigid platform (12 kg, 60 × 60 cm) connected to the main frame by eight 15 cm steel springs with identical stiffness and it can swing in the horizontal plane in all directions. In this study, an automatic custom-made release system was used to initiate mechanical perturbations in four different directions: back (B), front (F), left (L), right (R) (Fadillioglu et al., 2022). By convention, these directions indicate in which direction the platform was accelerated after release of the platform. In each trial, a perturbation in one of the four possible directions was applied in a randomized order. Participants stood on the platform on their dominant leg, which were determined based on self-reports. If the participants were not sure which leg was their dominant leg, it was determined by means of test trials on the Posturomed before the measurements (Ringhof and Stein, 2018). During single-leg stand, they kept their hands placed at the hips and their eyes focused on a target positioned at eye level and 4 m away from the center of the Posturomed (**Figure 1**).

Group Assignment and Masticatory System Statuses

For familiarization, each participant performed two trials without and two trials with perturbation on the Posturomed. After that, a baseline measurement with perturbation and in habitual stomatognathic motor condition was conducted to determine the initial balance performance quantified by Lehr's damping ratio (DR) (Kiss, 2011a). Based on both balance performance and gender, each participant was assigned to one of three groups, such that both gender and the initial level of performance of the groups were balanced. The statistical examination by a one-way ANOVA revealed no baseline performance differences between the three groups ($p = 0.767$). Each group (JAW, TON and HAB) consisted of 16 participants and performed one of the stomatognathic motor conditions simultaneously with the balance task during the measurements (**Table 1**).

The stomatognathic motor activity was recorded by an EMG system (detailed information in the Section Data Collection). To ensure that a force of 75 N was consistently applied, the participants in the JAW group trained just before data acquisition with a RehaBite® (Plastyle GmbH, Uttenreuth, Germany). This medical training device with liquid-filled plastic pads works based on hydrostatic principles, and can be used to control the applied force (Giannakopoulos et al., 2018b). As the participants trained with the RehaBite®, masseter activity was monitored to determine the corresponding muscle activity level associated with a jaw clenching force of 75 N. The determined masseter activity level was around 5% maximum voluntary contraction (MVC) for all participants and was used to control if the submaximal jaw clenching condition was fulfilled.

The TON group similarly trained to apply a submaximal force with the tip of the tongue against the anterior hard palate corresponding to an EMG activity level of the suprahyoid muscles of the floor of the mouth of 5% MVC. Both groups trained for the stomatognathic motor task for 5 min. The HAB did not receive any training or instructions. During the measurements, the JAW group performed the jaw clenching task on an Aqualizer® intraoral splint (medium volume; Dentrade International, Cologne, Germany).

Data Collection

A 3D motion capture system (Vicon Motion Systems; Oxford Metrics Group, Oxford, UK; 10 Vantage V8 and 6 Vero V2.2 cameras with a recording frequency of 200 Hz) was used to record the movements of the Posturomed platform and the participants. Four reflective markers were attached on the upper surface of the Posturomed platform. A further 42 reflective markers were attached to the participants' skin in accordance with the ALASKA modeling system (Advanced Lagrangian Solver in kinetic Analysis, Insys GmbH, Chemnitz, Germany; Härtel and Hermsdorf, 2006). Before data acquisition, 22 anthropometric measures were manually taken from each participant for the ALASKA modeling.

A wireless EMG system (Noraxon, Scottsdale, USA; recording frequency of 2,000 Hz) was used to measure EMG activity of the masseter for JAW and HAB; and of the suprahyoid muscles of the floor of the mouth measured in the region of the digastricus venter anterior muscle for TON. As preparation, the skin over the corresponding muscles was carefully shaved, abraded and rinsed with alcohol. Bipolar Ag/AgCl surface electrodes (diameter 14 mm, center-to-center distance 20 mm; Noraxon Dual Electrodes, Noraxon, Scottsdale, USA) were positioned in accordance with the European Recommendations for Surface Electromyography (Hermens et al., 1999). Afterwards, MVC tests were performed.

For each trial, participants received standardized instruction about the task to be performed and were asked to compensate the perturbation as quickly as possible and to stabilize their body. Between each trial, participants had 2 min of rest to prevent fatigue. Measurements ended after completion of 12 successful balance task trials (i.e., three trials for each direction) each lasting 20 s after initiation of the perturbation. During the measurements, EMG activity of the masseter (for the JAW group) or the suprahyoid muscles of the floor of the mouth (for the TON group) was monitored and compared with the individually determined EMG activity level corresponding to 5% of MVC. Recordings were stopped and the trial was considered invalid if participants stopped performing their stomatognathic motor task (JAW and TON), had ground contact with the non-standing foot, changed the placement of their standing foot, released one of the hands from the hip or lost their balance. All data were recorded in Vicon Nexus 2.10; where the EMG system was connected via the Noraxon plug-in.

Data Analysis

The collected data of 576 trials (48 participants × three valid trials × four directions) were exported from Vicon Nexus for further analysis in MATLAB R2020a (MathWorks; Natick, USA). For all data, the R and L directions were re-sorted as ipsilateral (I) and contralateral (C) according to the standing leg of the participants.

FIGURE 1 | Participant during single-leg stand on the Posturomed oscillating platform.

TABLE 1 | Stomatognathic motor conditions of the three groups, JAW, TON, and HAB.

JAW: instructed, controlled submaximal jaw clenching with a 75 N force—activity of the masticatory muscles during simultaneous occlusal loading
TON: instructed, controlled submaximal tongue pressing against the palate—stomatognathic muscle activity without occlusal loading
HAB: habitual stomatognathic behavior—jaw positioning without any instruction

The marker data were filtered with a Butterworth low-pass filter (fourth-order; cut-off frequency 10 Hz); and the EMG data with a Butterworth band-pass filter (fourth-order; cut-off frequency 10–500 Hz). The EMG data were then rectified and smoothed by averaging with a sliding window of 30 ms and normalized to the MVC amplitudes (Hellmann et al., 2011).

Based on Posturomed marker data, two critical time points were separately determined for each trial: (1) the start of the perturbation, and (2) the third highest amplitude of the Posturomed center in the direction of perturbation (Kiss, 2011a). The time span between (1) and (2) was assumed to be the phase in which the perturbation was maximally compensated, and the time frames after (2) were considered the dynamic steady-state phase of the movement. For all calculations, a time window of 12 s (Müller et al., 2004) was used which started at time point (2).

Spatial Dynamics of CoM Sway

To quantify the spatial dynamics of the CoM sway, the PL was calculated. The time series of CoM position was estimated by the subject-specific anthropometric 3D model referenced and explained below (see Section Uncontrolled Manifold Approach). The point-by-point Euclidean norm of the vectors containing the 3D coordinates of the CoM was calculated to convert the three components in the x, y and z coordinates into a single value k, where i stands for the frame number (Equation 1). PL was approximated by the sum of the distances between consecutive points of the time series k with a length of n (Prieto et al., 1996), where n equals the total number of frames in a 12 s interval ($n =$ 2,400; Equation 2).

$$k_i = \sqrt{{x_i}^2 + {y_i}^2 + {z_i}^2};\ where\ \ i = 1, 2, 3, \ldots, n \tag{1}$$

$$PL = \sum_{i=1}^{n-1} \left|k_{i+1} - k_i\right| \tag{2}$$

Temporal Dynamics of CoM Sway

A detrended fluctuation analysis was performed to quantify the temporal dynamics of CoM sway. Firstly, an integrated time series was calculated by subtracting its mean from it (Equation 3). Secondly, the data were divided into non-overlapping segments of length m and the linear approximation was estimated by a separate least square fit in each segment. Thirdly, average fluctuation of the time series around the trend was calculated as given in Equation (4). The last two steps were repeated for all the considered m.

$$y\left(b\right) = \sum_{i=1}^{b} \left(k\left(i\right) - k_{avg}\right) \tag{3}$$

$$F\left(m\right) = \sqrt{\frac{1}{n} \sum_{b=1}^{n} \left(y\left(b\right) - y_m(b)\right)^2} \tag{4}$$

In general $F(m)$ increases with the increasing m and a power law is expected where the scaling component α is a constant (Equation 5). If $\alpha < 0.5$ or $1 < \alpha < 1.5$, the time series interpreted as anti-persistent, where a smaller α indicates a more anti-persistent behavior. If $0.5 < \alpha < 1$ or $1.5 < \alpha < 2$, the time series is persistent and the larger the α, the more persistent is the time series (Lin et al., 2008).

$$F\left(m\right) \propto m^{\alpha} \tag{5}$$

Uncontrolled Manifold Approach

A UCM approach was applied to investigate if and how the co-variation of the joint movements led to the stabilization and control of the CoM. In accordance with the literature, the CoM and the joint angles were selected as PV and EVs, respectively (Krishnamoorthy et al., 2005; Freitas et al., 2006; Hsu et al., 2007, 2013; Scholz et al., 2007; Hagio et al., 2020). To obtain joint angles, an inverse kinematics calculation was performed using a modified version of the full-body Dynamicus (ALASKA) model (Härtel and Hermsdorf, 2006). A subject-specific anthropometric 3D model was used to estimate the CoM as the weighted sum of the body segments (Möhler et al., 2019).

The model was a modified version of the Hanavan model and had 50 degrees of freedom (Hanavan, 1964). Of the 36 subject-specific anthropometric measurements needed to calculate the CoM according to this model, 21 were taken manually and 15 were determined from the reflective markers. A constant density was assumed (Ackland et al., 1988) and the mass of each segment was estimated by volume integration. The whole-body CoM in 3D, r_{CoM} , was determined by calculating the weighted sum of the body segments using Equation (6), where N is the total number of segments ($N = 17$; V_m the volume of the m^{th} segment; and r_m the center of gravity vector of the m^{th} segment.

$$r_{CoM} = \frac{1}{\sum_{m=1}^{N} V_m} * \sum_{m=1}^{N} r_m V_m \tag{6}$$

The CoM, as the PV for the UCM, was defined as a function of the joint angles as the EVs ($r_{CoM} = f\left(\theta_1, \theta_2, \ldots, \theta_j\right)$, where j stands for the number of EVs). The mean joint configuration across each trial, θ^0, was calculated as an approximation of the desired configuration (Latash et al., 2007). The Jacobian matrix, $J(\theta^0)$, containing all first-order partial derivatives of the CoM coordinates with respect to the joint angles, was calculated at this reference joint configuration. Afterwards the null space of the matrix was computed as the linear estimate of the UCM

(Equation 7). The null space of the Jacobian matrix is the linear subspace of all joint angle combinations that does not affect the position of the CoM, and it is spanned by *j-d* number of basis vectors ε_i, where *j* and *d* are the number of dimensions of EVs and PV, respectively (Scholz and Schöner, 1999).

$$0 = J(\theta^0) \cdot \varepsilon_i \tag{7}$$

At each instant of $n = 2{,}400$ trials ($t = 12$ s, recording frequency 200 Hz), the deviation from the mean joint configuration ($\theta - \theta^0$) was calculated (Scholz et al., 2007; Hsu et al., 2013) and subsequently resolved into their projection on the null space to decompose it into the parallel, $\theta_{\parallel}$, and orthogonal, $\theta_{\perp}$, components (Scholz and Schöner, 1999; Möhler et al., 2019) (Equations 8, 9).

$$\theta_{\parallel} = \sum_{i=1}^{j-d} \varepsilon_i^T \left(\theta - \theta^0\right) \varepsilon_i \tag{8}$$

$$\theta_{\perp} = \left(\theta - \theta^0\right) - \theta_{\parallel} \tag{9}$$

Finally, the amount of variability parallel to the UCM ($UCM_{\parallel}$, i.e., stabilizing PV) and orthogonal to the UCM ($UCM_{\perp}$, i.e., changing PV) were estimated (Scholz and Schöner, 1999) (Equations 10, 11). UCM_{Ratio}, the ratio of the two UCM components was calculated as suggested by Papi et al. (2015) to obtain a symmetrical distribution (i.e., [−1 1]. The midpoint 0 is the threshold for "synergy" and therefore appropriate for statistical calculations (Equation 12).

$$UCM_{\parallel} = \sqrt{\frac{1}{n \cdot (j-d)} \sum_{i=1}^{n} \theta_{\parallel i}^2} \tag{10}$$

$$UCM_{\perp} = \sqrt{\frac{1}{n \cdot d} \sum_{i=1}^{n} \theta_{\perp i}^2} \tag{11}$$

$$UCM_{Ratio} = \frac{2 \cdot UCM_{\parallel}{}^2}{UCM_{\parallel}{}^2 + UCM_{\perp}{}^2} - 1 \tag{12}$$

The $UCM_{\parallel}$ component is a measure of the co-variation of EVs and therefore a measure for flexibility. A higher $UCM_{\parallel}$ indicates a higher variability on the level of the EVs that does not change the PV, and therefore a more flexible behavior of the control system, and vice versa. The $UCM_{\perp}$ component is a measure for control of the PV. The higher the $UCM_{\perp}$, the less controlled the PV, which in this study is the CoM. Lastly, UCM_{Ratio} indicates the stability of the PV by means of kinematic synergy of the EVs. A $UCM_{Ratio} > 0$ is interpreted as a synergy, whereas a $UCM_{Ratio} \leq 0$ indicates no synergy (Latash et al., 2007). In this study, UCM_{Ratio} and $UCM_{\perp}$ were used to quantify the stability and control of the CoM (i.e., the PV), respectively.

Statistics

Statistical analysis was performed with IBM SPSS Statistics 25.0 (IBM Corporation, Armonk, NY, USA). The PL of the CoM, DFA scaling component and two UCM parameters ($UCM_{\perp}$, UCM_{Ratio}) for three trials for each direction and for each subject were averaged. A Kolmogorov-Smirnov test was conducted to determine the normality of data distribution.

Each of the four directions of perturbation was analyzed separately because postural response may differ depending on the perturbation direction (Kiss, 2011b; Nonnekes et al., 2013; Chen et al., 2014; Freyler et al., 2015; Akay and Murray, 2021). For each of the four considered parameters and for each direction, a one-way ANOVA or a Kruskal-Wallis test was performed for the group comparisons depending on the normality of the distribution. The level of significance was set *a priori* to $p < 0.05$. Partial eta squared (η_p^2) or Cramer's V (ϕ_c) (small effect: $\eta_p^2 < 0.06$ or $\phi_c < 0.2$; medium effect: $0.06 < \eta_p^2 < 0.14$ or $0.2 < \phi_c < 0.6$; large effect: $\eta_p^2 > 0.14$ or $\phi_c > 0.6$; Cohen, 1988; Richardson, 2011) were calculated to estimate the effect sizes for normal and non-normal distribution of data, respectively.

RESULTS

Sway of the Center of Mass

The operationalization of CoM sway in relation to the different stomatognathic motor conditions was analyzed by the PL of the 3D CoM trajectory (**Table 2**). The PL results did not show any significant changes between different stomatognathic motor conditions in the four perturbation directions. Although B, I and C had small effect sizes, F had a medium effect size (B: $p = 0.429$, $\eta_p^2 = 0.037$; F: $p = 0.182$, $\eta_p^2 = 0.073$; I: $p = 0.461$, $\eta_p^2 = 0.034$; C: $p = 0.692$, $\eta_p^2 = 0.016$).

Detrended Fluctuation Analysis

Temporal dynamics of CoM sway was analyzed with a DFA (**Table 2**). The scaling components did not differ significantly between different stomatognathic motor conditions in the four perturbation directions (B: $p = 0.103$, $\eta_p^2 = 0.096$; F: $p = 0.724$, $\eta_p^2 = 0.014$; I: $p = 0.821$, $\eta_p^2 = 0.009$; C: $p = 0.689$, $\eta_p^2 = 0.016$).

Uncontrolled Manifold Analysis

A UCM analysis was performed aiming at analyzing the co-variation of joint angles in relation with the control as well as the stability of the CoM. The $UCM_{\perp}$ and UCM_{Ratio} components were utilized to quantify the control and the stability of the CoM, respectively. The results are represented in **Table 2**.

For the $UCM_{\perp}$ component, the groups did not show any significant differences in any of the perturbation directions and all the effect sizes were small (B: $p = 0.305$, $\phi_c = 0.157$; F: $p = 0.466$, $\eta_p^2 = 0.033$; I: $p = 0.947$, $\eta_p^2 = 0.002$; C: $p = 0.514$, $\eta_p^2 = 0.029$). This indicated the control of the CoM was not affected by the stomatognathic motor conditions (i.e., JAW, TON, and HAB).

Regarding the UCM_{Ratio}, the groups did not differ significantly in any of the perturbation directions and all of the results showed small effect sizes (B: $p = 0.865$, $\eta_p^2 = 0.006$; F: $p = 0.333$,

TABLE 2 | The UCM, the path length and the DFA scaling exponent (α) results are shown as mean ± standard deviation.

UCM$_\perp$ in rad^2/dof	**JAW**	**TON**	**HAB**	***p***	**η_p^2 or ϕ_c**
B	0.0134 ± 0.0124	0.0125 ± 0.0052	0.0134 ± 0.0064	0.305*	0.157*
F	0.0107 ± 0.0041	0.0127 ± 0.0060	0.0126 ± 0.0052	0.466	0.033
I	0.0173 ± 0.0144	0.0161 ± 0.0085	0.0163 ± 0.0077	0.947	0.002
C	0.0178 ± 0.0113	0.0149 ± 0.0062	0.0179 ± 0.0059	0.514	0.029
UCM$_{Ratio}$	**JAW**	**TON**	**HAB**	***p***	**η_p^2**
B	0.2092 ± 0.3179	0.2362 ± 0.2020	0.1852 ± 0.2710	0.865	0.006
F	0.2516 ± 0.3194	0.2204 ± 0.3176	0.1022 ± 0.2482	0.333	0.048
I	0.2492 ± 0.2663	0.2967 ± 0.1652	0.2095 ± 0.2653	0.585	0.024
C	0.1791 ± 0.4335	0.2373 ± 0.2096	0.1634 ± 0.2664	0.788	0.011
Path length in mm	**JAW**	**TON**	**HAB**	***p***	**η_p^2**
B	325.21 ± 174.28	408.17 ± 277.49	329.38 ± 119.33	0.429	0.037
F	267.15 ± 112.13	381.38 ± 253.10	304.44 ± 124.28	0.182	0.073
I	369.18 ± 186.20	423.32 ± 219.60	344.73 ± 125.14	0.461	0.034
C	366.66 ± 154.83	428.83 ± 295.28	395.59 ± 117.34	0.692	0.016
Scaling component, α	**JAW**	**TON**	**HAB**	***p***	**η_p^2**
B	1.68 ± 0.12	1.74 ± 0.09	1.76 ± 0.11	0.103	0.096
F	1.73 ± 0.08	1.75 ± 0.09	1.73 ± 0.08	0.724	0.014
I	1.73 ± 0.08	1.72 ± 0.08	1.74 ± 0.09	0.821	0.009
C	1.72 ± 0.09	1.70 ± 0.10	1.71 ± 0.08	0.689	0.016

The p-values and the effect sizes for group comparisons are represented in the last two columns. The asterisks (*) indicate the Kruskal-Wallis and Cramer's V (ϕ_C) calculations [otherwise a one-way ANOVA and partial eta squared (η_p^2) were calculated]. The level of significance was set a priori to $p < 0.05$. JAW, jaw clenching; TON, tongue pressing; HAB, habitual; B, backwards; F, forwards; I, Ipsilateral; C, contralateral; dof, degrees of freedom.

$\eta_p^2 = 0.048$; I: $p = 0.585$, $\eta_p^2 = 0.024$; C: $p = 0.788$, $\eta_p^2 = 0.011$). These results showed that the stability of the CoM was not affected by the stomatognathic motor conditions (i.e., JAW, TON, and HAB).

DISCUSSION

The aim of this study was to investigate the effects of different stomatognathic motor conditions on the sway, control and stability of the CoM during a dynamic steady-state balance task. The PL of the 3D CoM, a DFA and a UCM analyses were used to quantify the variables of interest. It was hypothesized that jaw clenching and tongue pressing decrease the total sway, increase the persistency of CoM fluctuations, increase both the control and stability of the CoM. Inclusion of the TON group would enable a differentiation between the specific effects of jaw clenching with occlusal load from the effects of stomatognathic motor activity in general, as well as from the dual-task effects. This could ultimately help to understand if the possible modulations of posture during jaw clenching occur basically due to a supra-postural task or stomatognathic activities in general; or if any additional functional interactions such as higher excitability of the human motor system (Boroojerdi et al., 2000), muscle co-contractions (Giannakopoulos et al., 2018a) or H-reflex responses (Miyahara et al., 1996) may exist. None of the considered parameters showed significant group effects in any of the directions. Based on these results, it can be concluded that deliberate jaw clenching or tongue pressing do not have significant effects on the control or stability of the CoM compared to habitual stomatognathic motor conditions in the steady-state phase of the task. At least, the effects could not be quantified by the used parameters. In contrast to the previously-found effects of jaw clenching on dynamic reactive balance performance (Fadillioglu et al., 2022), the findings in this study do not indicate any task-specific effects of stomatognathic motor activities on dynamic steady-state balance assessed by an oscillating platform. Because of the task-specificity of balance (Giboin et al., 2015; Kümmel et al., 2016; Ringhof and Stein, 2018), further research investigating the effects of stomatognathic motor activities on dynamic steady-state balance with different movement tasks are needed. To the best of our knowledge, the present study is the first to investigate the effects of stomatognathic motor activity on dynamic steady-state balance on an oscillating platform.

Quantification of CoM Sway

The PL of the 3D CoM position was calculated to quantify the distance covered by the CoM during the trials. The results in this study revealed no significant differences between the three groups for any of the directions. Nevertheless, the effect sizes for the direction F were medium ($p = 0.182$; $\eta_p^2 = 0.073$), where the JAW group had a slightly smaller PL compared to TON and HAB, indicating a higher dynamic steady-state stability. It should be noted that significant performance increases and decreased anatomical region speeds were seen in the F direction during the early reactive phase of the task (Fadillioglu et al., 2022). Although a medium effect size does not have a high statistical power, jaw clenching may have effects on the steady-state stability; but these were not high enough to be detected with the chosen methods.

The temporal dynamics of CoM sway was analyzed by a DFA, which did not show any significant differences between groups. Overall, the scaling exponent α was larger than 1.5 for all directions and all groups, indicating a Brownian noise (Stergiou, 2016). These results were slightly higher but similar to those of Liang et al. (2017) , which considered the CoM instead of center of pressure for DFA (Lin et al., 2008; Blázquez et al., 2010; Munafo et al., 2016) as in the present study. On the other hand, even though DFA has become a predominant method for fractal analysis, it may not provide useful results for short time series (Stergiou, 2016).

Analysis of Control and Stability of the CoM by a UCM Approach

A UCM approach was applied to investigate the co-variated movement of joints in relation to the CoM position as the PV (Krishnamoorthy et al., 2005; Freitas et al., 2006; Hsu et al., 2007, 2013; Scholz et al., 2007; Hagio et al., 2020) under different stomatognathic motor conditions. In this study, the $UCM_{\perp}$ and the UCM_{Ratio}were directly included in the analysis, whereas the $UCM_{\parallel}$ was only indirectly considered within the UCM_{Ratio}. The findings indicate that jaw clenching or tongue pressing do not lead to any effects that are quantifiable with the UCM approach.

The UCM_{Ratio}component was used to investigate the stabilization of the CoM through co-varied movements of the joint angles. The statistical analysis revealed that the three groups did not differ in UCM_{Ratio}. This showed that jaw clenching or tongue pressing did not lead to a more stable CoM compared to habitual stomatognathic motor conditions in the steady-state phase of the task. Therefore, it contradicted our first hypothesis regarding the stability of the CoM.

The $UCM_{\perp}$ component was used to quantify the control of the CoM. The results showed that the groups did not differ in terms of control of the CoM, which suggested that jaw clenching or tongue pressing did not lead to a better control compared to habitual conditions. Based on this result, the second hypothesis was rejected.

A UCM analysis was performed in the present study and a subject-specific anthropometric 3D model was used to estimate the CoM (Möhler et al., 2019). Therefore, the model covered all three movement planes and considered not only the lower body but also the upper body, which plays an important role in the movement of the CoM due to its high mass.

Effects of Masticatory System on Dynamic Stability

As already described in the introduction, there are some phenomena described in the literature that support the assumption of a close functional integration of the masticatory system in the human motor control processes (Miyahara et al., 1996; Boroojerdi et al., 2000; Bracha et al., 2005a,b; Julià-Sánchez et al., 2020). This could be because jaw movements are proportionally driven by anterior neck muscles, which inevitably requires co-contractions of the lateral and posterior neck muscles (Giannakopoulos et al., 2018a). The resulting movements of the head must in turn be matched centrally with the further proprioceptive input of postural control. Therefore, the integration of the stomatognathic system into postural control is not a phenomenon, but a basic requirement.

Jaw clenching during activities that involve the lower and upper limbs may enhance neuromotor stimulation by means of the H-reflex, and therefore increase the excitability of the motor system (de Souza et al., 2021). Furthermore, high activity was observed in the frontal, parietal, and temporal cortices and cerebellum during hand grip combined with jaw clenching compared to without jaw clenching (Kawakubo et al., 2014). In addition, there are also studies revealing that not only stomatognathic activity but also the use of occlusal splints (Battaglia et al., 2018) or mouthguards (Dias et al., 2022) influence the strength in the muscles of the other body parts. These findings indicate that there is a close relationship between the masticatory system and muscle strength or physical exertion. Although it is hard to verify the underlying mechanisms experimentally, based on these findings it was hypothesized that jaw clenching may lead to better dynamic steady-state stability also under dynamic conditions. However, the results in this study did not support this hypothesis.

Consideration of Methodical Aspects

Based on their gender and baseline performance, the participants were allocated into one of the three groups (JAW, TON, and HAB). The statistical examination by ANOVA revealed that there were no baseline performance differences between the three groups ($p = 0.767$). The purpose of building three groups with different stomatognathic motor conditions, instead of making all participants perform all the tasks, was due to three main reasons. Firstly, "habitual" in this study meant that no instruction was given regarding the stomatognathic motor activity. Therefore, an unconscious, natural behavioral pattern of the masticatory system was ensured. An instructed "habitual" would not be physiologically possible, because an "instructed" behavioral pattern cannot lead to an unconsciously performed behavior. Secondly, possible carry over effects between different stomatognathic motor tasks were avoided. After jaw clenching or tongue pressing, there could be sustained physiological effects such as an increased excitability of the motor system. Thirdly, fatigue effects were avoided. If all tasks were performed for each of the four directions separately, 36 valid trials would be needed. Considering the invalid trials as well, the total number performed would be too high.

In this study, the Posturomed, an oscillating platform, was used to assess dynamic balance performance. The platform was randomly perturbed in one of the four different directions. The perturbation directions were analyzed independently following the suggestions of Freyler et al. (2015), because muscle spindles provide different information depending on the direction as well the velocity of perturbations (Akay and Murray, 2021). In addition to this, the direction of surface translation is an important factor for the sensation, processing and output of the postural responses (Nonnekes et al., 2013; Freyler et al., 2015). Although it was suggested that the perturbation direction may not matter during the steady-state phase of the balancing task on an oscillating platform (Giboin et al., 2015), in this study the directions were analyzed separately since the research question was to further investigate the positive effects of jaw clenching, which was found only in one direction (Fadillioglu et al., 2022).

The focus was put on the CoM in this study because it is suggested to be the controlled variable in postural studies (Winter et al., 1998; Kilby et al., 2015; Nicolai and Audiffren, 2019; Richmond et al., 2021). Also, in studies assessing dynamic stability by means of an oscillating platform, the CoM was considered as an important variable (Pfusterschmied et al., 2013; Pohl et al., 2020). Even if it is widely chosen for postural studies and others from the field of motor control (Scholz et al., 2000, 2001; Reisman et al., 2002; Tseng et al., 2002; Domkin et al., 2005; Verrel et al., 2010; Qu, 2012; Maldonado et al., 2018; Möhler et al., 2019), it does not prove that it is the single right one. Another possible PV could be the distance between the CoM and the center of the platform.

Limitations

All the participants in this study were physically active adults. The homogeneity of this group helped to minimize altered postural control mechanisms due to, for example, age (Henry and Baudry, 2019) or neurological disorders (Delafontaine et al., 2020). Nevertheless, it should be noted that the findings cannot be directly transferred to other groups (e.g., elders or people with neurological disorders).

The stabilization of a moving platform and the stabilization of the body on a rigid surface are different balance tasks (Alizadehsaravi et al., 2020). Therefore, it should be added that the findings in this study may not be valid for balance tasks on stationary ground or for other dynamic tasks (Giboin et al., 2015; Kümmel et al., 2016; Ringhof and Stein, 2018).

It is possible that the UCM approach and the model used in the study were not sensitive enough to capture the possible effects due the different stomatognathic motor activities. Therefore, further research investigating the effects of stomatognathic motor activities on dynamic steady-state balance with other models could be useful. Additionally, further analysis by use of muscle synergies (d'Avella et al., 2003) or co-contractions (Hellmann et al., 2015; Munoz-Martel et al., 2019) may reveal effects on the level of muscles, which were not visible on the level of kinematics.

CONCLUSION

To the best of our knowledge, this study investigates for the first time the effects of different stomatognathic motor conditions (jaw clenching, tongue pressing and habitual condition) on dynamic steady-state balance. The aim was to analyze the effects particularly on the sway, control and stability of the CoM during a dynamic steady-state task (standing one-legged on an oscillating platform). The results revealed that deliberate jaw clenching or tongue pressing do not seem to affect the sway, the control or the stability of the CoM during a dynamic balance task on an oscillating platform. Due to the task-specificity of balance, further research investigating the effects of stomatognathic motor activities on dynamic steady-state balance with different movement tasks is needed. In addition, further analysis by use of muscle synergies or co-contractions may reveal effects on the level of muscles, which were not visible on the level of kinematics. This study can contribute to the understanding of postural control mechanisms, particularly in relation to stomatognathic motor activities.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics Committee of the Karlsruhe Institute of Technology. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

CF and LK conducted the experiment. FM provided the toolbox for data processing. CF carried out data analysis and took the lead in writing the manuscript. All authors were involved in the interpretation and discussion of the results, provided critical feedback. All authors contributed to the article and approved the submitted version.

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Balance Adaptation While Standing on a Compliant Base Depends on the Current Sensory Condition in Healthy Young Adults

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Background: Several investigations have addressed the process of balance adaptation to external perturbations. The adaptation during unperturbed stance has received little attention. Further, whether the current sensory conditions affect the adaptation rate has not been established. We have addressed the role of vision and haptic feedback on adaptation while standing on foam.

Methods: In 22 young subjects, the analysis of geometric (path length and sway area) and spectral variables (median frequency and mean level of both total spectrum and selected frequency windows) of the oscillation of the centre of feet pressure (CoP) identified the effects of vision, light-touch (LT) or both in the anteroposterior (AP) and mediolateral (ML) direction over 8 consecutive 90 s standing trials.

Results: Adaptation was obvious without vision (eyes closed; EC) and tenuous with vision (eyes open; EO). With trial repetition, path length and median frequency diminished with EC ($p < 0.001$) while sway area and mean level of the spectrum increased ($p < 0.001$). The low- and high-frequency range of the spectrum increased and decreased in AP and ML directions, respectively. Touch compared to no-touch enhanced the rate of increase of the low-frequency power ($p < 0.05$). Spectral differences in distinct sensory conditions persisted after adaptation.

Conclusion: Balance adaptation occurs during standing on foam. Adaptation leads to a progressive increase in the amplitude of the lowest frequencies of the spectrum and a concurrent decrease in the high-frequency range. Within this common behaviour, touch adds to its stabilising action a modest effect on the adaptation rate. Stabilisation is improved by favouring slow oscillations at the expense of sway minimisation. These findings are preliminary to investigations of balance problems in persons with sensory deficits, ageing, and peripheral or central nervous lesion.

Keywords: balance, adaptation, stance, repeated trials, sensory conditions, compliant surface, vision, touch

INTRODUCTION

Standing quietly may not require a particular neural effort (Morasso and Schieppati, 1999; Ivanenko and Gurfinkel, 2018) such that even a cognitive task is performed better while standing than sitting in young healthy subjects (Rosenbaum et al., 2017). Understandably, the metabolic cost of quiet stance is low in healthy subjects unless they have to counteract perturbations (Houdijk et al., 2009). However, standing in critical conditions, such as in tandem stance or on foam, is an attention-demanding task, where the cognitive task may affect the level of instability (Boisgontier et al., 2013; Honeine et al., 2017). Further, maintaining the equilibrium in critical conditions requires continuous activity in several muscles (Schieppati et al., 1995; Tokuno et al., 2007; Kelly et al., 2012; Sozzi et al., 2013), implying a considerable effort. When standing on foam, muscle activity and metabolic cost increases (Fransson et al., 2007; Mohapatra et al., 2014; Mademli et al., 2021) and body sway increases concurrently (Teasdale et al., 1991; Patel et al., 2011; Anson et al., 2019; Hsiao et al., 2020).

Keeping the body vertical, the ultimate goal of the control of upright stance (Hlavačka et al., 1996; Jahn and Wühr, 2020) is achieved thanks to continuous firing of the receptors activated by gravity, movement of body segments and centre of mass, and by changes in the relation between foot sole and basis of support (Mittelstaedt, 1998; see Felicetti et al., 2021). An inventory of all responsible receptors is implausible. One can only hope to itemise most of those for which reports are available, based on experiments conducted in quiet and perturbed stance conditions (Borel et al., 2008; Creath et al., 2008; Pettorossi and Schieppati, 2014; Cheung and Schmuckler, 2021; Nedelkou et al., 2021). In the same vein, it is virtually impossible to indicate all the muscles more or less active when standing on a compliant surface or under similarly critical postures, such as standing in tandem or on one leg. Standing on foam requires continuous adjustments of all body segments, aiming at achieving core stability by appropriate recruitment of muscle activity (Grüneberg et al., 2005; Gurfinkel et al., 2006; Sozzi et al., 2013; Küng et al., 2009). Proprioceptive inflow must be strongly enhanced when standing on foam, including the input from the intrinsic foot muscles (Schieppati et al., 1995; Abbruzzese et al., 1996; Forestier et al., 2015). A proprioception-based mechanism is at the origin of implicit learning and adaptation (Fitzpatrick et al., 1992; Rosenkranz and Rothwell, 2012; see, in a different context, Tsay et al., 2021) and favours the synergistic activity of the postural muscles (Danna-Dos-Santos et al., 2008).

We posited that a large amount of sensory barrage during standing on foam (Jeka et al., 2004), the unfamiliar and atypical activation of foot sole skin and intrinsic muscle receptors (Duysens et al., 2008), and the abundance in muscle activation (see Latash, 2012) would be conditions favouring sensory channel reweighting and adaptation over time of body sway with task repetition, ideally leading to enhanced body stability (Lhomond et al., 2021). Therefore, we explored the possibility that balance control adaptation would actually occur, and investigated the process of adaptation under different sensory conditions during the repetition of standing on-foam trials, as a development of a recently published paper on the specific effects of distinct sensory conditions on standing balance (Sozzi et al., 2021). Different sensory channels would be more or less sensitive to standing on foam, and the complex task of maintaining the equilibrium would be assisted by appropriately modulating the weight of the information from these channels (Mahboobin et al., 2009).

Adaptation has received much attention when repeated perturbations of balance have been investigated. The responses to a consecutive set of perturbation trials become usually smaller than those to the initial trials. Perturbations of balance can have various configurations, like translations or tilts of the base of support (Gollhofer et al., 1989; Corna et al., 1999; Schmid et al., 2011; Welch and Ting, 2014; Sozzi et al., 2020) or sudden changes in the velocity of the treadmill upon which subjects walk (Patel and Bhatt, 2015; McCrum et al., 2017) or else strikes of a loaded pendulum hitting the upper body of a standing person (Kaewmanee et al., 2020). Also, translational visual stimuli challenge the sensory reweighting mechanisms (Isableu et al., 2011; Fransson et al., 2019), much as occurs for vestibular stimulation (St George and Fitzpatrick, 2011; Barmack and Pettorossi, 2021) or muscle vibration (Bove et al., 2006; Duysens et al., 2008).

Under all settings, modulation of balancing behaviour appears in various forms and time scales. When balancing on a platform translating in AP direction, virtually no adaptation with vision takes place, and the steady-state of body segment motion and muscle activity is reached immediately after the startle reaction (Oude Nijhuis et al., 2010). This is explained by the strong stabilising effect of vision during responses to support surface translation (Akçay et al., 2021). Conversely, occluding vision produces a prolonged adaptation process, whereby the values of mechanical and EMG variables decay over time and reflex responses to muscle stretch diminish at a higher rate than the anticipatory activities (Sozzi et al., 2016). The weight of vision increases under challenging conditions (Loughlin et al., 1996; Patel et al., 2011; Lee et al., 2018) and when proprioception is disturbed (Isableu et al., 2011; Goodworth et al., 2014).

During postural perturbations and during walking, body stability is enhanced by light touch to a fixed frame (Dickstein and Laufer, 2004; Forero and Misiaszek, 2013; Martinelli et al., 2015; Kaulmann et al., 2020). Light touch rapidly reaches the somatosensory cortex (Schieppati and Ducati, 1984) and deactivation of the sensory cortex is related to worse balance performance in old age (Noohi et al., 2019). Haptic information, be it by tactile passive sensory inputs or light fingertip touch, has an effect comparable to vision (Holden et al., 1994; Rogers et al., 2001; Honeine and Schieppati, 2014; Chen et al., 2018) or to the vestibular system (Creath et al., 2008), even when standing on foam (Krishnan and Aruin, 2011; Albertsen et al., 2012). The continuous haptic input from the fingertip lightly touching an earth-fixed surface improves the control of upright stance under both stable or unstable stance (Clapp and Wing, 1999), after a perturbation (Johannsen et al., 2007) and during walking (Dickstein and Laufer, 2004; Forero and Misiaszek, 2013; Martinelli et al., 2015; Kaulmann et al., 2020) and in patients with poor balance (see Horak, 2009; Baldan et al., 2014). Hence, vision, proprioception, and haptic feedback have substantial effects in helping the body cope with perturbations of upright stance.

In the present study, we have addressed in particular the role of vision and haptic sense in the process of adaptation of stance on a compliant surface, in the absence of any external perturbation (Singh et al., 2012; Sozzi et al., 2012). We have hypothesised that adaptation would occur during repeated and prolonged standing-on-foam trials, that stance control would progressively focus on and rely on the sensory input available, and that vision and touch would differently affect adaptation. We asked the following questions. Does adaptation depend on the type (or number) of the available sensory information? Does adaptation modify the quality of stance by attaining a "default" oscillation mode, equal for distinct sensory condition(s)? Would the adapted state witness the nature of the postural control mode across specific sensory conditions? Does adaptation under light-touch conditions depend on progressive modifications of the level of force exerted by the finger? Do excursions in the frontal and sagittal plane undergo similar or different changes with adaptation, and if so is this contingent on the sensory condition?

Instead of focussing the analysis on the conventional metrics of sway, such as area and length of excursions of the centre of feet pressure (CoP), we asked whether particular frequencies of body oscillations would show specific changes during and at the end of the adaptation period, depending on the sensory information available. Hence, we leveraged the frequency analysis of the CoP excursions, based on the recent demonstration of the differences in the pattern of body stabilisation in the presence of vision compared to haptic sense (Sozzi et al., 2021). Here, we hypothesised that these inputs, yielding information of a different nature about the environment and probably processed through different brain pathways, also affect in a partially different way the process of adaptation.

MATERIALS AND METHODS

Participants

In this study, 22 healthy young adults (10 men and 12 women) participated. Age was 28.6 ± 4.4 years (mean ± *SD*), height 172.1 ± 7.1 cm and weight 67.5 ± 13.2 kg. The participants were resident physicians or physiotherapists at the Istituti Clinici Scientifici Maugeri SB. All were in good conditions, had no sight problems, or their visual acuity was corrected, were free of neurological and musculo-skeletal disorders, and none had had vertigo episodes in the past. No participant reported injuries or occurrences of falls in the previous year. All provided written, informed consent to participate in the experiment as conformed to the Declaration of Helsinki. The local review board approved the research protocol (Istituti Clinici Scientifici Maugeri SB, approval number #2564-CE). The size of the target population was based on previous experience and on the convenience of collecting the data. Since prior information is lacking, the power of the applied statistical test was assessed once the sample was collected.

Procedures

Subjects stood barefoot, for at least 100 s on a force platform (Kistler 9286BA, Switzerland) on which a foam pad (Airex Balance Pad, Sins, Switzerland, L 50 cm × W 41 cm × H 6 cm, density 55 g/dm^3, Young's module 260 kP) was placed (Lin et al., 2015). The outer profiles of the parallel feet were set at hip-width. Balance was measured under four standing conditions: with eyes open (EO), with eyes closed in the absence of touch (EC), and with light-touch without (EC-LT) or with vision (EO-LT). The position of the feet was marked on a paper sheet placed on top of the foam pad for consistency across consecutive trials. Subjects were asked to stand at ease (Hufschmidt et al., 1980) and to look at the visual scene of the laboratory wall at a 6 m distance (Sozzi et al., 2021). All subjects were naive to foam-standing. They were asked to avoid voluntary head movements in pitch, roll, and yaw, and to minimally move the eyes during the EO trials. In the EC conditions (EC and EC-LT), subjects closed their eyes before the start of the acquisition and kept their eyes closed until the end of the acquisition epoch. In the touch conditions (EC-LT and EO-LT), the instruction was to maintain a constant light touch with the index finger on the surface of the haptic device. They spontaneously chose their hand for the finger-touch task. At the end of the first trial, we asked whether that hand was the hand they used most frequently for the daily activities. The answer was affirmative in all cases and the hand was the right one. The haptic device consisted of a flat horizontal wooden square (10 cm × 10 cm) fixed on top of a strain gauge, located at about the height of the belly button and distant about 15 cm from it in the sagittal plane. When the force applied to the touchpad exceeded 1 N, the device beeped and subjects adjusted the force applied with the fingertip. This occurred rarely, mostly in the initial time period before the start of the acquisition. There was no instruction to keep the finger immobile on the force pad, so that small fluctuations in the hand and finger position were allowed. The finger never slipped off the force pad. During the no-touch conditions, both with eyes closed and eyes open, participants kept their arms relaxed by their sides. Intervals between trials were 15–30 s long when the subject stepped off the force platform and made a few steps. Before starting each consecutive trial, the experimenter verified the foot's position on the foam pad.

Each volunteer came to the laboratory four times, on separate days. Each day subjects completed eight equal-duration (∼100 s) consecutive standing trials in one of the sensory conditions of interest (EC, EO, EC-LT, and EO-LT). The order of the sensory conditions was randomised across subjects. There was no preliminary practice trial. The last 90 s-epoch of each 100 s-stance trial was acquired, to exclude the adjusting phase on stepping onto the foam pad. None of the subjects lost balance while standing on foam despite the increase in sway compared to a solid base of support (Sozzi et al., 2021). When asked at the end of the trial sequence, subjects reported no fatigue.

Data Acquisition and Processing

Details about data acquisition, processing, and identification of the frequency windows of interest in the spectrum are reported in Sozzi et al. (2021). Briefly, the platform data, from which the CoP was calculated, and the force data from the haptic device onto which the fingertip was resting, were acquired at the sampling frequency of 140 Hz by dedicated software

(Smart-D, BTS, Italy). *Post hoc* analysis was done using Excel (Microsoft), MATLAB (Mathworks), and LabVIEW (National Instrument). The force platform signals of the CoP excursions along with the anteroposterior (AP) and mediolateral (ML) directions were high-pass filtered at 0.01 Hz and low-pass filtered at 20 Hz with a 4th order Butterworth filter, after removing the respective mean values. The length of the sway path was the total length of the wandering CoP, and the sway area was the surface of the 95% ellipse fitted to the dispersion of the time-series of the AP plotted against the corresponding ML data (Schubert and Kirchner, 2014).

The frequency analysis was performed by the fast Fourier transform of the CoP ML and AP time-series of each trial, subject, and sensory condition, employing the Auto power spectrum VI algorithms of the LabVIEW functions. The duration of the acquired epoch and the sampling rate defined the lowest and the highest detectable frequency, respectively (Michalak and Jakowski, 2002; Sozzi et al., 2021). The resolution (sample frequency/sample number) was 0.011 Hz for a sampling frequency of 140 Hz and a sample number of 12,600 (equal to 90*140). The power spectrum signal was expressed in cm^2_{rms}. For each subject, trial, and sensory condition, we calculated the median frequency (that divides the power spectrum into two parts of equal area) and the mean level of the spectrum (the arithmetic mean of the amplitude values at each sampled frequency), which represents an index of the oscillation amplitude (Krizková et al., 1993; Nagy et al., 2004). These variables were calculated between 0.01 and 2 Hz and for both AP and ML directions. Distinct frequency windows (W) were identified to investigate in some detail the effect of adaptation on the frequency content of the spectrum of the ML and AP excursions of the CoP. The segmentation was based on the power spectrum profile of the EC condition, which featured the greatest overall amplitude compared to all other conditions. The procedure was based on the study of Sozzi et al. (2021) and consisted in locating the boundaries of the windows by detecting the local minima of the profile of the mean power spectrum and its standard deviation in successive epochs of 0.05 Hz. Six minima were thus defined (Sozzi et al., 2021). The same minima were used for the window definition in both ML and AP directions. Hence, ranges were: W1, 0.01–0.055 Hz; W2, 0.056–0.2 Hz; W3, 0.21–0.43 Hz; W4, 0.44–0.8 Hz; W5, 0.81–1.31 Hz; and W6, 1.32–2 Hz.

Data Treatment and Statistics

Centre of feet pressure (CoP) path length and sway area, median spectrum frequency, the mean level of the full spectrum (from 0.01 to 2 Hz) and of the distinct frequency windows were used to assess the time course of the changes in these variables over the eight consecutive repetitions. We postulated that an exponential model would fit the trends, as described on several other occasions (Tjernström et al., 2010; Schmid et al., 2011; Welch and Ting, 2014; Patel and Bhatt, 2015; Sozzi et al., 2020). To compare these trends across conditions, all data were log-transformed, and regression lines fitted to these data, even if for certain participants, sensory conditions, metrics, and frequency windows an exponential trend was not obvious. The slope of the regression lines was considered an index of the adaptation rate. The slopes of the lines obtained for each subject were averaged in each sensory condition and compared with zero (no adaptation) by the Student's *t*-test. In addition, separately for each sensory condition, the mean level of the spectrum in each window for the trials from 2 to 8 was expressed as a percent of the mean level of the spectrum of the first trial and graphed in a radar plot.

Assumptions for parametric statistics, as assessed by the Kolmogorov-Smirnov test and Levene's test, were met for all variables of interest. The effects of the different sensory conditions on path length and sway area of the CoP were compared by 4 (sensory conditions) × 8 (trial repetition) repeated measure (rm) ANOVA. The slopes of the regression lines fitted on path length and sway area were compared by a 1-way rm ANOVA. A 4 (sensory conditions) × 8 (trial repetition) rm ANOVA was used to compare the median frequency and the mean level of the spectrum between 0.01 and 2 Hz, separately for the ML and AP directions. The slope of the regression lines fitted to the median frequency and the mean level of the spectrum were compared by a 2 (ML and AP directions) × 4 (sensory conditions) rm ANOVA. A 2 (ML and AP directions) × 4 (sensory conditions) × 6 (frequency windows) rm ANOVA was used to compare the slope of the regression lines fitted to the mean level of the spectrum in each frequency window. A 2 (EC-LT and EO-LT conditions) × 8 (trial repetition) rm ANOVA was used to compare the force exerted by the subjects on the touchpad between EC-LT and EO-LT conditions. Where the differences were significant, the η^2_p was reported. The minimum effect size given our sample size (n = 22) was calculated using G*Power (Faul et al., 2007). Given our sample, the study proved to have a sufficient power (>80%) to detect an effect size d of 0.55. The *post hoc* test was Fisher's LSD test. The significance level was set at 0.05. Statistical tests were performed using Statistica (Statsoft, United States).

RESULTS

Figure 1 shows examples of the recorded time series (a segment of 60 s is depicted) in one representative subject for two different sensory conditions, namely EC and EO-LT. **Figures 1A–L** show the CoP traces recorded from the force platform in the ML (left column) and AP (middle column) directions. The recordings of the first and the last (adapted) trial are shown for each sensory condition. In the rightmost panels, the excursion of the CoP in the horizontal plane is reported. The sketch in the bottom right shows the experimental condition. CoP excursions in both ML and AP directions and sway areas were much reduced with vision and touch compared to the EC condition. The two bottom rows show the median frequency values and the mean level of the spectrum for each of all eight trials of the same subject and conditions above. The median frequencies diminished with trial repetition. The level of the spectrum was roughly constant in the ML and increased in the AP direction with trial repetition.

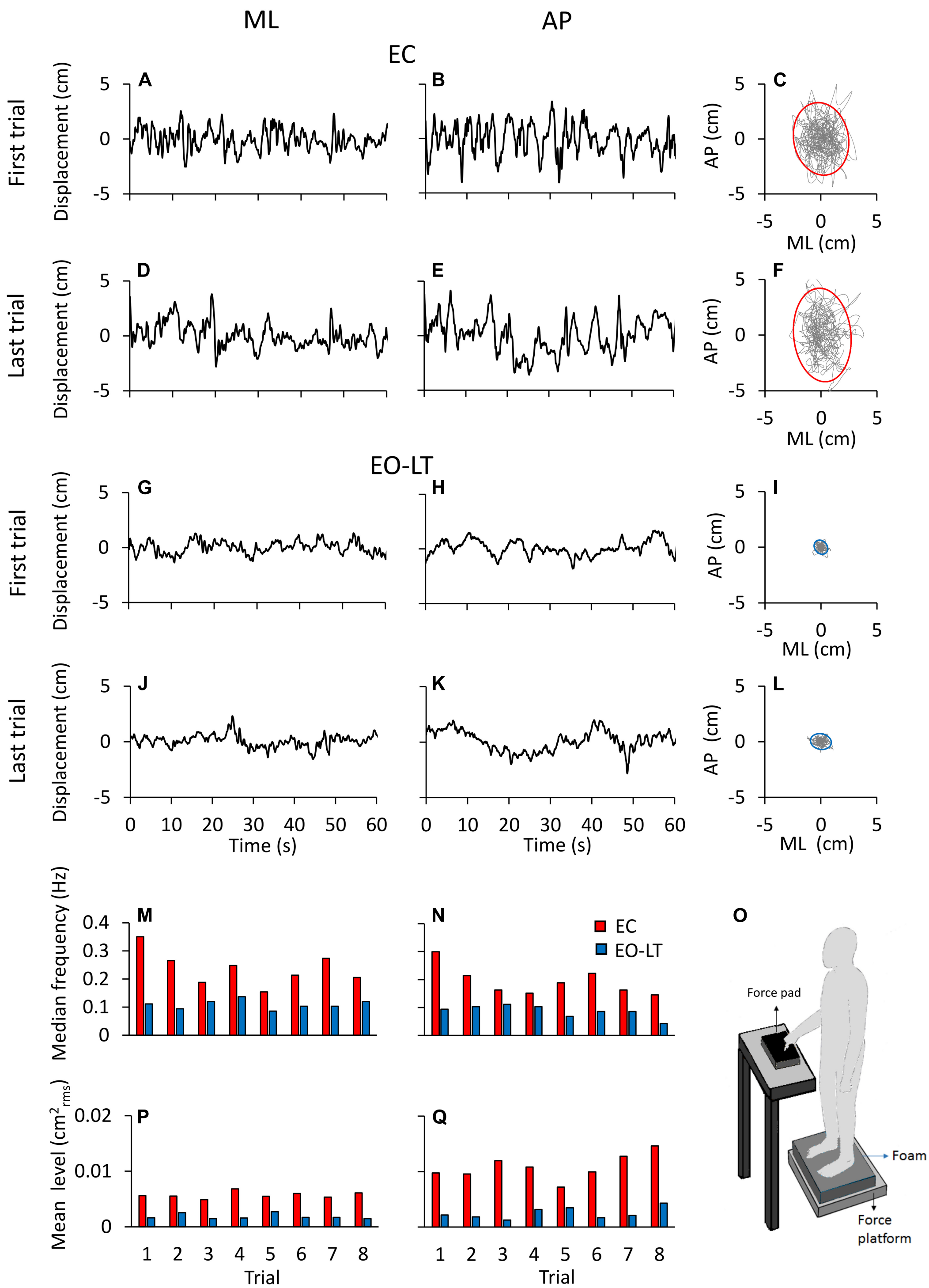

FIGURE 1 | Centre of feet pressure (CoP) excursions and effects of trial repetition on the power spectrum in a representative subject. The recorded CoP displacement in the ML **(A,D,G,J)** and AP **(B,E,H,K)** directions are reported for the first and last trials in eyes closed in the absence of touch (EC) **(A–E)** and eyes open with light-touch (EO-LT) **(G–K)** conditions. The diagrams of the CoP sway trajectory (grey line) and the 95% ellipse profile (red for EC and blue for EO-LT) are reported in **(C,F,I,L)**. **(M–Q)** Show the median frequency and the mean level of the spectrum computed for the 8 trials performed in the EC and EO-LT conditions. In **(O)**, a sketch of the experimental condition is shown.

Effects of the Trial Repetition on Path Length and Sway Area of the Excursions of the Centre of Feet Pressure

The mean path length of the CoP trace (**Figure 2A**) during the first trial was clearly longer in EC condition (red), and declined in the order EC > EC-LT > EO > EO-LT. The path lengths in EC and EC-LT (yellow) conditions diminished with trial repetition. Path lengths in EO (green) and EO-LT (blue) conditions faced no progressive decrease over the trials. ANOVA main effect showed a difference across sensory conditions [$F_{(3,63)} = 151.6$, $p < 0.0001$, $\eta^2_p = 0.88$]. The *post hoc* test found a difference for all paired comparison between sensory conditions ($p < 0.001$). ANOVA showed also a main effect of trial repetition [$F_{(7,147)} = 12.1$, $p < 0.0001$, $\eta^2_p = 0.37$] and an interaction between sensory conditions and trial repetition [$F_{(21,441)} = 7.4$, $p < 0.0001$, $\eta^2_p = 0.26$]. At the end of trial repetition, CoP path length was still the greatest in EC condition (*post hoc*, $p < 0.001$, for all comparisons) and the smallest in EO-LT ($p < 0.001$, for all comparisons). With touch (EC-LT), the path length decreased and became similar to that in the EO condition ($p = 0.32$).

There was an unspecific correspondence between path length and sway area (both were large with EC and small with EO-LT). However, the sway area (**Figure 2B**) did not follow the same adaptation pattern of path length, because the sway area increased with trial repetition. There was a difference across sensory conditions [$F_{(3,63)} = 98.8$, $p < 0.001$, $\eta^2_p = 0.82$]. The *post hoc* test found a difference between sensory conditions as well ($p < 0.05$, for all comparisons). ANOVA showed a difference among trial repetitions [$F_{(7,147)} = 12.1$, $p < 0.01$, $\eta^2_p = 0.14$]. The interaction between conditions and trial repetition [$F_{(21,441)} = 0.5$, $p = 0.95$] was not significant. **Supplementary Tables 1**, **2** show the probability of all the *post hoc* paired comparisons across repetitions for both path length and sway area, respectively. The path length of the first trial was different from that of all the others in EC and EC-LT conditions, and there was a progressive reduction in the mean path length (to about 75 and 80% of their initial value, respectively) that almost reached a floor at about the 4th–5th trial ($p < 0.001$ for all comparisons). In EO and EO-LT, there was no difference between the first trial and that of all the other trials ($p > 0.19$, for all comparisons). Sway area of the first and last trials were not different in EC ($p = 0.92$) and EO ($p = 0.11$) conditions, while a difference between these trials was observed in EC-LT ($p < 0.05$) and EO-LT ($p < 0.01$) conditions. At the end of trial repetition, the sway area was the greatest under EC condition (*post hoc*, $p < 0.001$, for all comparisons) and the smallest under EO-LT ($p < 0.001$, for all comparisons). There was no difference in sway area between EC-LT and EO conditions ($p = 0.58$).

The bottom panels of **Figure 2** summarise the effects of the repetitions on CoP path length (**Figure 2C**) and sway area (**Figure 2D**) across sensory conditions, as assessed by the adaptation rate. For each subject, a regression line was drawn through the log-transformed values of the successive trials under each of the four sensory conditions. Then, the slopes of the regression lines were averaged. ANOVA showed a difference in the mean slope of path length between sensory conditions [$F_{(3,63)} = 11.6$, $p < 0.001$, $\eta^2_p = 0.36$]. These slopes were different without vision (EC and EC-LT) from those with vision (EO and EO-LT) (*post hoc*, $p < 0.01$, for all comparisons). Conversely, there was no difference in slope between EC and EC-LT ($p = 0.99$) and between EO and EO-LT ($p = 0.1$). **Figure 2C** shows that the slope of the lines for EC and EC-LT conditions pointed downward (triangles indicate a significant difference from zero, $p < 0.001$). Hence, path length progressively decreased in these conditions, while a flat line fitted EO and EO-LT conditions, implying no adaptation. On the contrary, sway area (**Figure 2D**) showed no significant progressive reduction for the EC condition, and a moderate but definitely increasing trend (positive slope) for EC-LT, EO, and EO-LT conditions. The slope of sway area was different between sensory conditions [$F_{(3,63)} = 6.7$, $p < 0.001$, $\eta^2_p = 0.24$]. With EC, the slope was the smallest compared to the other sensory conditions (*post hoc*, $p < 0.05$ for all comparisons). There was a slight difference in slope between EO and EO-LT ($p < 0.05$).

Effect of Trial Repetition on the Mean Level of the Power Spectrum of the Centre of Feet Pressure Excursions in the Distinct Sensory Conditions

Figure 3 shows the difference in the spectrum amplitudes of the last and the first trial for each of the four sensory conditions and the difference between them. The upper traces of each of the four conditions (**Figures 3A–D,I–L**) show the average profiles of the power spectra of all the subjects. For each panel, the "adapted" profile of the power spectrum (the last trial in the series of eight, red trace) is superimposed to that of the first trial (black trace). All traces pertinent to the ML and AP directions are presented next to each other. The lower traces of each panel are the difference between the spectrum profiles (trial 8 minus trial 1; when the trace moves to the negative part of the graph (light blue area), the amplitude of the last is smaller than that of the first trial, and vice versa for the pink area). Note that in **Figure 3**, the abscissa has been limited to 1 Hz and the ordinate to 0.05 cm^2_{rms} for a better definition of the result in the lower frequency band.

Overall, the general pattern in the sagittal and frontal planes was roughly similar. Adaptation consistently increased the spectra in the last (red trace) compared to the first trial (black trace) in the low and very-low-frequency range (<0.2–0.3 Hz, pink areas in **Figures 3E–H,M–P**) in all conditions. These differences were larger without (EC and EC-LT) than with vision (EO and EO-LT) and larger in the AP than ML direction. Conversely, the spectrum mean levels diminished with adaptation in the higher frequency range (from 0.2–0.3 to 1 Hz, light blue areas) particularly with EC, whereas with vision (EO and EO-LT) all the spectra were small and the higher frequencies less represented. With the addition of both vision and touch (EO-LT), subjects felt quite stable from the beginning, as if the stabilising effects of vision and touch added up in assisting the control of stance. Nonetheless, the comparison between the mean profiles of the last and of the first trial featured an isolated, large increase in the lowermost frequencies (from about 0.01 to about

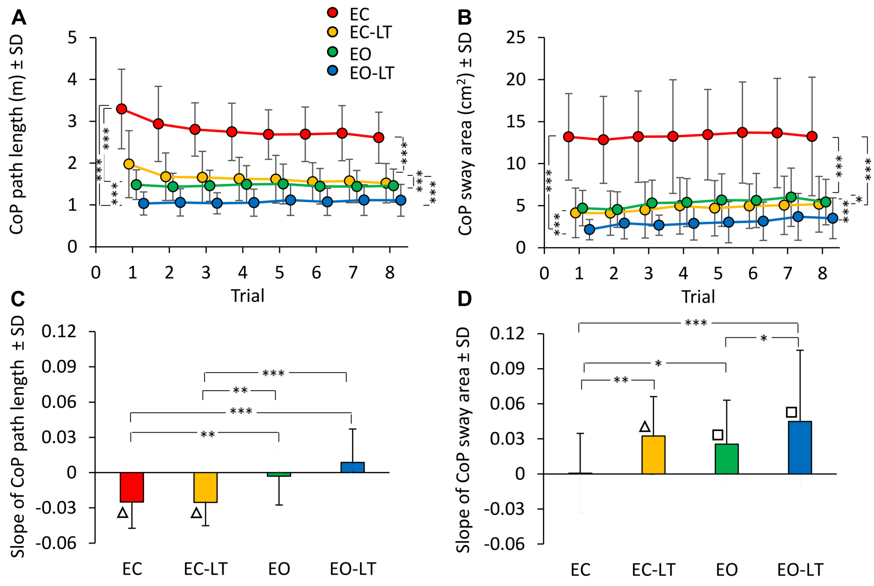

FIGURE 2 | Effect of trial repetition on path length and sway area of CoP excursions under different sensory conditions. Path length **(A)** diminished with trial repetition with EC (red) and EC with light-touch (EC-LT) (yellow), while no adaptation was evident with vision (eyes open (EO), green, and EO-LT, blue). Sway area **(B)** moderately increased with trial repetition for all conditions (less so for EC). **(C,D)** Show the mean slope of the lines fitted to path length and sway area values across the trials. The negative slopes with EC and EC-LT indicate a decrease in path length. Slopes were not different from zero with EO and EO-LT. The slopes for the sway area were positive, indicating an increase, except for EC. Asterisks indicate significant differences between sensory conditions (*$p < 0.05$; **$p < 0.01$; ***$p < 0.001$). Symbols indicate slopes different from zero (□$p < 0.01$; Δ$p < 0.001$).

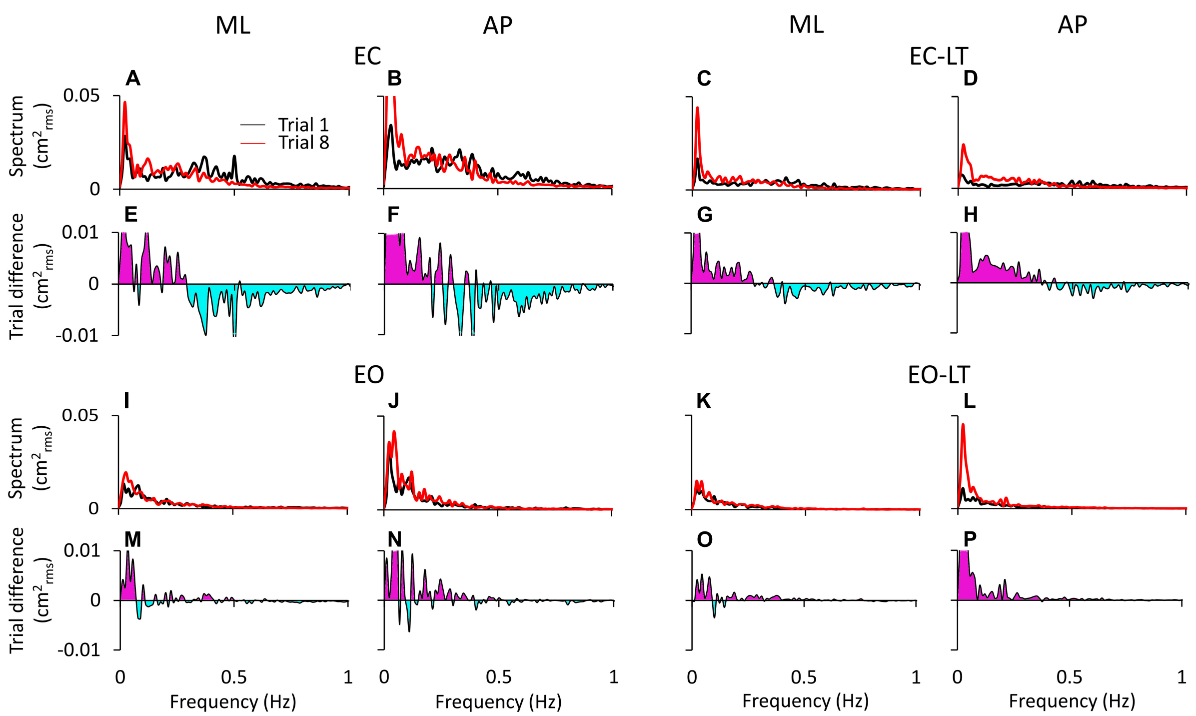

FIGURE 3 | Power spectrum differences between the first and last trials. The profiles of the mean power spectrum of the first (black traces) and last (red traces) trials for both ML and AP directions are superimposed for EC **(A,B)**, EC-LT **(C,D)**, EO **(I,J)**, and EO-LT **(K,L)**. In the panels from **(E–H,M–P)**, the differences between the spectrum of the last and the first trial (trial 8 minus trial 1) are reported. The pink area indicates a positive difference, i.e., in the corresponding frequency range, the spectrum of the last trial is greater than that of the first trial. The negative differences are highlighted in light blue, indicating that the spectrum of the last trial is smaller than that of the first trial.

0.1 Hz), larger in AP. The frequencies around 0.2–0.4 Hz, i.e., where the traces of the differences cross the abscissa, did not apparently change between the first and the last trial (A to D).

Effect of Trial Repetition on the Median Frequency of the Spectrum of the Centre of Feet Pressure Excursions in the Different Sensory Conditions

The median frequency of the entire power spectrum (0.01–2 Hz) of the CoP excursions of the first trial showed distinct values for each sensory condition. Differences were present for the ML and AP directions as well. The mean values of the median frequency diminished with adaptation in the ML and in the AP direction (**Figures 4A,B**). The adaptation pattern of the median frequency was similar for EC and EC-LT, as if the absence of vision, regardless of touch, was the main cause of the shift toward low frequencies with trial repetition. Conversely, with vision (EO and EO-LT), the median frequency of the spectra showed little changes over time and was overall smaller (approximately 0.1–0.15 Hz) than that observed for the EC and EC-LT conditions.

ANOVA showed a difference in the median frequency between sensory conditions for ML [$F_{(3,\ 63)} = 58.7$, $p < 0.0001$, $\eta^2_p = 0.74$] and AP directions [$F_{(3,\ 63)} = 61.05$, $p < 0.0001$, $\eta^2_p = 0.74$]. The median frequency was definitely higher with EC and EC-LT than with vision (EO and EO-LT), both in ML (*post hoc*, $p < 0.001$ for all comparisons) and in AP directions (*post hoc*, $p < 0.0001$ for all comparisons). ANOVA showed also a difference between trial repetitions for ML [$F_{(7,\ 147)} = 16.8$, $p < 0.0001$, $\eta^2_p = 0.44$] and AP directions [$F_{(7,\ 147)} = 22.4$, $p < 0.0001$, $\eta^2_p = 0.52$]. There was also a significant interaction between conditions and trial repetition for both ML [$F_{(21,\ 441)} = 4.75$, $p < 0.0001$, $\eta^2_p = 0.18$] and AP directions [$F_{(21,\ 441)} = 7$, $p < 0.0001$, $\eta^2_p = 0.25$]. Hence, the median frequency patently diminished with the successive trials, a sign of relative progressive prevalence of lower over higher body oscillation frequencies, particularly without vision. All the *post hoc* comparisons are reported in **Supplementary Tables 3, 4**. For both ML and AP directions, median frequency was different between the first and last trials under EC and EC-LT conditions ($p < 0.001$, for all comparisons). Under EO and EO-LT conditions, median frequency was not different between the first and the last trial ($p > 0.09$), except for AP EO-LT condition ($p < 0.05$).

The mean slopes of the regression lines fitted to the log-transformed values of the successive trials are reported in **Figure 4C** for the median frequency data of each sensory condition. ANOVA showed a difference in the slopes between sensory conditions [$F_{(3,\ 63)} = 8.4$, $p < 0.0001$, $\eta^2_p = 0.29$] and an interaction between ML and AP directions and sensory conditions [$F_{(3,\ 63)} = 2.95$, $p < 0.05$, $\eta^2_p = 0.12$]. There was no difference between ML and AP directions in the slope of

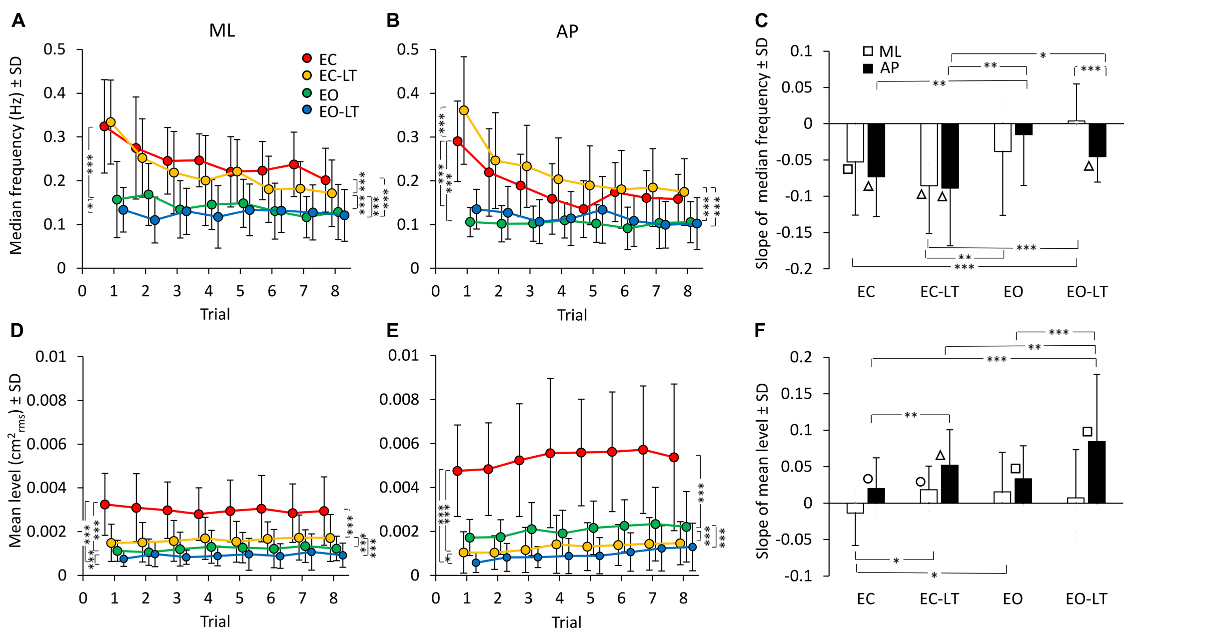

FIGURE 4 | Effect of trial repetition on the median frequency and mean level of the spectrum under the different sensory conditions. Median frequency decreased with trial repetition in the no-vision conditions for both ML **(A)** and AP **(B)** directions. Median frequency was higher in EC (red) and EC-LT (yellow) than in EO (green) and EO-LT (blue) conditions. **(C)** Shows the mean slope of the lines fitted to the median frequency data. Except for EO-LT in the ML direction, all the slopes were negative, indicating a reduction in the value of the median frequency. The mean level of the spectrum for the ML **(D)** and AP **(E)** direction is also reported. For the ML direction, the mean level of the spectrum showed no clear changes with trial repetition, while in the AP direction the mean level of the spectrum increased with trial repetition. **(F)** Shows the mean slope of the lines fitted to the mean levels of the spectrum. Except for the EC in ML direction, all the slopes were positive, indicating an increase of the mean level of the spectrum with the trial repetition. Asterisks indicate significant differences between sensory conditions (*$p < 0.05$; **$p < 0.01$; ***$p < 0.001$). Symbols indicate slopes significantly different from zero (○$p < 0.05$; □$p < 0.01$; △$p < 0.001$).

the median frequency [$F_{(1,\ 21)}$ = 1.2, p = 0.28]. For the EC and EC-LT conditions, the slopes were definitely decreasing (the white and black bars in **C**) in the successive trials, the slopes of the lines fitted to the EO and EO-LT data reached significance only for AP EO-LT. The slope of the median frequency in ML direction was different between EO-LT and the other conditions (*post hoc*, $p < 0.05$, for all comparisons). The slope of EC-LT was greater than that of the EO condition ($p < 0.01$) for both ML and AP directions. The slope was not different between EC and EO conditions in the ML direction (p = 0.4), but there was a difference between these two conditions in the AP direction ($p < 0.01$). Obviously, the adaptation trends in the median frequencies were different between no-vision and vision, while touch had a small effect on these trends. The difference between the amplitude of the profiles of the last (red) and first (black) trials reported in **Figure 3**, added to the increase in the amplitude of the low-frequencies, likely explain the relative decrease in the values of the median frequency of the last compared to the first trial.

Effect of Trial Repetition on the Mean Level of the Spectrum of the Centre of Feet Pressure Excursions in the Different Sensory Conditions

The mean level of the spectrum for the subsequent trials in each sensory condition is reported for the ML and AP directions in the lower panels of **Figures 4D,E**. While the adaptation effect on the median frequencies was plain to see, there was no progressive decrease in the mean level of the spectrum in the ML direction and a moderate, albeit definite, progressive increase in the level of AP spectrum. The mean level of the spectrum was different between sensory conditions in both ML [$F_{(3,\ 63)}$ = 65.7, $p < 0.0001$, η^2_p = 0.76] and AP directions [$F_{(3,\ 63)}$ = 95.5, $p < 0.0001$, η^2_p = 0.82]. The mean spectrum level for the "stabilised" sensory conditions (EC-LT, EO, and EO-LT) was clearly smaller than with EC (*post hoc*, $p < 0.0001$ for all comparisons). ANOVA showed no significant effect of trial repetitions for the ML direction [$F_{(7,\ 147)}$ = 0.38, p = 0.91], but an effect for the AP direction [$F_{(7,\ 147)}$ = 4.9, $p < 0.0001$, η^2_p = 0.19]. There was no significant interaction between sensory conditions and trial repetition for either ML [$F_{(21,\ 441)}$ = 1.4, p = 0.1] or AP directions [$F_{(21,\ 441)}$ = 0.66, p = 0.87]. All the *post hoc* comparisons are reported in **Supplementary Tables 5**, **6**. In general, there was no significant difference in the mean level of the spectrum along the ML direction between the first and last trials ($p > 0.12$), except for EC condition ($p < 0.05$). For the AP direction, there was a difference between the first and last trials (last > first) for all sensory conditions ($p < 0.05$, except EC-LT, p = 0.09).

The slopes of the lines fitted to the log-transformed data of the mean levels of the spectrum are reported in **Figure 4F**. Slopes were generally close to zero in ML direction, but greater than zero in AP direction, indicating a progressive increase in the spectrum level in the sagittal plane. ANOVA showed a difference between ML and AP directions [$F_{(1,\ 21)}$ = 20.3, $p < 0.001$, η^2_p = 0.49], a difference between sensory conditions [$F_{(3,\ 63)}$ = 5.2, $p < 0.01$, η^2_p = 0.2] and an interaction between ML and AP directions and sensory conditions [$F_{(3,\ 63)}$ = 4.5, $p < 0.01$, η^2_p = 0.18]. The *post hoc* analysis showed lower slopes in the ML than in AP direction ($p < 0.01$ for all comparisons) except for the EO condition (p = 0.14). For ML direction, the slope of the EC was different from that of EC-LT and EO conditions (*post hoc*, $p < 0.05$, for both comparisons), but not from EO-LT (p = 0.09). For the AP direction, the slope in the EO-LT condition was greater than that in the other sensory conditions ($p < 0.01$, for all comparisons). The slope in EC was also smaller than that in the EC-LT condition ($p < 0.01$). The differences between the amplitude of the profiles of the last (red) and first (black) trials reported in **Figure 3** explain the relative constancy of the mean levels of the power spectra compared to the large decrease in the values of median frequencies because the adaptation-induced increase in the lower frequencies is compensated by the decrease in the higher frequencies.

Distinctive Changes in the Mean Level of the Spectrum of the Frequency Windows

There were similarities and differences in adaptation across conditions and directions of CoP excursion. The radar charts of **Figure 5** show the progressive amplitude changes (successive trials are indicated in different colours) of the six frequency windows, for the four sensory conditions (from left to right) and for the ML and AP directions (top and bottom, respectively). Note that the levels of the spectrum are reported in these charts in *percent* of the level of the first trial (the inner black 100% hexagon) for each window, to render the changes in the higher frequency windows more conspicuous compared to their absolute values. This procedure allows to compare the different trials within a given sensory condition, but not between different sensory conditions or ML and AP directions. A major rise in the level of the spectra of the low-frequency windows occurred with EC, EC-LT, and EO-LT. Its percent increase compared to the first trial was prominent in W1 and W2 (from 0.01 Hz to 0.2 Hz) with EC-LT in the AP direction. Concurrently, definite reductions occurred in the high-frequency windows (W4–W6). These reductions were common to ML and AP directions. A further noticeable finding was the largely unaltered amplitude of the spectrum of W3, mostly common to all sensory conditions and directions.

The graphs in **Figure 6** add information on the mean adaptation rates of the mean levels of the spectra of each frequency window in the four sensory conditions (same colour code as in **Figures 2**, **4**) in ML (**Figure 6A**) and AP directions (**Figure 6B**). Sensory conditions [$F_{(3,\ 63)}$ = 6.51, $p < 0.001$, η^2_p = 0.24], frequency windows [$F_{(5,\ 105)}$ = 65.01, $p < 0.0001$, η^2_p = 0.76] and ML and AP directions [$F_{(1,\ 21)}$ = 20.2, $p < 0.001$, η^2_p = 0.49] affected the adaptation pattern. Some patterns were common to all conditions, some were different. There was an interaction between ML and AP directions and frequency windows [$F_{(5,\ 105)}$ = 2.44, $p < 0.05$, η^2_p = 0.1], between ML and AP directions and sensory conditions [$F_{(3,\ 63)}$ = 3.26, $p < 0.05$, η^2_p = 0.13], between frequency windows and sensory conditions [$F_{(15,\ 315)}$ = 8.17, $p < 0.0001$, η^2_p = 0.28] and between ML and AP directions, frequency windows and sensory conditions [$F_{(15,\ 315)}$ = 2.5, $p < 0.01$, η^2_p = 0.11]. Specifically, some frequency windows were unaffected by adaptation, while some were deeply

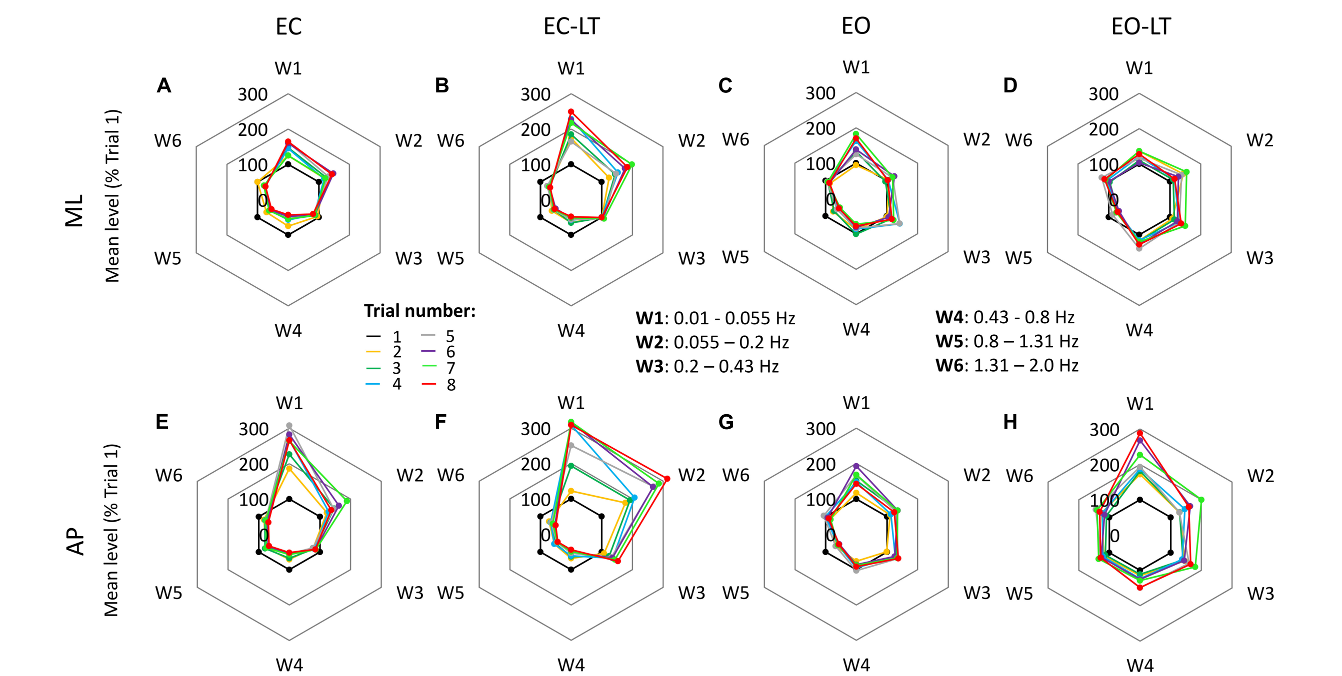

FIGURE 5 | Changes in the mean level of the power spectrum in the six frequency windows, in which the full spectrum (from 0.01 to 2.0 Hz) was separated, for all trials and sensory conditions. Data are expressed as a percent of the spectrum mean level of the first trial (black dots at 100%) in each condition and direction (this prevents a quick comparison of the graphs between conditions). In general, for both ML **(A–D)** and AP **(E–H)** directions and for all sensory conditions, the mean level of the spectrum increased over the successive trials in the first two windows (W1 and W2), showed minor changes in W3, and decreased in the windows from W4 to W6.

modified with growing or diminishing spectrum amplitude. The amplitude of the low-frequency windows (W1 and W2) had a definite increasing trend (positive slope) with trial repetition, W3 persisted almost unchanged with a slope not different from zero for all conditions in the frontal plane and minimally changed in the sagittal plane, the last three frequency windows progressively decreased (except with EO-LT).

Common patterns were the larger positive slope of the low-frequency windows in AP than in ML (*post hoc*, W1, $p < 0.05$ and W2, $p < 0.01$, except for EO-LT), and the generally negative slopes in W4–W6, similar between ML and AP ($p > 0.07$ for all comparisons except for EC-LT in W6, $p < 0.01$). Moreover, there was no statistical difference from zero in the slope of EO-LT condition in the frontal plane for all frequency windows, and a positive slope of the adaptation rate in the EO-LT in the sagittal plane for all windows up to W4. The results of the *post hoc* comparisons between sensory conditions in the distinct frequency windows are reported in **Supplementary Table 7**.

Differences Between the Mediolateral and Anteroposterior Spectra of the Adapted Trials in the Different Sensory Conditions

The power spectra of the adapted (last) trials have been compared to each other between sensory conditions. For instance, the comparison of the spectra of the EC and EC-LT conditions would disclose the unique effect of touch in the absence of vision, while the comparison between EO and EO-LT would reveal the effect of touch in the presence of vision. In **Figure 7**, the left and right columns refer to the ML and AP directions, respectively. In each panel of **Figures 7A–E**, the upper traces show the superimposed profiles of the adapted spectra. The bottom traces show the amplitude of the differences between the spectrum profiles (the minuend and the subtrahend are indicated in the legend). In general, most adapted spectra showed substantial differences depending on the selected paired conditions, the ML or AP directions, and the range of oscillation frequencies.

In **Figure 7A** (EC-LT compared to no-touch EC), touch reduced both the ML (left) and the AP (right) adapted spectra. The differences were scattered across all frequencies (mostly below 0.5 Hz) but were much larger for AP than ML (AP > ML, compare the light blue areas in the lower traces of **Figure 7A**). Vision (**Figure 7B**, EC compared to EO) led to changes similar to those produced by touch in the EC adapted trial. Adding touch to vision (**Figure 7C**, EO-LT compared to EO), in the context of an overall much smaller spectrum level, slightly reduced the very low frequencies in ML direction in the adapted trial. Again, however, the reduction in the amplitude of the power spectrum was of some importance for the AP excursions, where differences were recurrent also at higher frequencies. In other words, the reduction in the mean level of the spectrum by adaptation was more substantial in AP than ML when touch was present. The comparison of the adapted trials with touch no-vision and with

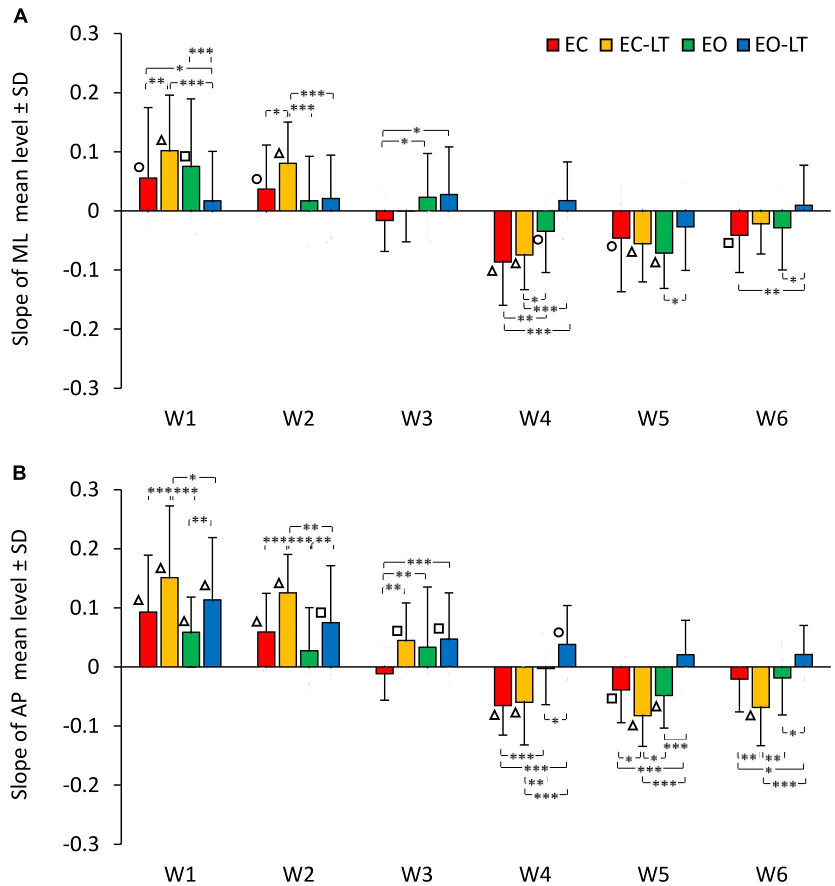

FIGURE 6 | Adaptation rate in the frequency windows (Ws) of ML **(A)** and AP **(B)** spectra. The mean slopes of the lines calculated for each frequency window (W) are reported for ML and AP in the different sensory conditions (red, EC; yellow, EC-LT; green, EO; blue, EO-LT). For all sensory conditions, the slopes of W1 and W2 are always positive indicating an increment, during the trial repetition, in the mean level of the spectrum. In W3, the slope is not different from zero in the ML direction for all sensory conditions, while in the AP direction was positive, except for the EC condition, and different from zero only for touch (EC-LT and EO-LT). From W4 to W6, except for the EO-LT condition, the slopes are always negative indicating a reduction in the mean level of the spectrum during the trial repetition. Asterisks indicate significant differences between sensory conditions (*$p < 0.05$; **$p < 0.01$; ***$p < 0.001$). Symbols indicate slopes significantly different from zero (○$p < 0.05$; □$p < 0.01$; △$p < 0.001$).

vision no-touch (EC-LT vs. EO, **Figure 7D**) yielded modest differences in the ML direction. The touch no-vision adapted trials showed a lower amplitude of the spectrum up to 0.2 Hz in AP direction compared to EO. **Figure 7E** shows the difference between the adapted trials with and without vision in presence of touch (EC-LT vs. EO-LT). With touch and vision (EO-LT) compared to touch no-vision (EC-LT), adaptation reduced the spectrum between 0.1 and 0.5 Hz in both ML and AP direction. At the lowest frequencies, the spectrum was larger in AP than ML directions.

Force of Finger Touch Across Trials

Figure 8 shows the mean level of the force applied by the subjects on the touchpad during the EC-LT and EO-LT trials. The mean force never exceeded 1 N, despite non-negligible variability across subjects. ANOVA showed no difference between conditions [EC-LT vs. EO-LT, $F_{(1,\ 20)} = 1.1$, $p = 0.31$] and no main effect of trial repetition [$F_{(7,\ 140)} = 1$, $p = 0.46$]. There was also no significant interaction between conditions and trial repetition [$F_{(7,\ 140)} = 1.3$, $p = 0.26$]. Therefore, the observed changes in the geometric and spectral measures were not related to a graded change in the fingertip force exerted during the subsequent trials.

DISCUSSION

Body Sway Adaptation to the Repeated Standing Trials

We had postulated that a process of postural adaptation would take place, also in the absence of reactions to external artificial stimulation(s), and reflect a progressive involvement of higher centres (Mierau et al., 2015; Kaulmann et al., 2020). We leveraged the use of the modulations in the frequency of the power spectrum of the CoP excursions (Schumann et al., 1995), following the approach exploited in several studies on balance adaptation to postural disturbance (Loughlin et al., 1996;

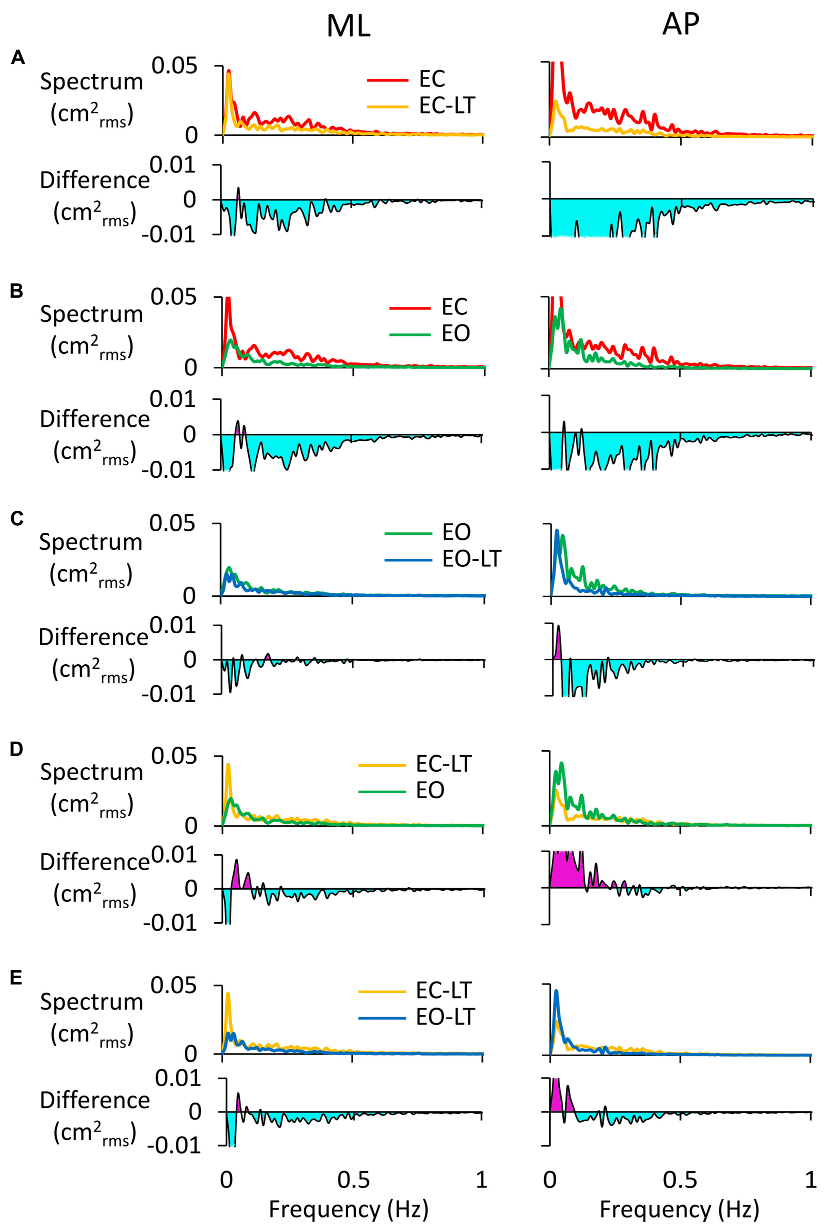

FIGURE 7 | Comparisons of adapted trials between sensory conditions (coloured). In the upper traces of each panel, the mean power spectra of ML (left column) and AP directions (right column) of the last trial are reported and compared. In the lower traces, the difference between spectra is shown: **(A)**, EC-LT minus EC; **(B)**, EO minus EC; **(C)**, EO-LT minus EO; **(D)**, EO minus EC-LT; **(E)**, EO-LT minus EC-LT. Light blue areas highlight negative differences between the pairs of sensory conditions. The pink area in **(D,E)** indicates that the spectrum of the most stabilised condition (e.g., EO in **D** and EO-LT in **E**) is greater than the contrast condition (EC-LT in **D,E**).

Kiers et al., 2015; Borel and Ribot-Ciscar, 2016; Assländer et al., 2020). In accordance with our assumptions, we have seen that adaptation occurs during repeated, prolonged standing-on-foam trials. This adaptation affects both geometric variables (path length and sway area) and spectral markers (median frequency and mean level of frequency spectrum) of the CoP excursion. Where significant differences were observed, the adaptation impact had an effect size greater than the minimum effect size detectable based on our sample of 22 participants.

We have found peculiar features of adaptation (meaning here both a decrease *and* an increase in the value of distinct variables) in the oscillations of the CoP, with features depending on availability of vision and touch. The changes in the responsiveness to the closely recurrent exposures to stance trials were contingent on the metric considered and on the direction (ML or AP) of the CoP excursion. The power spectrum of the oscillations showed distinct features in the different sensory conditions, and the mean level of distinct frequency windows

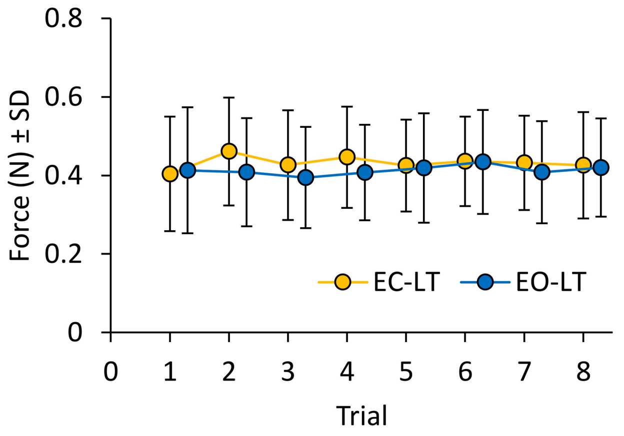

FIGURE 8 | Force applied on the touchpad during trials (means of all subjects). Force was not different between EC-LT (yellow symbols) and EO-LT (blue symbols). In both conditions, the force never exceeded1 N and remained constant across trials.

of the spectrum changed over the trials depending on the conditions. Finally, the adapted behaviour at the end of the trial sequence did not overlap for all sensory conditions, indicating that there is no unique, common "default" or "optimal" adapted behaviour, as a necessary outcome of trial repetition.

Several new findings of the present study deserve attention. (1) Adaptation is present under some but not all sensory conditions. (2) Adaptation does not eventually lead to a stable state, as when a ball in a parabolic bowl of any shape (here, the sensory conditions) is released from some initial height and allowed to roll back and forth inside the bowl. (3) Adaptation rate is not necessarily related to the amplitude of the given variable in the first trial. (4) Adaptation reduces the amplitude of some sway variables and increases that of others. (5) CoP path length generally decreases while sway area generally increases. (6) In the spectral dimension, the median frequency generally decreases, whereas the mean level of the spectrum tends to increase over time, mainly in the AP direction. (7) Importantly, adaptation differently affects distinct frequency ranges, whereby the low frequencies increase in amplitude, middle-range frequencies remain constant, high frequencies decrease. (8) The adaptation process differently affects the mean level of the spectrum and its adaptation rate in the frontal and the sagittal planes.

The subjects' report about absence of fatigue or "muscle tension" at the end of the session, and the declared negligible physical effort for standing, rule out the possibility that fatigue plays a role in the adaptation by increasing the CoP oscillations (Nardone et al., 1997; Bisson et al., 2012; Bermejo et al., 2015). Incidentally, an increase in the high-frequency band was observed with leg muscle fatigue (Bizid et al., 2009), but is not seen in the present study. Moreover, the adaptation process was not accompanied by matching changes in haptic input over time, since the force applied to the touchpad was constant across the repeated trials, regardless of vision availability. In a sense, the absence of the stabilising visual input was not compensated by a stronger haptic inflow, in keeping with the conclusions by Garbus et al. (2019).

The Adaptation Process Occurs in the Absence of Repeated Balance Perturbations

It might appear extravagant to address the adaptation process of standing upright without administering external or artificial perturbations. Rarely if ever it happens that people stand in place for minutes, except perhaps when standing at attention if you are a soldier (normally on a non-compliant base of support). Anyhow, for that matter, repeated perturbations hardly ever come about in real life. However, both approaches to this issue are valuable, because they deal with the general theme of changes in motor behaviour taking place with task repetition. In accordance with our expectations, an adaptation process occurs during repeated, prolonged standing-on-foam trials. In particular, standing on foam induces self-imposed (or internally generated) perturbations of the stance that need to be counteracted by reactions to the elicited stimuli. Both foam and perturbations imply balance threat, reflex responses, descending modulation, and both ultimately call for mastering balance (Tjernström et al., 2002, 2005; Clair et al., 2009). But standing on foam does not imply any degree of preparatory or expectancy activity that may interfere with responses to perturbations.

Just to mention distant examples of "adapting" behaviours, impressive long-term plasticity helps control balance when the underpinning neural processes slowly deteriorate over time (see Tighilet and Chabbert, 2019; Barmack and Pettorossi, 2021). On a very short time scale, post-effects of a particular sensory condition are observed on shifting to a different condition (Hay et al., 1996) or are produced by sensory reweighting after attenuation of one modality (Billot et al., 2013; Honeine and Schieppati, 2014), or simply modify postural adjustments after a single training session of a catching task (Kanekar and Aruin, 2015). What triggers adaptation, whether it is dependent on predictions of balance threats or on the capacity of reweighting the sensory information, or else on explicit learning is a matter of controversy (Peterka and Loughlin, 2004; Assländer et al., 2020; Bakker et al., 2021; see Rothwell et al., 2021, for an excellent short account of neural adaptation recourses). The present investigation claims to be a preliminary methodological approach to such a complex question.

A decrease in body sway occurs with trial repetition when standing on a solid base of support in the absence of vision, but not when vision is available (Tarantola et al., 1997; Reed et al., 2020). Drawing on that former finding, we have investigated here the balance adaptation process of young healthy subjects during standing on foam, having in mind that presence or absence of touch and vision can entail different "attractors" (Lee et al., 2018) and disclose uncharted modulations of central integration processes (Sozzi et al., 2021). The foam was selected because it is more demanding compared to standing on a solid base of support (Allum et al., 2002). Foam challenges control because the reaction of the compliant surface to muscle action is unfamiliar and unpractised and enhances the level of attention and voluntary corrections (Di Fabio et al., 1990).

Eyes Closed, Without, and With Touch

At variance with quiet stance on a solid surface, where geometric markers of body oscillation are almost superimposable without (EC) and with vision (EO) (Sozzi et al., 2021), sway on foam is much larger EC than EO. Adaptation is clearly observable with EC, where the body oscillations, measured by CoP path length, are the largest and progressively diminish (Jeka and Lackner, 1995), but remain larger at the end of the adaptation period, compared to the other sensory conditions. This suggests that the absence of visual information can be compensated—not completely—by the progressive up-weighting of the remaining senses (proprioception, vestibular, plantar cutaneous) (Bernard-Demanze et al., 2004; Šarabon et al., 2013a; Goodworth et al., 2014). As observed under different conditions, i.e., perturbation of stance by a mobile platform (Sozzi et al., 2016), the adaptation is obvious with EC (again not with EO) and is not immediate.

With EC, adaptation entrains a pronounced reduction of the median frequency of the spectrum (for both ML and AP directions), while the mean level of the spectrum remains constant in the ML direction and slightly increases in the AP direction. In other words, the excursions of the CoP become slower, but the size of the surface covered by the wandering of the CoP remains large across the repetitions. It seems that accuracy in maintaining the CoP within a limited space of the support base is not required when standing on foam. It might be useless to strictly control the CoP excursion when no steps or gait initiation are planned, or in the absence of an explicit instruction to stand as still as possible (Bonnet, 2016), or else because muscle co-contraction in addition to being costly (Houdijk et al., 2015) would be unsafe. Rather, exploiting the redundancy of the body segments' degrees of freedom can enhance the chance of finding a safe posture (Rasouli et al., 2017) or several safe postures over time despite the absence of vision.

These features are broadly common to EC-LT when touch is added in the absence of vision. Sway is reduced with touch (EC-LT) to values a little larger than with vision. However, the rate of decrease in path length is analogous to that in EC condition, even though path length is initially only two-thirds of that with EC. This relative difference remains in the final trials as well. Much as for EC, the sway area with EC-LT slightly increases in size and the median frequency decreases with adaptation. Hence, touch eliminates a certain constant amount of oscillation amplitudes and (high) frequencies, while the adaptation rate itself is set by the absence of vision. The mechanisms through which touch reduces sway are not flagrantly time-dependent. Kaulmann et al. (2020) also observed that light touch did not reduce the time constant of postural compensation following a perturbation. They suggested that owing to the overall smaller sway with than without touch, an invariant time constant would nonetheless allow reaching a steady-state earlier with than without touch.

Eyes Open, Without, and With Touch

With vision (EO), the progressive decrease in path length is absent. In the beginning, the path length is much smaller than with EC (<50%) and remains constant. With touch added to vision (EO-LT), the path length is the shortest and remains constant over time as well. Sway area is also small, but slightly and steadily increases. The mild rise is accompanied by an increase in the mean level of the spectrum in the very-low-frequency range (<0.1 Hz) and by a minor reduction in the value of the median frequency. It has been shown already that vision compared to no-vision reduces the sway frequencies below 1 Hz (Dichgans et al., 1976; Mezzarane and Kohn, 2007; Sozzi et al., 2021). However, the lowest frequencies (W1 and W2) do not further decline with the adaptation, rather they relatively build up over the successive trials instead (more in AP than ML). Perhaps, with touch and vision (EO-LT), subjects feel safe and free to sway thanks to the haptic input, meaning that they can change freely the activation turnover across many relevant postural muscles. While this seems to indicate evidence of a reweighting of the haptic input itself, the concurrent slow increase in the sway area (EO-LT) might indicate a shift to progressive better exploitation of the exploratory behaviour (Carpenter et al., 2010; Fabre et al., 2020). This could also contribute to preventing receptor and muscle fatigue.

The Contribution of Vision and Touch to the Adaptation

Initial values of body sway with vision or touch are comparable to those found previously (Rogers et al., 2001; Wing et al., 2011; Sozzi et al., 2021). We confirm here that both vision and touch exert a powerful stabilising effect even when standing on foam and that adaptation is present under some but not all sensory conditions. The adaptation is prominent without vision in the geometrical markers and in the median frequency. Despite the differences in the amplitude of the spectra and in the median frequency, the adaptation rates are similar in the EC and EC-LT. Adaptation rates are also similar between EO and EO-LT, but their value is much slower without vision, possibly because of a floor effect given by the low mean level of CoP excursion and low median frequencies of the spectra. During the adaptation period, the spectrum shifts toward the low frequencies (W1 and W2, in ML and AP directions). Touch, much as vision, progressively reduces the amplitude of the spectrum in the high-frequency range (W4–W6) in both ML and AP directions (Riley et al., 1997; Rogers et al., 2001). Whereas, touch speeds up the rise of the low frequencies compared to the same visual condition without touch. Touch reduces by a constant amount path length, sway area and mean level of the spectrum compared to no-touch, regardless of visual condition (EC-LT and EO-LT). This "offset" is large with EC and minimal with EO, but in both cases, it does not vary with adaptation. It would be safe to conclude that the process of postural adaptation is set when vision is not available and that, with vision, adaptation is almost ineffectual. Oddly enough, with touch and vision, all geometrical and spectral markers are manifestly diminished compared to EC, but during the adaptation process, a slight but significant increment in these markers is observed as if the stabilising effect of vision and touch would be down-weighted. The rate of this increment is slightly greater with than without touch. Perhaps, in the adaptation period, a trade-off is reached whereby the postural control system accepts a minor stabilisation (in the sagittal but not in the frontal plane) in exchange for a more relaxed attitude and less attention spent in the control of the finger force. Likely, touch acted mainly as a task demand, i.e., vision set by trial and error the stabilisation process while touch was a “supra-postural task” exploiting but not contributing to the stabilisation process (Lee et al., 2018).

Non-identical Adaptation Along the Frontal and Sagittal Planes

The control of balance in the frontal and sagittal plane is peculiar (Day et al., 1993; Winter et al., 1996) and conditional on the interaction of two independent postural sub-systems, the synergic action of which complies with the demands of a precision task (here represented by the maintenance of the equilibrium, with or without the light touch) (see Balasubramaniam et al., 2000). The excursions of the CoP in the ML and AP directions are produced by coordinated activation of different muscles (about the hip and the ankles, respectively) (Winter et al., 1996; Zhang et al., 2007). Singh et al. (2012) showed that visual information focuses the control of sway in AP direction at almost all frequencies, whereas a foam surface rather affects sway in ML direction at the middle (from about 0.6 to 0.7 Hz) and high (from about 2 to 7 Hz) frequencies. Importantly, sway amplitude in the frontal plane is proportional to the threshold of vestibular encoding of lateral body translation (Karmali et al., 2021), and ageing affects both the vestibular and the balance systems (see Wagner et al., 2021). When standing on foam, the head is stabilised better in the frontal than the sagittal plane in healthy subjects, showing accurate control of hip motion in roll (Fino et al., 2020). Hence, any effect of the adaptation on the ML or AP excursions seems to be of interest in the light of potential interventions.

Adaptation manifests itself along both the AP and ML directions in all sensory conditions. This is shown by a decrease in the median frequency (EC and EC-LT) and by the increase in the mean spectrum level, especially in AP. This increase is caused by the shift of the oscillation frequencies toward very low values (<0.2 Hz, W1 and W2) and a decrease in the medium-high frequencies (0.3–2 Hz, corresponding to about W4–W6). The increase in the frequencies lower than 0.1 Hz confirms the conclusions of Rasouli et al. (2017) and Yamagata et al. (2019) about the presence of slow “drifts.” We do not know why the slow drifts build up with adaptation to a greater degree in the sagittal plane, but this is probably related to the foot-ankle architecture, whereby larger and slower torques are allowed in the sagittal than the frontal plane (Murray and Sepic, 1968; Pai and Patton, 1997). This favours a “safer” fore-aft balancing strategy by the slow ankle or hip rotations in the sagittal plane (Ogaya et al., 2016), likely assisted by the elastic properties of the foam. With touch, the spectrum amplitude remarkably diminishes, but part of the “drift” effect in the AP direction that builds up with adaptation can be connected to changes in the tone of the axial muscles (Johannsen et al., 2007; Franzén et al., 2011; Wing et al., 2011). This would be favoured by the position of the haptic device just in front of the subjects (Rabin et al., 1999). With vision, the adaptation rate of the spectrum amplitude is smaller in ML than in AP direction, likely because the large distance between the feet minimises the role of vision in ML balancing (Day et al., 1993; Šarabon et al., 2013a) and assures a safe ML sway from the onset of the trials.

Potential Processes Underpinning Adaptation

The adaptation process of body sway, described here by geometrical and spectral analysis, can hardly be traced to a few definite sources. As mentioned, foot plantar and proprioceptive input from many muscles must be enhanced by standing on foam. In a sense, standing on a compliant base of support might have analogies with some sensory augmentation conditions, and adaptation to this state implies several different operations (Sienko et al., 2018). At least some of the mechanisms implied in the adaptation to repeated predictable perturbations should not be disregarded. Stretch reflex modulation is a candidate, and changes in its excitability have been shown during adaptation processes (see Rothwell et al., 1986; Schmid and Sozzi, 2016; Sozzi et al., 2016; Mrachacz-Kersting et al., 2019). Here, the rate of the adaptation with eyes closed reminds that observed in young subjects exposed to continuous predictable perturbation of stance by sinusoidal AP displacement of the supporting platform (Sozzi et al., 2016). In that case, as much as in the present investigation, a progressive decrease in the leg muscle activity was present with eyes closed but not with eyes open, where muscle activity was smaller.

In the absence of vision, the nervous system would exploit a sensory inflow to which it is unaccustomed (standing on foam), by initially increasing muscle stiffness (Winter et al., 1998; Craig et al., 2016) to later learn to appropriately reweigh and select the helpful and cancel the disturbing information. We suggest that when subjects realise the absence of a real risk of toppling over by standing on foam [as well as during continuous predictable perturbations as in Castro et al. (2019) in older adults] they feel safe, reduce postural muscle co-contraction, and energy expenditure (Welch and Ting, 2014) and accept larger, slower CoP excursions (Karmali et al., 2021). This would lead to a reduction of the median frequency of the spectrum. Moreover, the progressive increase in the amplitude of the low-frequency excursion is compatible with the high vestibular thresholds at these frequencies (Valko et al., 2012), as if the adaptation process shifted body oscillations toward those at which the vestibular input can be best exploited (Karmali et al., 2014). Of note, in spite of a substantial difference between the studies, activation of the midline fronto-central cortex is associated with adaptive behaviour to repeated postural perturbations unpredictable in timing much as occurs when standing on foam (Mierau et al., 2015; Varghese et al., 2019).

In our experiment, adding touch to EC further enhances the shift toward the low frequencies of the spectrum. Light-touch would “optimise” the processing of the relevant sensory information, both from the body and the touching limb as well, leading to sparing of motor actions around 0.4 and 2 Hz, possibly by enhancing coordination of muscle actions (Albertsen et al., 2012). Consequently, the control of stance moves from a strategy whereby a high oscillation rate dominates postural control to one where slower CoP sway predominates (Rasku et al., 2012).

We would speculate that a sway "threshold" exists, below which the adaptation mechanisms would hardly (EO, with or without LT) be called into action. For example, in our case (see **Figures 2**, **4**), the threshold should be just below 0.2 Hz for median frequency values and just below 2 m (over 90 s) for path length. Which is or are the responsible excessive-sway-detecting receptors or the brain centres possessing some kind of "velocity-storage" mechanisms would require a different experimental and analytical approach (Jeka et al., 2004; Assländer and Peterka, 2016; Appiah-Kubi and Wright, 2019).

Rambling and Trembling

The reduction of path length is accompanied by an increase in the amplitude of the low-frequency displacements of the CoP with a reduction in the median frequency of the full spectrum. A description of body sway has identified two processes under the names of rambling and trembling (Zatsiorsky and Duarte, 1999, 2000). Rambling has been suggested to be the expression of supraspinal control, while trembling attests spinal control. During prolonged standing, large-amplitude changes in rambling may be observed. Adaptation on foam could be instructed by the same processes originally described for stance on solid ground (Duarte and Sternad, 2008). The "trembling" component might imply inordinate CoP excursions, not immediately filtered by the supra-spinal modulating influences. The high frequencies might be more unfavourable than serviceable for the "exploration" task. The progressive reduction in trembling can avoid the blurring cutaneous and proprioceptive inputs from the feet and legs, and reduce the neural and muscle cost. However, the median frequency diminishes to only about half of the initial values without vision, where adaptation is obvious and declines more smoothly with vision (except with vision *and* touch in the ML direction, where no adaptation is obvious). Therefore, a certain "share" of trembling is present and continues to control sway and help equilibrium maintenance (Yamamoto et al., 2015; Gerber et al., 2022), probably because difficult standing tasks favour automaticity (Haith and Krakauer, 2018) compared to tasks requiring less cognitive involvement (Hsiao et al., 2020; Leung et al., 2021). The shift from trembling to rambling would be controlled and checked by the cerebellum and higher centres (Reynolds, 2010; Colnaghi et al., 2017; Patikas et al., 2019). Anyhow, we cannot leave out the intriguing observation that a certain frequency range (~W3) was largely unaffected by the adaptation. This frequency is intermediate between rambling and trembling, where the former vanishes and the latter starts to grow (Zatsiorsky and Duarte, 1999). Further, as the lower frequencies increased in amplitude, higher frequencies decreased with adaptation. The unvarying W3 window shows that adaptation is not a generalised depression of some collective activity, but reflects a hard-wired neural mechanism. There is an edge in the low-frequency CoP oscillations, whereby slow "rambling" displacements overcome fast, small, higher-frequency "trembling" components (Zatsiorsky and Duarte, 2000; Mochizuki et al., 2006; Yamamoto et al., 2015). Certainly, rambling becomes progressively more important during the adaptation process described here. This occurs also with EO and EO-LT, where adaptation is less obvious, and where the high frequencies are less represented than without vision (EC and EC-LT). Interestingly, older subjects show reduced "trembling" and increased "rambling" frequencies (Šarabon et al., 2013b), much as happens with younger subjects during the adaptation process.

The Shift to Low-Frequency Oscillations Does Not Reflect Enhanced Automaticity

Our subjects made deliberate efforts to control balance, certainly at the beginning and throughout the standing trials as well. During the adaptation period, they would continuously seek to anticipate a forthcoming instability and produce preparatory or expectancy activity, in addition to controlling instability. However, no subject could trace any distinct voluntary activation of selected postural muscles when asked at the end of the session. In a sense, they seemed to implicitly learn to cope with the compliant base of support before the end of the trials.

The stabilising effect of touch is in keeping with its capacity to modulate the responses to unanticipated perturbation of stance (Martinelli et al., 2015), an effect accompanied by a decrease in muscle stiffness and prevalence of reciprocal activity in antagonist's muscles (Sozzi et al., 2013; Dos Santos et al., 2019; Kaulmann et al., 2020). Light-touch with the constraint of keeping force level below 1 N is a precision task, and a dual-task might be elicited (Rabin et al., 1999; Chen and Tsai, 2015). Under different experimental conditions, sensory attention tasks modify the integration of proprioceptive input into the motor cortex, modulating the cortical learning processes (Bolton et al., 2011; Rosenkranz and Rothwell, 2012; see for a recent review, Dijkstra et al., 2020). In our case, light-touch required some attention, even if quite different from an explicit arithmetic task (Vuillerme et al., 2006; Honeine et al., 2017; Lee et al., 2019). A decreasing contribution from the visual system, with a concurrent increase in contribution from the cerebellum and vestibular system in dual-task conditions, represents a shift from controlled to automatic postural behaviour (Lang and Bastian, 2002; St-Amant et al., 2020). The automatic behaviour would show a greater contribution of higher frequency bands in cognitive task conditions, though (Potvin-Desrochers et al., 2017; Richer and Lajoie, 2020), which is different from what we found with adaptation.

The interference of a dual-task with balance control is an open issue (Chen and Tsai, 2015; Bayot et al., 2018; Costa et al., 2021), including the simultaneous performance of a cognitive task with finger-touch stabilisation (Lee et al., 2018; Dos Santos et al., 2019). Here, touch does not increase sway compared to no-touch (without and with vision). Hence, no dual-task property should be conferred to the standing behaviour by light touch, at least not of greater moment than free viewing, which likely requires some attention as well (Bonnet et al., 2017). Possibly, the control of stance is prioritised when standing on foam (contrary to the "posture-second strategy," see Bloem et al., 2006 in a different context), or else maintaining the light fingertip force is an easy task, as shown by the lack of changes over time in the fingertip force produced. In addition, a light touch is strongly stabilising in itself, thereby likely requiring a minor level of attention devoted to the control of standing upright.

The Function of the Adaptation

There is no doubt that task repetition leads to improved stance control, for example, assessed behaviourally, where the "time in balance" increases with short-term training (Schedler et al., 2021). Getting closer to the underpinning mechanisms, we suggest that, when subjects stand on foam, the brain soon realises that minimisation of the body displacement *per se* is not an efficient way of coping with the critical condition (Kiemel et al., 2011). Conversely, the goal would be to set an oscillation amplitude compatible with minimal but *effective* muscle activation. In the end, a passive rigid body, even with a non-point-like base of support, falls more easily on a compliant than the solid base of support, whereas a continuous excursion of the CoP allows appropriate activation of selected muscles (be it reflex or voluntary) to create the necessary torques for adaptively controlling the excursions of the centre of mass. The reasoning is in keeping with several previous findings, based on theoretical and experimental approaches, which point to the inadequacy of stiffness *per se* to maintain equilibrium (Morasso and Schieppati, 1999; Moorhouse and Granata, 2007; Kiemel et al., 2011; Gorjan et al., 2019; Nandi et al., 2019). As a consequence, the oscillation frequencies gradually move toward low values *and* the high frequencies tend to disappear, while the overall level of the spectrum does not decrease but rather increases because of the large contribution of the low frequencies. This pattern is roughly common to the four sensory conditions, naturally graded to the overall amplitude of the spectra (e.g., quantitatively smaller with EC-LT than EC). Notably, even where the markers of adaptation are modest (as path length or median frequency in the stabilised EO and EO-LT conditions), a certain increase in the low frequencies is still present in the adapted trials, while the higher frequencies change little or diminish. It seems safe to conclude that adaptation modifies the quality of the oscillation, which becomes greater in amplitude but slower, so that balance control shifts from a stiff attitude to a more relaxed attitude and slower motion (Karmali et al., 2021). These conclusions do not seem to contradict a hypothesis put forward by Cherif et al. (2020). Their experimental setup is quite complex, but it might be more close to the unsophisticated foam-standing protocol, inasmuch as it delivers perturbations of the base of support to which subjects must react producing a focused corrective muscle activity. Their findings show that learning a force-accuracy control mode, producing minimisation of acceleration, was more effective than minimising sway by stiffness control.

Differences in the Adaptation Pattern in the Geometrical and Spectral Measures

In conformity with published data (Tarantola et al., 1997), where subjects stood for repeated trials without vision on a solid base of support, reduction of path length over time occurs here when standing on foam with EC. About the same decrease is observed when touch is added to EC, despite an overall shorter path length. Sway area shows a trend over time as well, whereby its value moderately increased in the stabilised conditions (EC-LT, EO, EO-LT). This increase in sway area is not obvious with EC, where the area is by far the largest of all conditions. Therefore, assuming and not granting that both path length and sway area are considered appropriate tools for judging postural stability (Danna-Dos-Santos et al., 2008), these do not appear to be the most apposite markers for addressing the various aspects of adaptation of stance over time.

Moving to the spectral analysis, a different picture emerges. A clear-cut reduction in the median frequency of the full spectrum occurs with trial repetitions for the CoP oscillations along with both the frontal and the sagittal plane. The median frequency drops from more than 0.3 Hz to less than 0.2 Hz with EC and EC-LT, without notable changes in the mean level of the spectrum in the ML direction. The changes in the median frequency originate in an increased amplitude of the low- and a decrease in amplitude of the high-frequency windows, respectively. The findings suggest that conventional parameters, such as sway length (or velocity) and amplitude (Hufschmidt et al., 1980), do not provide sufficient information regarding a person's ability to maintain an upright stance (see Gerber et al., 2022).

Limitations

We have restricted the analysis to the spectral frequencies below 2 Hz, as we did in a recent paper (Sozzi et al., 2021). The spectrum level in the 0.01–2 Hz range corresponds to about 98% of the entire spectrum from 0 to 70 Hz. Our choice was in line with other studies that have limited the analysis to this range (Hayes, 1982; Day et al., 1993; Zatsiorsky and Duarte, 1999; Rougier and Farenc, 2000; Mezzarane and Kohn, 2007; Zhang et al., 2007; Demura et al., 2008; Mahboobin et al., 2009; Halická et al., 2014; Kanekar et al., 2014). Others have extended the range of interest up to higher frequencies. Some authors have posited that frequencies in a certain range (around 1.5–6.5 Hz) reflect the somatosensory contribution to balance control and represent the "moderate" band (Taguchi, 1977; Krafczyk et al., 1999). The lower band would express the contributions from the cerebellum (Diener et al., 1984), the very-low bands from the vestibular and the visual systems (Soames and Atha, 1982; Chagdes et al., 2009). We admit that the information on the exact frequencies at which distinct effects of diverse sensory input occur has not been acquired here, even if tackled by Sozzi et al. (2021) for vision and touch. The absence of recording of electrical activity from the many muscles potentially contributing to the CoP oscillations prevents a direct match of the changes in the spectral frequencies to the modulation of muscle activity.

The adaptation process has been studied by repeating 90 s standing trials with a short interval between trials not superior to half a minute, intended to soothe the subjects and discontinue peripheral processes like receptor adaptation or muscular fatigue. The 90 s trial duration and the number of successive trials has little correspondence with previous studies, possibly leading to incongruences in the findings (Cofré Lizama et al., 2016). However, the relatively long duration of the acquisition for each trial seemed appropriate based on previous research (Winter et al., 1998; Michalak and Jakowski, 2002; Sozzi et al., 2021). This epoch was not further divided into time segments and formally analysed to see if an adaptation occurs between the first and last part of the same trial. A related unanswered issue is the duration and amplitude of the adapted effects and the capacity to exploit learning for coping with difficult tasks (Schedler et al., 2021). In other words, we do not know whether and how post-effects of adaptation in sway or spectral measures fade over time.

In addition, there was some inter-subject variability in all sway metrics. Whether this reflects idiosyncratic postural sway in different subjects (Di Berardino et al., 2009; Yamamoto et al., 2015; Sakanaka et al., 2021) is not easy to tell based on our findings. In addition, we have not clustered the subjects into groups presumably showing different visuo-postural dependency (Lacour et al., 1997; Rasku et al., 2012) or differences in the body resonant frequency (Tarabini et al., 2014). It is also possible that different subjects became more or less tired out toward the end of the trials, despite none complaining of fatigue or dizziness, or requested to stop the trial, or asked for a longer rest period to be granted between trials. It must also be mentioned that only young adults but not younger or older subjects were investigated in this study. By exploiting a frequency-based analysis, Pauelsen et al. (2020) have recently shown issues in postural control in older adults with fall-related concerns and declining strength.

CONCLUSION AND PERSPECTIVES

The present investigation exploited the use of the frequency analysis of the CoP time series in a protocol implying prolonged standing on foam. The emerging picture is that repetition of stance trials leads to definite modulation of the standing behaviour. This consists of distinct but moderate changes in path length or sway area, and in qualitative and quantitative changes in the frequency content of the sway and in the amplitude of the frequency spectra.

The adaptation pattern reflects the current sensory conditions. Sway area increases over time, particularly in the stabilised conditions (EC-LT, EO, EO-LT) whereas the median frequencies of oscillation move toward low values, particularly without vision (EC and EC-LT). This occurs by enhancing the relative amplitude of the very low frequencies and reducing the higher frequencies over time. Throughout this process, the control of balance would be shifted from the lower to the higher nervous centres, with the aim to functionally incorporate the integrative capacities of the cortex and resolve the sensory ambiguity (Lhomond et al., 2021). In the sagittal plane, the mean level of the spectrum slowly increases in amplitude, regardless of its initial value, more than in the frontal plane, concurrently with a steeper rate of increase of the low-frequency windows. The different effects on the frontal or sagittal plane suggest that standing subjects can implicitly learn how to recruit the optimal strategies for controlling unstable stances. Of note, balance in the frontal plane is precarious in elderly subjects (Lord et al., 1999), where multiple sensory conditions can degrade the postural control (Morrison et al., 2016; see Paillard, 2021). These data give a rationale, though still preliminary, for explaining the results of training balance on a compliant support base (Strang et al., 2011) compared to other training modes (see Taube et al., 2008; Hirase et al., 2015; Nagy et al., 2018; König et al., 2019). If our paradigm captures some of the underlying causal mechanisms of adaptation, then adaptation features can become a standard marker of deficits in balance control in various populations at risk of falling (Schinkel-Ivy et al., 2016; Ruffieux et al., 2017). Understanding the interaction of balance control with the sensory condition and time has clinical implications. For instance, it would help investigate whether the adaptation rate differs where the postural disorders originate from peripheral neuropathy or a central condition (Asan et al., 2022). In this light, it is notable that patients with Parkinson's disease show undamaged central processing of haptic cues and vision despite their motor problems (Rabin et al., 2013; Engel et al., 2021), even if their adaptability to prolonged standing seems to be impaired (Moretto et al., 2021). Further, challenging balance under the manifold and combined sensory states (Taube et al., 2007; Allison et al., 2018) rather than aiming at enhancing muscle strength (Thompson et al., 2020; Ramachandran et al., 2021) might exert positive effects in persons with precarious balance and older people.

DATA AVAILABILITY STATEMENT

The original data supporting the conclusions of this study are included in the article. Further inquiries can be directed to the corresponding author upon reasonable request.

ETHICS STATEMENT

These studies involving human participants were reviewed and approved by the Ethics Committee of the IRCCS Istituti Clinici Scientifici Maugeri SB. The participants provided their written informed consent to participate in the present study.

AUTHOR CONTRIBUTIONS

MS conceived the idea for the study. SS performed the recruitment of participants and the collection of data. SS and MS performed the data analysis and drafted the article. Both authors approved the submitted version.

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SUPPLEMENTARY MATERIAL

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BRIEF RESEARCH REPORT
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The Effects of Intermittent Trunk Flexion With and Without Support on Sitting Balance in Young Adults

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Prolonged trunk flexion is known to affect passive and active stabilization of the trunk. Previous studies have evaluated changes in spinal range of motion, muscle activity and reflex behavior induced by prolonged trunk flexion, whereas the effect on sitting postural control is vastly underexplored. In this study, we compared the effects of supported and unsupported intermittent trunk flexion on center of pressure (CoP) motion during sitting on an unstable seat. Participants (n = 21; 11 males, 23.2 ± 2.0 years; 10 females, age 24.3 ± 4.0) were exposed to 1-h intermittent (60-s sets with 30 s of rest) trunk flexion (80% of the maximal range of motion) and CoP root mean square distance, velocity and frequency before and after the exposure were assessed. Contrary to our hypothesis, there were no main effects of exposure (pre. vs. post flexion protocol; p = 0.128–0.709), no main effects of condition (supported vs. unsupported; p = 0.134–0.931), and no interaction between exposure and condition (p = 0.163–0.912). Our results indicate that prolonged intermittent flexion does not induce any changes in CoP motion during a seated balance task, regardless of the presence of a trunk support during prolonged intermittent flexion. This suggests a successful compensation of decreased passive stiffness by increased reflex activity.

Keywords: trunk stiffness, reflex gain, postural control, postural stability, spine stability

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INTRODUCTION

In the working-age population, between 20 and 40% of persons suffer from low back pain (LBP) annually (Hoy et al., 2012). Physically demanding professions that include manual material handling and working in awkward postures have been considered to increase the risk of LBP (Heneweer et al., 2011; Griffith et al., 2012; Fatoye et al., 2019), although recent publications report no clear consensus regarding the causality between posture and presence of LBP symptoms (Swain et al., 2020). A strong case for an association between LBP risk and working with twisted or bent trunk has been reported in a systematic review, although cause-and-effect relationship could not be conclusively confirmed (Wai et al., 2010). Prolonged trunk flexion has been shown to alter both passive mechanical properties of the spinal column as well as active control of spinal stability (Sánchez-Zuriaga et al., 2010; Bazrgari et al., 2011; Hendershot et al., 2011; Howarth et al., 2013; Voglar et al., 2016), which is believed to increase the risk of LBP.

The effect of prolonged trunk flexion on spinal stability has been mostly evaluated by measuring maximal lumbar flexion range of motion, which is used as an indicator of the creep deformation and consequent decrease in passive trunk stiffness. Creep deformation after prolonged trunk flexion is consistently shown (McGill and Brown, 1992; Rogers and Granata, 2006; Sánchez-Zuriaga et al., 2010), and probably follows a dose-response relationship (Hendershot et al., 2011; Muslim et al., 2013). On the other hand, the effects of prolonged trunk flexion on intrinsic stiffness (i.e., stiffness due to passive tissues and pre-activated muscles) and reflexive trunk stiffness (i.e., stiffness due to feedback activation of muscles) are less clear. For instance, Hendershot et al. (2011) reported decreased intrinsic trunk stiffness and increased reflex gains after 2 and 16 min of sustained trunk flexion. While the trunk stiffness rapidly returned to baseline levels, the reflex gains remained elevated at least 60 min. In contrast, another study by the same research group reported quick restoration of reflex gains, but slower recovery of intrinsic trunk stiffness following 10 min of sustained trunk flexion, although in males the reflex gains were again elevated 30 min after the onset of recovery period (Bazrgari et al., 2011). Finally, in two studies from another group, both increased and decreased reflex gais were found after exposure to spinal flexion (Granata et al., 2005b; Rogers and Granata, 2006).

In addition to analyzing discrete responses to external mechanical perturbations, evaluation of postural stability through center of pressure (CoP) movement quantification in a sitting posture has also been used to assess trunk stability (Van Dieën et al., 2012; Hendershot et al., 2013; Leban et al., 2017). Sitting postural sway has been shown to increase consistently throughout a simulated shift in crane operators (Leban et al., 2017), as well as throughout long-distance shifts in bus riders (Arippa et al., 2021). These observations could be explained by muscle fatigue, which has consistently been shown to deteriorate standing postural sway (Paillard, 2012). Specifically, trunk extensor muscle fatigue induced by crane operator workload could translate to increased sway in sitting position, as these muscles are paramount for sitting postural control (Curtis et al., 2015). On the other hand, studies investigating electromyographic (EMG) muscle responses indicate that reflexive trunk stability after an exposure to flexion is impaired primarily due to the creep deformation of soft tissues, and not due to muscle fatigue (Sánchez-Zuriaga et al., 2010). Assessing CoP behavior during sitting seems as a promising tool to investigate changes in trunk stability.

To the best of our knowledge, only one study has investigated the effects of exposure to flexion on CoP behavior during sitting (Hendershot et al., 2013). This study showed that both creep deformation and CoP movements increased in a dose-response fashion (2-, 4-, and 10-min exposures were used), while there were no significant differences in recovery patterns between different exposures, with 10 min being sufficient to restore CoP behavior to baseline. It remains to be determined how CoP behavior in seated position is affected by longer flexion periods, and whether the provision of trunk support can attenuate the decrements in sitting postural control. In this paper, we report the results pertaining to seated postural stability, obtained in a larger experiment (Voglar et al., 2016) that was conducted to compare the effects of supported and unsupported prolonged intermittent flexion exposure on trunk stability. The support in the former condition was provided as a padded bar, on which the participants leaned with their chests and shoulders. Total trunk stiffness increased after unsupported flexion only, while reflex gains increased after both conditions. A larger increase in lumbar range of motion and reflex gains were noted following unsupported flexion in comparison to supported flexion (Voglar et al., 2016), which indicates a potentially beneficial effect of trunk support during working in flexed postures. The aim of this paper is to analyze the effects of exposure to prolonged intermittent flexion with and without support on sitting postural stability, assessed through CoP movement recording. Considering previous evidence (Hendershot et al., 2013; Voglar et al., 2016), we hypothesized that CoP movement amplitude and velocity will increase, with a concomitant decrease in CoP frequency, indicating impaired postural control after the exposure to intermittent flexion. We also hypothesized that the provision of support will eliminate or attenuate this effect.

MATERIALS AND METHODS

Participants

Twenty-one young participants were included in the present study [11 males, (age 23.2 ± 2.0 years, height 182.3 ± 6.2 cm, and body mass 73.9 ± 8.2 kg) and 10 females, (age 24.3 ± 4.0) years, height 168.3 ± 7.2 cm, and body mass 62.1 ± 9.0 kg)]. Participants who reported LBP within the last 6 months, or any history of LBP that impaired their physical activity for at least 1 day, were excluded. Moreover, participants with any known sensory or neuromuscular pathologies were also excluded. The experimental protocol was approved by the Ethics committee for Movement Sciences at the Vrije Universiteit, Amsterdam (approval number: ECB 2015–18). All subjects were required to sign an informed consent statement prior to the experiment. The study was conducted in accordance with the Helsinki Declaration.

Pilot Experiment

The purpose of the pilot experiment was to establish the number of trials of seated balance assessment needed to obtain reliable performance. For this purpose, 12 participants (seven males, five females; age: 28.2 ± 5.8) performed a two-session experiment, with 10 repetitions of the seated balance task (see Section Sitting Balance Assessment for details). The reliability in a pilot experiment was assessed with single-measures, two-way random model intra-class correlation coefficients (ICCs) for absolute agreement. Reliability was considered as excellent when ICC was > 0.90, and good when ICC > 0.75 (Koo and Li, 2016). Values below 0.75 were considered as unacceptable. None of the participants in the pilot experiment participated in the main experiment.

Study Design

The study protocol consisted of two visits, each consisting of one of the two exposure conditions: supported flexion (SF)

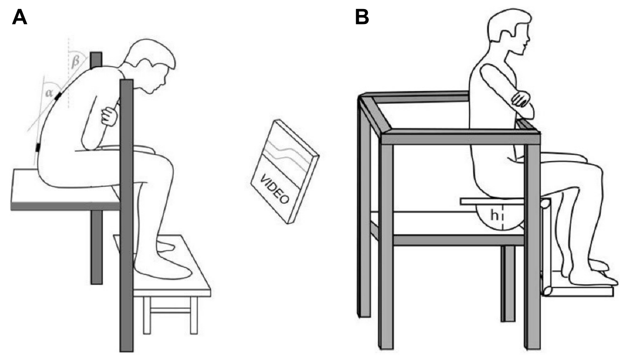

FIGURE 1 | (A) During intermitant flexion exposure participants obtained ~80% of their available lumbar flexion (α) and 35° of trunk inclination (β) using real-time visual feedback. Additionally in the supported condition a padded bar was placed in front of the participant at appropriate height in order to lean on it while obtaining the required trunk position. **(B)** Postural control assessment on a custom-built chair with a hemi-sphere attached below the seat surface [height (h) = 18 cm].

and unsupported flexion (USF) (see Section Intermittent Trunk Flexion Task; **Figure 1A**). The experiment also involved the assessment of trunk reflex gains and range of motion. Sitting balance was always measured after trunk reflex gains and range of motion assessments. The details regarding the full procedure are available in our previous paper (Voglar et al., 2016). Before the baseline measurements, the participants performed a set of measurements followed by siting for 30 min on an office chair with their backs supported against the backrest. This was done to provide a washout period, avoiding any effects of previous activities of the participants. The measurements included assessment of postural control in a sitting posture via CoP movement analysis (see Section Sitting Balance Assessment; **Figure 1B**). After the baseline measurements, participants were exposed to one of the two experimental conditions (SF and USF). The conditions were introduced on separate visits, in a quasi-randomized counterbalanced order, with a minimum of 4 days between visits. After the experimental condition, the sitting postural control assessment test was repeated. In all participants, less than 10 min passed between the end of the exposure and the assessment of postural control. During this time range of motion measurements and two perturbation trials were performed (Voglar et al., 2016).

Intermittent Trunk Flexion Task

The participants were exposed to prolonged intermittent flexion, as described in detail in our previous study (Voglar et al., 2016). Briefly, the participant was seated on a raised platform with the feet supported and rails provided for safety (**Figure 1A**). Real-time feedback on lumbar flexion and trunk inclination was provided (see section Sitting Balance Assessment). Participants were reminded to adjust their position if they drifted more than 2° from the target position as checked by feedback on the computer screen. The target lumbar flexion angle was determined as 80% of the range of motion from erect stance to maximal forward flexion. To standardize loading and avoid that subject would obtain lumbar flexion by slumped sitting, trunk inclination and lumbar flexion were monitored. To determine the target posture, participants flexed forward until the trunk inclination reached 35°, and then adjusted the lumbar flexion position by tilting the pelvis until reaching 80% of lumbar flexion RoM. Target lumbar flexion percentage, inclination angle and exposure duration were adopted from one of the rare previous studies investigating effects of longer lasting (60 min) supported flexion on trunk neuromuscular control (Sánchez-Zuriaga et al., 2010). Both the target flexion percentage and inclination angle were slightly increased in comparison to the aforementioned study. This was done to increase the load on the trunk muscles during the unsupported condition, creating a greater contrast with the condition with trunk support. Intermittent trunk flexion was used to simulate the main characteristics of the harbor crane operator workload. The ratio between flexion and upright sitting was set based on our previous experiences and observation of crane operators work. Perceived discomfort or fatigue were not systematically assessed, but was frequently reported spontaneously during the unsupported condition, but not during the supported condition.

Muscle activity was tracked using surface EMG (REFA, TMSi, Netherlands), with the electrodes placed bilaterally over two segments of the erector spinae muscle (3 cm lateral to interspinous space between L4 and L3 and 6 cm lateral to the L2 spinous process). The details regarding EMG procedures are available in our previous report (Voglar et al., 2016). In case the participant presented with EMG silence of the back muscles due to the flexion relaxation phenomenon, the lumbar flexion angle was reduced until marked activation could be seen. The same lumbar flexion angle was used in both conditions. The flexion task was intermittent and consisted of 40 cycles including 1 min of target flexion level, followed by 30 s of upright active sitting, resulting in total duration of 60 min. Auditory cues were used to indicated the time to change position. In the unsupported flexion condition, a thin rope was placed horizontally to provide the participant with a mechanical orientation to indicate the required trunk inclination. Participants had their hands crossed across the chest and were touching the rope slightly with their shoulders. In the supported flexion the rope was replaced with a padded bar that provided passive support. The participants leaned on the pad with their chests and shoulders.

Sitting Balance Assessment

Sitting postural control assessment involved the analysis of CoP movements whilst the participants were seated on a custom-built chair with a spherical surface attached below the seat surface (radius = 22 cm; height = 18 cm). The seat was placed on a platform where a force plate (KAP-E, AST, Germany) was installed and safety rails provided (**Figure 1B**). To control for task familiarization and ensure reliability, the participants performed six trials, each lasting 1-min, with 1-min breaks in between. The instruction was to stay as still as possible and reach for the rail only in case of losing balance. In case that balance was lost, the trial was repeated (one participant at baseline and one participant after the intermittent flexion). The force plate signal was sampled at 200 Hz and all CoP data were demeaned prior to further analyses. The postural control of the trunk was quantified by three variables (each calculated for anterior-posterior and medial-lateral direction) derived from CoP time series: root mean square (RMS) distance (mm), mean sway velocity (mm/s), and mean sway frequency (Hz), following the study by Larivière et al. (2013), which also showed high reliability of these measures specific for the sitting position.

Lumbar Range of Motion

Pelvis and thorax orientations were estimated using two inertial measurement units (IMU) (Xsens Technologies X-bus, Enschede, Netherlands) positioned at the T12 and S2 level, directly on the spinous processes. Maximal lumbar flexion RoM was calculated as the difference in the inclination angles of the sensors in the sagittal plane. To achieve full lumbar flexion in standing position, the participants were instructed to maximally flex forward while keeping their knees slightly bent. Each participant performed two repetitions at baseline and after the exposure to the intermittent flexion protocol, and the highest value of the two repetitions was used for further analyses. The real-life feedback from the same two sensors was used to control the position of the spine during the intermittent flexion protocol (Section Study Design).

Statistical Analysis

Statistical analyses were done with SPSS (version 25.0, SPSS Inc., Chicago, IL, United States). Descriptive statistics are reported as mean ± standard deviation. The normality of the data distribution was verified with Shapiro–Wilk tests. The inter-visit reliability of baseline values in the main experiment was assessed with ICC, as well as with 1-way repeated measures analysis of variance (ANOVA). Two-way ANOVA was used to analyze the effect of the exposure (pre. vs. post protocol) and the inclusion of support (supported vs. unsupported condition), as well as the interaction between the two main effects. Effect sizes were expressed as partial eta-squared (η^2) and interpreted as trivial (<0.01), small (0.01–0.06), medium (0.06–0.14) and large (>0.14) (Bakeman, 2005). The threshold for statistical significance was set at $\alpha < 0.05$.

RESULTS

Pilot Reliability Experiment

Our pilot experiments showed that 6 repetitions are needed to achieve stable sitting balance performance outcomes. The within-session reliability for CoP velocity was poor (ICC = 0.42) when considering repetitions 1–3, good when considering repetitions 2–4, 3–5, or 4–6 (ICC = 0.77–0.85), and excellent when considering repetitions 5–7 (ICC = 0.94). Considering even higher repetition numbers did not increase the ICC further. The inter-visit reliability for the average of repetitions 4–6 was also close to excellent (ICC = 0.89). Therefore, six repetitions were performed in the main experiment, and repetitions 4–6 were considered for further analyses. **Figure 2** shows the behavior of CoP velocity across trials within the first session of the pilot experiment. Other variables showed very similar patterns (data not shown).

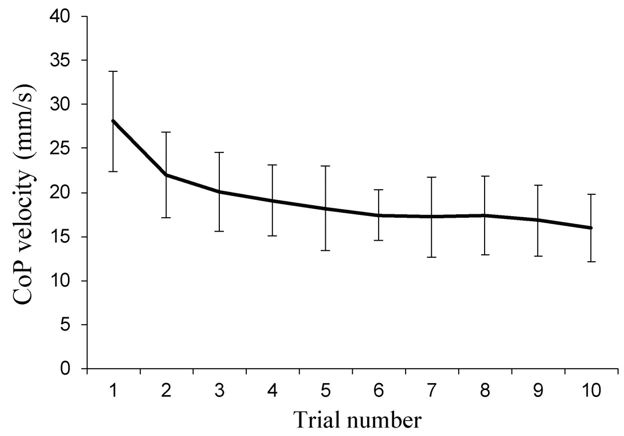

FIGURE 2 | Within-session behavior of center of pressure (CoP) velocity in the pilot experiment (n = 12).

Reliability of Baseline Measures in the Main Experiment

The analysis of variance showed no differences between sessions of baseline values for CoP RMS distance (F = 0.513–0.906; p = 0.352–0.482 over AP, ML and combined directions), CoP velocity (F = 0.396–0.958; p = 0.333–0.536), or CoP frequency (F = 0.015–0.222; p = 0.643–0.905). The ICC indicated moderate reliability for RMS outcomes (ICC = 0.61–0.75) and velocity outcomes (ICC = 0.64–0.72), but good reliability for frequency outcomes (ICC = 0.77–0.84).

Effects of Intermittent Trunk Flexion on Sitting Balance

Intermittent trunk flexion had no effect on CoP RMS distance, CoP velocity and CoP frequency in any direction (**Figure 3**). There were no main effects of exposure (pre. vs. post flexion protocol; F = 0.212–2.515; p = 0.128–0.709 over outcomes and directions), no main effects of condition (supported vs. unsupported; F = 0.003–2.443; p = 0.134–0.931), nor was there any exposure × condition interaction (F = 0.013–2.097; p = 0.163–0.912). All the associated effect sizes were small to medium for the effects of exposure (η^2 = 0.01–0.11), trivial to medium for support condition (η^2 = 0.00–0.11) and trivial to medium for the interaction (η^2 = 0.00–0.10).

DISCUSSION

The purpose of this study was to assess the effects of supported and unsupported prolonged intermittent trunk flexion on postural control during sitting. Based on the previous studies reporting increases in CoP movements after much shorter exposures to flexion (Hendershot et al., 2013), we expected to observe an increase in CoP movements. In addition, we

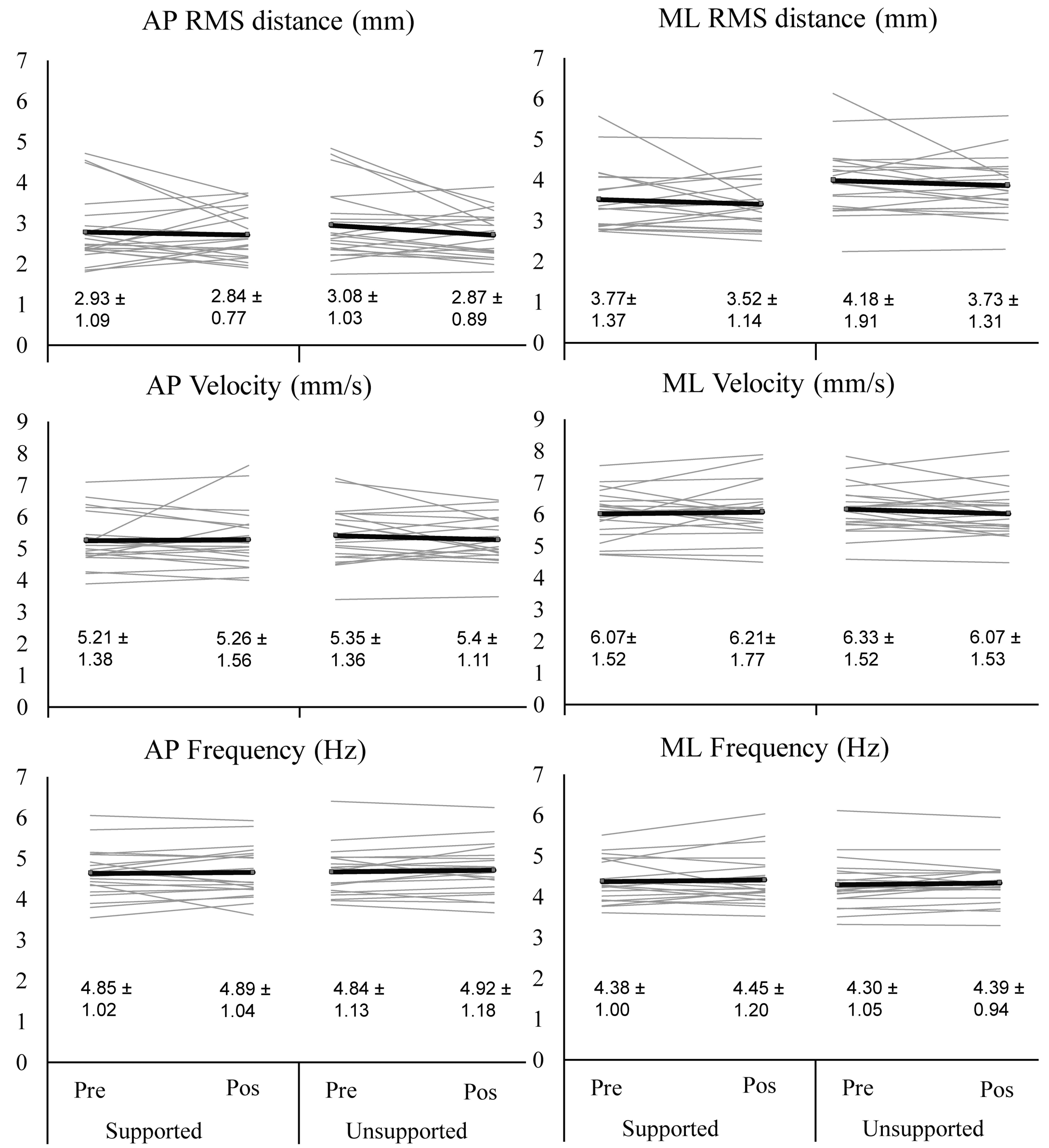

FIGURE 3 | Center of pressure RMS (root mean square) distance, velocity and frequency before (Pre) and after (Pos) the exposure to supported and unsupported flexion. Gray lines indicate individual datapoints, and the black lines represent mean values. Note that combined antero-posterior (AP) and medio-lateral (ML) trajectory showed similar results and none of the effects were significant. The values below the lines are mean ± standard deviation.

hypothesized that trunk support would attenuate or eliminate the effects of prolonged flexion in CoP movements, as we previously found that support attenuated or negated effects of lumbar flexion exposure on lumbar range of motion, admittance and reflex gains (Voglar et al., 2016). However, the hypotheses of the present paper were rejected, as the CoP movement was not affected by prolonged intermittent flexion, and there was no difference between supported and unsupported conditions. This was somewhat surprising, as we reported in the previous paper that the range motion was increased in both conditions, confirming the presence of viscoelastic deformation of passive tissues, which should decrease the stability of the trunk. It could be that the decrease in passive stiffness was successfully compensated by increase in reflex gains (Voglar et al., 2016). As reported in the previous paper (Voglar et al., 2016), the activation of erector spine (pars lumborum and pars iliocostalis) in response to forward directed force (60 N) was increased after the exposure to intermittent flexion and the increase in activation was higher following the unsupported condition. The increase in muscle activation could be a result of an increased neural drive to compensate for a reduced force production capacity of fatigued muscles and reduced intrinsic stiffness.

Contrary to our findings, Hendershot et al. (2013) reported significant increases in CoP RMS and velocity after 2–10 min of exposure to sustained flexion. Moreover, an increase in sitting postural sway was seen throughout a simulated crane operator shift (Leban et al., 2017). An important limitation of our study is the break (5–10 min) between the end of the flexion exposure and sitting balance assessment. It could be that this time was sufficient to restore CoP behavior to baseline. Another possible explanation for these discrepancies is the nature of the task, as we used intermittent flexion, whereas Hendershot et al. (2013) used sustained flexion. However, considering that changes induced by trunk flexion are reported to last longer than the exposure itself (McGill and Brown, 1992; Hendershot et al., 2011), we think it is unlikely that the breaks in our study were sufficient to eliminate all the effects. Another issue to consider is the range of motion, as we determined flexion position at 80% of full range of motion. The angle used in the flexion task is a significant determinant of the changes in trunk stiffness and reflex gain (Hendershot et al., 2011). Moreover, previous studies have also reported significant creep deformation and alteration of reflex behavior following the exposure to flexion at non-maximal ranges of motion (Sánchez-Zuriaga et al., 2010; Hendershot et al., 2011). Finally, the first three repetitions in our pilot experiment showed poor agreement (ICC = 0.42), indicating the need for sufficient familiarization before baseline assessment. Therefore, we used an extensive familiarization procedure to ensure stable baseline performance before proceeding with the exposure to trunk flexion. It could be that previous results regarding sitting postural sway (Hendershot et al., 2013) were confounded by poorer reliability, although learning effects would work in the opposite direction.

Despite the factors discussed above, a complete lack of effect of prolonged intermittent flexion on sitting postural control is surprising, considering that previous studies have shown alterations in intrinsic trunk stiffness and reflex behavior (Granata et al., 2005b; Rogers and Granata, 2006; Sánchez-Zuriaga et al., 2010; Bazrgari et al., 2011; Hendershot et al., 2011; Voglar et al., 2016). It seems that the body can successfully adapt to effects of fatigue by increasing background muscle activation (Grondin and Potvin, 2009) and possibly by increasing reflex gains (Hendershot et al., 2011). Moreover, our previous finding of increased reflex gains alongside significant creep deformation, supports the assumption that the body can successfully compensate for decreased passive stiffness (Voglar et al., 2016). As trunk muscle activation is increased when the sitting surface becomes less stable (Oomen et al., 2015), such an increase in muscle activation likely enhances stabilization. Importantly, the changes in reflex behavior following prolonged flexion seem to be driven primarily by creep deformation, and to a lesser extent by muscle fatigue (Sánchez-Zuriaga et al., 2010; Voglar et al., 2016). However, postural sway is known to be heavily influenced by both local and global muscular fatigue (Springer and Pincivero, 2009; Paillard, 2012; Van Dieën et al., 2012; Garcia-Gallart and Encarnacion-Martinez, 2019), as well as cognitive fatigue (Noé et al., 2021). Therefore, monitoring muscular and cognitive fatigue during/after exposure is recommended for easier interpretation of results pertaining to CoP movements. Despite the fact that lumbar muscles were more active in the unsupported condition (Voglar et al., 2016), the resulting fatigue, if any, did not affect CoP behavior during sitting.

Limitations

There are a few limitations of the present study that need to be acknowledged. Firstly, a convenience sample of young healthy adults was used. It could be that different responses would be seen in participants of different ages, LBP history, and professions. Next, the sample size of the study was relatively small. It could be that differences in CoP variables would be seen if larger sample was used. Given the high variability across participants, a very large sample size would likely be needed. Note that effects would be small relative to this variability and hence likely not important. While the protocol was well-controlled, performing a task that would simulate actual work activities (e.g., crane operator shift) would increase the ecological validity of the study. Despite providing extensive familiarization and thus high reliability, another limitation of our measurements is a rather high between-participant variability, with coefficients of variation ranging from 21.4 to 51.3%. This could mask small changes induced by exposure to prolonged trunk flexion. Moreover, postural stability is a phenomenon with many underlying factors, including sensory contributions from vestibular, visual and somatosensory systems (Horak et al., 2009), sensory integration (Goodworth and Peterka, 2009), reflex behavior (Chen and Zhou, 2011), and joint stiffness (Sakanaka et al., 2021), contributions of which are in part modulated to deal with surface instability (Andreopoulou et al., 2015) and other task constraints (van Drunen et al., 2015). Moreover, small increases in CoP movements are not necessarily indicative of deteriorated balance, but may simply reflect altered balance maintained strategy, which may not be related to injury risk (Shumway-Cook and Woollacott, 2017). Somewhat different responses across participants (**Figure 3**) could indicate that they used different strategies to adapt to the changes induced by the exposure to flexion. It is not clear whether small increases in CoP movement actually reflect decreased stability during sitting. It is suggested that future studies explore the effects of trunk flexion on postural control in more detail. For instance, it would be interesting to explore specific strategies used to maintain trunk stability during sitting. In LBP patients, an increased reliance on co-contraction and lower reliance on cognitive control for static standing posture has been reported (Kiers et al., 2015). Co-contraction may not be an optimal strategy to maintain balance, as it is associated with increased spinal compressive forces (Granata et al., 2005a).

CONCLUSION

In conclusion, regardless of the presence of a trunk support, prolonged intermittent flexion did not induce any changes in CoP behavior during a seated balance task. This suggests a successful compensation of decreased passive stiffness by increased reflex activity. Future studies should assess the effects of longer exposures to flexion on postural control, possibly simulating real-life work shifts (e.g., crane operator shift).

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics committee for Movement Sciences (Ethische Commissie Bewegingswetenschappen) at the Vrije Universiteit, Amsterdam. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MV, IK, JD, and NŠ conceptualized the idea. MV carried out the measurements. JD and NŠ were overviewing the measurement procedures and administration. MV and ŽK analyzed the collected data and wrote the first draft of the manuscript. All authors worked on finalizing the manuscript and approved the submitted version.

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Decreased Postural Complexity in Overweight to Obese Children and Adolescents: A Cross-Sectional Study

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Introduction: Although a few studies suggest that young overweight to obese children and adolescents (YO) may have impaired postural control compared to young normal-weight (YN) peers, little information exists about how these two groups differ in the quality of the underlying balance strategies employed. Hence, the aim of the present study was a first comprehensive examination of the structural complexity of postural sways in these two cohorts during quiet bilateral standing.

Methods: Nineteen YO secondary school students (13.0 ± 1.4 years; male = 10, female = 9) were carefully matched to YN controls (13.0 ± 1.5 years) for age, sex, height, and school. Mediolateral (ML) and anteriorposterior (AP) acceleration signals were recorded with an inertial measurement unit (IMU) positioned at the trunk while standing barefoot in two conditions: firm and foam support surface. The magnitude of postural fluctuations was obtained using the root mean square (RMS). The temporal structure of the signals was analyzed *via* sample entropy (SEn), largest Lyapunov exponent (LyE), and detrended fluctuation analysis (α-DFA) algorithm. Reliability was assessed using a test–retest design.

Results: In both groups, foam standing caused higher postural fluctuations (higher RMS values) and reduced structural complexity (lower SEn values, higher LyE values, higher α-DFA values). In comparison to YN, YO exhibited a higher RMS_{AP}. Especially in ML direction, the acceleration signals of the YO had higher repeatability (smaller SEn values), greater long-range correlations (higher α-DFA values), and lower local stability (higher LyE values). However, these observations were largely independent of the task difficulty. Except for $\alpha\text{-DFA}_{AP}$, the IMU approach proved reliable to characterize posture control.

Discussion: Our outcomes confirm postural control deficits in YO compared to their YN peers and indicate impaired regulatory mechanisms reflected as rigidity. Such less complex patterns usually reflect diverse pathologies, are detrimental to compensate for internal or external perturbations, and are attributed to lower adaptability and task performance. Without targeted balance stimuli, YO likely end in a lifelong vicious circle of mutually dependent poor balance regulation and low physical activity.

Keywords: inertial measurement unit, root mean square, sample entropy, largest Lyapunov exponent, detrended fluctuation analysis, reproducibility

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INTRODUCTION

According to the most recent report published by the Non-Communicable Disease Risk Factor Collaboration, more than 340 million children and adolescents aged 5–19 years worldwide were overweight or obese (girls: 18% and boys: 19%) in 2016 (Abarca-Gómez et al., 2017). Increasing physical activity would be a valuable instrument to counteract this pediatric condition and its detrimental effects on health (Kelley et al., 2014). However, several studies have discovered a negative association between adiposity levels and physical activity participation in childhood and adolescence (Jimenez-Pavon et al., 2010; Prentice-Dunn and Prentice-Dunn, 2012). Following, for example, McGraw et al. (2000), a contributing factor for the poor adherence to physical activity in overweight to obese children and adolescents (YO) could be their well-established difficulties in maintaining balance (Steinberg et al., 2018). Consequently, YO likely feel less confident to execute basic motor tasks and thus less motivated to engage in sports and exercise than their young normal-weight peers (YN; McGraw et al., 2000; Jimenez-Pavon et al., 2010; Prentice-Dunn and Prentice-Dunn, 2012).

Postural control deficits in YO have been systematically demonstrated with static post-urography, particularly under challenging conditions like standing under proprioceptive manipulation (e.g., D'Hondt et al., 2011; Steinberg et al., 2018). Center of pressure (CoP) fluctuations, usually registered with force plates, are acknowledged as not purely random but to have a deterministic origin containing crucial information about the time-evolving dynamics of balance control. However, these structural features embedded in the time series remained ignored by previous studies, that only focused on the outcome of performance by analyzing the magnitude instead of the structure of CoP variability in YO. Best to our knowledge, only three studies examined CoP trajectories of YO at the underlying control level. They indicated proprioceptive impairments (Fink et al., 2019), less sensitivity to correct for small shifts in the CoP (Villarrasa-Sapina et al., 2016), and altered behavioral strategies governing standing (Pau et al., 2012). Given the few original and explorative studies on measures of the structure of variability, it remains unclear to what extent an excessive body mass in childhood and adolescence affects the collective regulation of the postural system, or in other words, its fundamental patterns of coordination during postural balance tasks. Hence, additional research is required to provide more detailed insights into the complexity of these regulation processes unveiled by non-linear dynamical analyses and give trainers and clinicians better knowledge to promote more sustainable exercise behavior in this specific cohort.

Unlike the perturbed conditions in previous non-linear analyses of YO, we used a foam surface to enhance the task demands in a manner that challenges the whole sensorimotor balance control system and is significant for a plethora of physical activities. In addition, we aimed to perform an exhaustive but routinely applicable assessment of postural balance in YO and YN by unfolding postural control dynamical properties *via* the first parallel use of the most prominent non-linear tools, known as sample entropy (SEn), largest Lyapunov exponent (LyE), and detrended fluctuation analysis (α-DFA; Stergiou and Decker, 2011; Mukherjee and Yentes, 2018). In brief, SEn estimates the rate of regularity, LyE extracts the trajectory divergence, and α-DFA quantifies long-range correlations (Peng et al., 1995; Richman and Moorman, 2000) in the signal in question. In contrast to the amount of variability, each of these techniques evaluates the structural characteristics embedded in the CoP trajectory from different conceptual perspectives (e.g., Goldberger et al., 2002; Stergiou and Decker, 2011). However, all output measures of SEn, LyE, and α-DFA belong to the same construct, usually called complexity (Roerdink et al., 2006; Lamoth et al., 2009; Lamoth and van Heuvelen, 2012; Mukherjee and Yentes, 2018). High complexity in human motion is thought to reflect a rich repertoire of coordinative solutions (Harrison and Stergiou, 2015) that allow the controller to readily respond to the diverse internal and external stressors encountered in daily life. Inversely, pathology, aging, increased task demands, or a lack of physical experience are prominent causes that render systems physiologically less complex and thus less adaptive (Goldberger et al., 2002; Roerdink et al., 2006; Lamoth et al., 2009; Stergiou and Decker, 2011; Lamoth and van Heuvelen, 2012; van Emmerik et al., 2016). Although not uncontroversial, a loss of complexity in CoP displacement of static posturography tests would be accompanied by a breakdown in the number of active degrees of freedom and, as a result, maladaptive responses to perturbation (Lipsitz, 2002). Accordingly, postural sway dynamics of YO and elderly typically show more repetitive (analyzed via SEn) or less locally stable (analyzed via LyE) patterns or a decrease in the fractal organization of the scaling exponent α (analyzed via α-DFA) (Ko and Newell, 2016; Villarrasa-Sapina et al., 2016).

Since single non-linear metrics cannot comprehensively quantify the complexity of a one-dimensional time series (e.g., Goldberger et al., 2002; Stergiou and Decker, 2011), the application of three complementary mathematical tools is a strength of our study. In this way, we reveal the overall impression of the temporal organization of the sway signal while reducing the risk of misinterpretations. Considering the earlier findings described above, we expected a reduced postural complexity in terms of higher repeatability (lower SEn values), lower local stability (higher LyE values), and higher long-range correlation (higher α-DFA values) in YO compared to their YN peers. Further, we hypothesized that this functional deficit would become even more evident in the foam standing condition. Since sway data were captured using simple accelerometry within a non-laboratory setting, we also tested the reliability of the methodology employed.

MATERIALS AND METHODS

A two-group cross-sectional comparative design was used to examine the hypothesis. Participating children were recruited from three local secondary schools, and all subjects were informed about the objectives, risks, and benefits of the study before providing their written consent. Signed parental permission was sought for children younger than 14 years. The project was approved by the local ethics review board

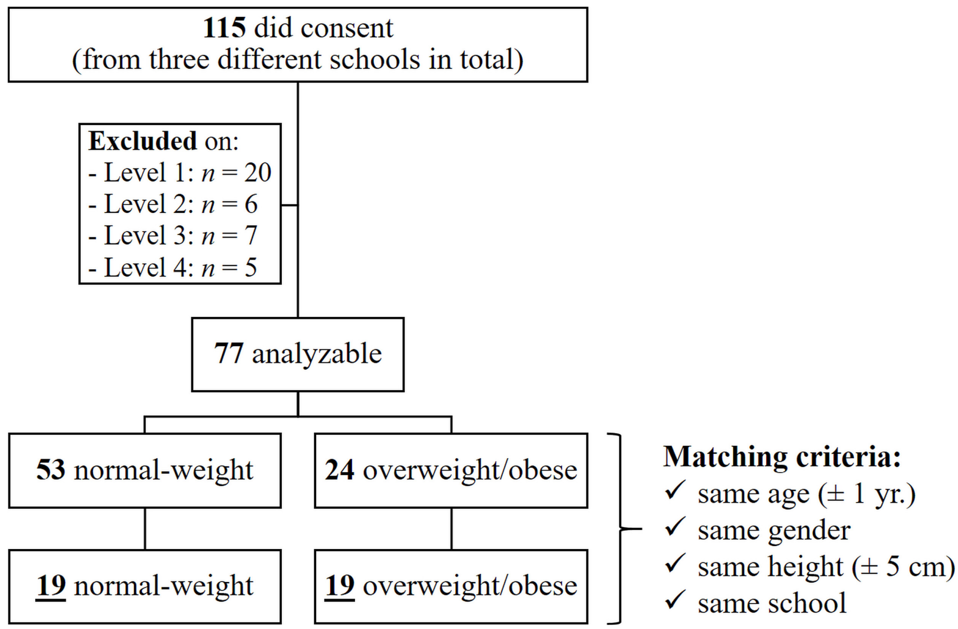

FIGURE 1 | Summary of volunteer recruitment and participation. Children were excluded due to self- or parental-reported acute injury or orthopedic, neurological, cardiac, or cognitive disorders, including medication that could affect balance (Level 1), or in case one of the following applied during any test session: unavailability (Level 2), feeling unwell or non-compliant behavior (both Level 3), or corrupted signals (Level 4).

TABLE 1 | Group statistics of the young overweight/obese (*n* = 19) and their matched normal-weight controls (*n* = 19).

	Overweight/Obese	Normal-weight	t_{36}	Difference BCa 95% CI	*p*	*f*
Girls/Boys	9/10	9/10				
Age (Years)	13.0 ± 1.4	13.0 ± 1.5	0.09	[−0.88, 0.88]	0.930	0.01
Body height (cm)	158.2 ± 7.9	157.8 ± 8.4	0.14	[−4.80, 5.15]	0.892	0.02
Body mass (kg)	64.0 ± 10.8	46.4 ± 9.6	5.32	[10.92, 24.12]	<0.001	0.89
BMI ($kg \cdot m^{-2}$)	25.4 ± 2.5	18.4 ± 2.1	9.22	[5.64, 8.44]	<0.001	1.54
BMI percentile (%)	93.4 ± 3.7	45.1 ± 22.5	9.22	[38.49, 57.35]	<0.001	1.54

Descriptive statistics are expressed as mean ± standard deviation values. t-values, p-values, and effect size f are presented for the independent t-test. Bias-corrected and accelerated (BCa) 95 percent bootstrap confidence interval for mean differences. Age, body height, and body mass information were obtained during the first visit. BMI, body mass index.

(EK-GZ: 17/2018) and was conducted in accordance with the Declaration of Helsinki.

Participants

The detailed participant selection process and the participants' characteristics are presented in **Figure 1** and **Table 1**. In sum, 38 out of the 115 10–16-year-old YO and YN had to be excluded for further analysis due to medical, personal, or technical reasons. Of the remaining 77 children, 24 were overweight or obese, while 53 were normal-weight. In the end, 19 (overweight, *n* = 9; obese, *n* = 10) out of 24 of the YO could be *a posteriori* strictly age-, sex-, height-, and school-matched to a YN control, thus leaving 38 children to be thoroughly evaluated. Age- and sex-specific body mass index percentiles for classifying weight status among children were computed using the online calculator provided by the Centers for Disease Control and Prevention[1]: normal-weight, 5th ≤ BMI < 85th percentile; overweight, 85th ≤ BMI < 95th percentile; obese, BMI ≥ 95th percentile). No child stated to have ever been involved in a weight loss or obesity treatment program.

[1]https://www.cdc.gov/healthyweight/bmi/calculator.html

Data Collection

For reliability testing, the same principal investigator accomplished experimental data collection twice within approximately 12 days (SD: ± 2 days) at the same location and at a comparable time of the day.

Children were tested individually in a quiet room at their regular school during morning classes. After height and weight assessment, the young participants were asked to stand (two-legged) as still as possible with eyes open, first, on a firm, and then on a foam (Balance-pad Solid, Airex, Switzerland) support surface. We refrained from using randomization since we expected higher transfer effects from foam to the firm condition than *vice versa*. Both trials lasted 60 s with a 2 min rest period in between. All tests were performed with minimal clothing and without shoes. The detailed instructions included (a) to look straight ahead at an eye-level target (positioned 3 m away), (b) to keep the feet hip-width apart and the arms relaxed at the sides, and (c) to breath as usual. Simultaneously, postural sway acceleration (ACC) signals were captured through a commercially available 50 mm × 70 mm × 20 mm (mass: 35 g) inertial measurement unit (IMU) (GyKo, Microgate, Italy) sampling data routinely at 500 Hz. The IMU was attached following the manufacturer's recommendation to the children's thoracic spine using the associated semi-elastic vest. The vest was equipped with a specific bib with magnetic support to avoid relative mediolateral (ML) and anteriorposterior (AP) IMU movements during the balance tests. The position of the IMU was noted thoroughly for reproducibility measures. All records were transferred *via* Bluetooth to a personal computer and stored as text files by the manufacturer's RePower software version 1.1.1.7 (Microgate).

Data Pre-processing

Raw ACC signals were initially down-sampled to 100 Hz and cropped to the middle 50 s, yielding 5,000 data points entering in-depth analysis. Further pre-processing steps involved (a) tilt-correction to obtain values concerning the anatomic frame of reference (Moe-Nilssen, 1998) and (b) bandpass filtering (Butterworth filter, 4th order, zero-lag, bandpass frequency: 0.3–10 Hz) to reduce both biases caused by respiration and by low amplitude measurement noise (Ruhe et al., 2010; Martinez-Mendez et al., 2012). Subsequently, given time series were inspected for potential (previously overlooked) abnormalities in task performance (Mancini et al., 2012). More precisely, the resultant ACC vector time series (the square root of the sum of squares of the ACC vector in ML and AP direction) was divided into five 10 s windows. The standard deviation was evaluated in each window. If one of these standard deviation values exceeded the fivefold of the smallest standard deviation value, all data from the particular child were ignored (Mancini et al., 2012). For this reason, three children had to be excluded for further analysis and another two because of unexpected signal gaps (see Level 4 in **Figure 1**).

Outcome Variables Overview

The primary outcome was the measure of postural complexity, including the structure related scaling exponent α-DFA, as well as SEn and LyE estimates. The secondary outcome was the root mean square (RMS) quantifying the magnitude of sway variability. Differences in the dependent variables were assessed separately for ACC_{ML} and ACC_{AP} profiles.

Sample Entropy

The SEn procedure indexes the predictability (regularity or orderliness) of a signal and is mathematically defined as the negative natural logarithm of the conditional probability that a time series of length N, having repeated itself for M samples within a tolerance r, will also repeat itself for $M + 1$ samples, but without allowing self-matches (Richman and Moorman, 2000). Thus, a SEn value becomes zero for a perfectly repeatable signal where sub-sequences exhibit the same configuration upon comparison, and *vice versa*, a perfectly random signal elicits a SEn value converging toward infinity. Healthy human functioning is located in the intermediate region of this continuum (Stergiou and Decker, 2011). Although not undisputed, pathological states typically involve more regular and constrained movements (Goldberger et al., 2002; Stergiou and Decker, 2011; van Emmerik et al., 2016). Here, the input parameters M and r were set in line with guidelines suggested by Ramdani et al. (2009), showing that the referenced estimate of the relative error of the determined SEn values across the present records was as small as possible when $M = 3$ with $r = 0.07$·standard deviation regarding ACC_{ML} data and $r = 0.06$·standard deviation regarding ACC_{AP} data.

Largest Lyapunov Exponent

To extract LyE estimates from one-dimensional observations, an m-dimensional state space was reconstructed in the first step with the method of embedding delays (Rosenstein et al., 1993; Stergiou et al., 2004). Therefore, classical concepts of "average mutual information" and of "false nearest neighbors" were applied to obtain the required delay τ and the embedding dimension m, respectively (Stergiou et al., 2004). In fact, τ was chosen as the first minimum in the average mutual information function, and m when the percentage of false nearest neighbors as a function of the embedding dimension dropped to near zero. Altogether, $m = 4$ with $\tau = 17$ samples for ACC_{ML} and with $\tau = 20$ samples for ACC_{AP} data were found acceptable for unfolding certain attractors. In the second step, the average exponential rate of divergence of initially nearby trajectories in each reconstructed state space was calculated following Rosenstein et al. (1993). In the third and last step, LyE values were quantified as the slope of the first linear growing part (0–75 samples) of the resulting divergence curves. Finally, they were converted to bits per second by multiplying them by the sampling frequency (100 Hz). As a general rule, the larger a LyE estimate, the greater the system's sensitivity to infinitesimal perturbations, or in other words, the lower the local dynamic stability (Roerdink et al., 2006; Donker et al., 2007; Lamoth et al., 2009).

Detrended Fluctuation Analysis

Detrended fluctuation analysis is a modification of the random walk analysis and has been proven suitable for revealing the extent of long-range correlations in biological signals (Peng et al., 1995; Goldberger et al., 2002; Lipsitz, 2002; Harrison and Stergiou, 2015). In brief, DFA fits a power law to the time series' average detrended fluctuations, $F(n)$, across different box sizes (or scales), n, and evaluates the scaling exponent α by determining the slope of the linear regression line of the log-log graph of $F(n)$ vs. n (Damouras et al., 2010). For example, if data are completely uncorrelated (white noise), the scaling exponent $\alpha = 0.5$, while for the opposite extreme (brown noise) of $\alpha = 1.5$, a value at any given moment is strongly correlated to the previous one. In contrast, $\alpha = 1$ reflects $1/f$ (pink) noise and implies that an event at every point approximately depends equally on those from the recent and those from the very distant past (Peng et al., 1995). Referring to Peng et al. (1995) and Lipsitz (2002), this class of fractal-like (self-similar) processes can be supposed to be quite adaptive. It should be interpreted as a broad compromise between fairly "rough" and fairly "smooth" assembled control mechanisms. Of note, to more specifically address the proprioceptive feedback loop, only α values derived from a range of scales associated with the higher frequencies in postural balance signals, that is, between 2 and 10 Hz, were currently examined. Detailed theoretical and technical aspects of this approach are described in Gilfriche et al. (2018).

Code Availability and Statistical Analyses

The SEn and LyE algorithms were taken from PhysioNet (Goldberger et al., 2000), whereas for DFA the source code provided by Gilfriche et al. (2018) was employed. These tools were implemented in Matlab software version R2020b (The MathWorks Inc., Natick, MA, United States). Data analysis, including state space reconstruction through its integrated "Predictive Maintenance Toolbox" was entirely managed with Matlab–unless otherwise specified.

For statistical analyses, data sets were initially transformed by the natural logarithm (after adding a constant of 1) to better approximate normality (Osborne, 2002). A three-way mixed analysis of variance (ANOVA) with "condition" (firm vs. foam) and "visit" (first vs. second) as the within-subject factors and "group" (YO vs. YN) as the between-subjects factor was completed for each dependent variable. In case of significant main or interaction effects, follow-up pairwise comparisons were performed using the Benjamini-Hochberg procedure (false discovery rate) to control for family wise type I error. However, if the effect of "visit" was statistically negligible, mean values were pooled across this factor before *post-hoc* testing. The Alpha level for rejecting the null hypothesis was set at 0.05. In addition, the differences between the values obtained were evaluated by computing Cohen's (1988) effect size f (small: $f > 0.10$, medium: $f > 0.25$, large: $f > 0.40$). For assessing test-retest reliability, a one-way random model of intra-class correlation ($ICC_{1,1}$) was employed (cf. Mancini et al., 2012). Based on the work of Fleiss (1986), particular results were interpreted as poor when $ICC_{1,1} \leq 0.40$, fair when $ICC_{1,1} \leq 0.60$, good when $ICC_{1,1} \leq 0.75$, and excellent when $ICC_{1,1} \leq 1.00$. IBM SPSS version 27.0 (SPSS Inc., Chicago, IL, United States) software was applied for all statistical analyses.

RESULTS

Reliability

Table 2 summarizes the test–retest reliability of ACC_{ML} and ACC_{AP} measures for the YO and their matched YN controls while standing on firm and foam support surfaces. The reliability of the non-linear measures was fair to excellent, except for the α-DFA_{AP} variable in terms of the YN controls. Therefore, the reports on this variable remain a purely descriptive one (see the section below). Despite the occasional only fair reliability in some measures of YN controls, it seems justified to assume that the given experimental field approach investigates the distinct aspects of postural control with sufficient precision.

TABLE 2 | Test–retest reliability, as expressed with the intraclass correlation coefficient$_{(1,1)}$, of postural sway measures.

Variable	Direction	Overweight/Obese		Normal-weight	
		Firm	Foam	Firm	Foam
SEn (bit)	ML	0.74	0.75	0.81	0.57
	AP	0.76	0.86	0.71	0.43
LyE (bit·s^{-1})	ML	0.70	0.79	0.52	0.71
	AP	0.68	0.73	0.54	0.55
α-DFA	ML	0.64	0.75	0.59	0.69
	AP	0.51	0.52	0.40	0.20
RMS (mm·s^{-2})	ML	0.83	0.79	0.64	0.49
	AP	0.85	0.84	0.75	0.59

SEn, sample entropy; LyE, largest Lyapunov exponent; α-DFA, scaling exponent α derived from detrended fluctuation analysis; RMS, root mean square; ML, mediolateral direction; AP, anteriorposterior direction.

Differences

Descriptive, non-log-transformed statistics are shown in **Table 3**, while a summary of ANOVA results is provided in **Table 4**. All significant *post-hoc* comparisons are reported in **Figures 2A–H**.

There was no significant main effect of visit nor any group × visit, condition × visit, or group × condition × visit interaction across the variables considered (all $p > 0.05$). Each outcome measure revealed a significant main effect of condition (all $p < 0.001$). More precisely, SEn_{ML} (YO: $p < 0.001, f = 1.09$; YN: $p < 0.001, f = 1.02$) and SEn_{AP} (YO: $p < 0.001, f = 1.24$; YN: $p = 0.001, f = 0.91$) values were lower for standing on a foam compared to standing on a firm support surface, whereas the LyE_{ML} (YO: $p < 0.001, f = 0.97$; YN: $p < 0.001, f = 1.14$), LyE_{AP} (YO: $p = 0.003, f = 0.87$; YN: $p = 0.039, f = 0.52$), α-DFA_{ML} (YO: $p = 0.031, f = 0.55$; YN: $p < 0.001, f = 1.37$), RMS_{ML} (YO: $p < 0.001, f = 2.13$; YN: $p < 0.001, f = 2.61$), and RMS_{AP} (YO: $p < 0.001, f = 1.84$; YN: $p < 0.001, f = 2.70$) variables exhibited higher values in the foam than in the firm condition.

A significant main effect of group was found for the SEn_{ML}, SEn_{AP}, LyE_{ML}, α-DFA_{ML}, and RMS_{AP} (all $p < 0.05$) but not for the LyE_{AP} and RMS_{ML} variables (all $p > 0.05$). In detail, the YO had smaller SEn_{ML} (firm: $p = 0.030, f = 0.38$; foam: $p = 0.030, f = 0.41$) and SEn_{AP} (firm: $p = 0.059, f = 0.33$; foam: $p = 0.018, f = 0.47$) values as well as increased LyE_{ML} (firm: $p = 0.006, f = 0.53$; foam: $p = 0.017, f = 0.42$), α-DFA_{ML} (firm: $p = 0.004, f = 0.56$; foam: $p = 0.040, f = 0.35$), and RMS_{AP} (firm: $p = 0.049, f = 0.34$; foam: $p = 0.049, f = 0.34$) values compared to their YN controls.

No group × condition interaction was detected throughout (all $p > 0.05$) except for the α-DFA_{ML} variable ($p = 0.030$), where the differences between groups were noticed to be more pronounced in the firm compared to foam condition.

DISCUSSION

In agreement with our primary hypothesis and the loss-of-complexity hypothesis (Lipsitz and Goldberger, 1992), YO display postural control deficits with a more conservative and

TABLE 3 | Descriptive statistics of postural sway measures in mediolateral and anteriorposterior direction for the young overweight/obese (n = 19) and their matched normal-weight controls (n = 19) while standing on firm and foam support surfaces according to both visits.

Variable	Visit	Mediolateral direction				Anteriorposterior direction			
		Overweight/Obese		Normal-weight		Overweight/Obese		Normal-weight	
		Firm	Foam	Firm	Foam	Firm	Foam	Firm	Foam
SEn (bit)	Visit 1	0.920 ± 0.131	0.854 ± 0.130	0.996 ± 0.065	0.933 ± 0.079	0.986 ± 0.126	0.894 ± 0.130	1.040 ± 0.068	0.982 ± 0.078
	Visit 2	0.923 ± 0.137	0.877 ± 0.096	0.992 ± 0.064	0.941 ± 0.069	0.967 ± 0.128	0.903 ± 0.114	1.036 ± 0.082	0.988 ± 0.079
LyE (bit·s^{-1})	Visit 1	0.817 ± 0.144	0.897 ± 0.131	0.695 ± 0.089	0.792 ± 0.123	0.702 ± 0.114	0.771 ± 0.108	0.673 ± 0.106	0.728 ± 0.105
	Visit 2	0.818 ± 0.164	0.871 ± 0.125	0.707 ± 0.094	0.789 ± 0.122	0.722 ± 0.120	0.765 ± 0.144	0.688 ± 0.099	0.717 ± 0.080
α-DFA	Visit 1	1.257 ± 0.135	1.307 ± 0.102	1.129 ± 0.112	1.215 ± 0.097	1.233 ± 0.107	1.313 ± 0.111	1.200 ± 0.108	1.239 ± 0.093
	Visit 2	1.238 ± 0.130	1.268 ± 0.103	1.129 ± 0.106	1.228 ± 0.109	1.240 ± 0.114	1.242 ± 0.157	1.208 ± 0.100	1.207 ± 0.108
RMS (mm·s^{-2})	Visit 1	40.3 ± 16.8	61.1 ± 30.5	34.7 ± 11.6	51.2 ± 14.5	47.4 ± 18.3	65.8 ± 26.1	37.1 ± 8.5	52.6 ± 13.4
	Visit 2	43.2 ± 24.3	56.1 ± 17.1	38.8 ± 11.4	55.4 ± 18.4	50.5 ± 24.4	64.8 ± 20.0	39.5 ± 10.1	53.9 ± 15.5

Data are expressed as mean ± standard deviation values. SEn, sample entropy; LyE, largest Lyapunov exponent; α-DFA, scaling exponent α derived from detrended fluctuation analysis; RMS, root mean square.

TABLE 4 | Summary of inferential ANOVA statistics. F-values, p-values, and effect size f are presented for a three-way mixed analysis of variance.

Variable	Direction	Statistics	Group	Condition	Visit	Group × Condition	Group × Visit	Condition × Visit	Group × Condition × Visit
SEn (bit)	ML	$F_{1,36}$	6.11	39.78	0.81	0.00	0.41	0.95	0.11
		p	0.018	<0.001	0.373	0.996	0.528	0.336	0.739
		f	0.41	1.05	0.15	0.00	0.11	0.16	0.06
	AP	$F_{1,36}$	6.34	42.15	0.03	1.83	0.06	2.15	0.47
		p	0.016	<0.001	0.867	0.184	0.801	0.152	0.497
		f	0.42	1.08	0.03	0.23	0.04	0.24	0.11
LyE (bit·s^{-1})	ML	$F_{1,36}$	9.05	40.39	0.09	0.93	0.40	1.55	0.08
		p	0.005	<0.001	0.765	0.341	0.532	0.221	0.779
		f	0.50	1.06	0.05	0.16	0.11	0.21	0.05
	AP	$F_{1,36}$	1.56	16.15	0.10	0.26	0.01	2.84	0.00
		p	0.220	<0.001	0.749	0.613	0.914	0.100	0.945
		f	0.21	0.67	0.05	0.08	0.02	0.28	0.01
α-DFA	ML	$F_{1,36}$	8.81	32.38	0.72	5.10	1.76	0.06	0.95
		p	0.005	<0.001	0.403	0.030	0.194	0.810	0.337
		f	0.49	0.95	0.14	0.38	0.22	0.04	0.16
	AP	$F_{1,36}$							
		p	not inferentially tested due to poor reliability (see **Table 2**)						
		f							
RMS (mm·s^{-2})	ML	$F_{1,36}$	0.45	197.05	2.33	0.06	1.83	1.36	0.12
		p	0.505	<0.001	0.136	0.806	0.185	0.251	0.734
		f	0.11	2.34	0.25	0.04	0.23	0.19	0.06
	AP	$F_{1,36}$	4.47	169.39	1.80	0.10	0.02	1.09	0.01
		p	0.042	<0.001	0.188	0.748	0.878	0.303	0.942
		f	0.35	2.17	0.22	0.05	0.03	0.17	0.01

SEn, sample entropy; LyE, largest Lyapunov exponent; α-DFA, scaling exponent α derived from detrended fluctuation analysis; RMS, root mean square; ML, mediolateral direction; AP, anteriorposterior direction.

constrained balance strategy in double-legged quiet standing than their YN peers. We were the first to show that this decomplexification of postural fluctuations in YO have already been clearly detectable during firm standing. We expand this finding to standing under proprioceptive manipulation and provide evidence that the regulation of the trunk becomes systematically more repeatable, less fractally organized, and more sensitive to initial perturbations on a foam surface. However, contrary to our secondary hypothesis, the sway differences between YO and YN remained largely independent of the task difficulty (**Figures 2A–H**). Another crucial discovery of the present study was that a single inexpensive IMU enables linear and non-linear trunk excursion analyzes of similar reproducibility to gold-standard tests on a force plate (**Table 2**). Subsequently, this IMU approach featured a high construct validity and proved sensitive to discriminating strategies

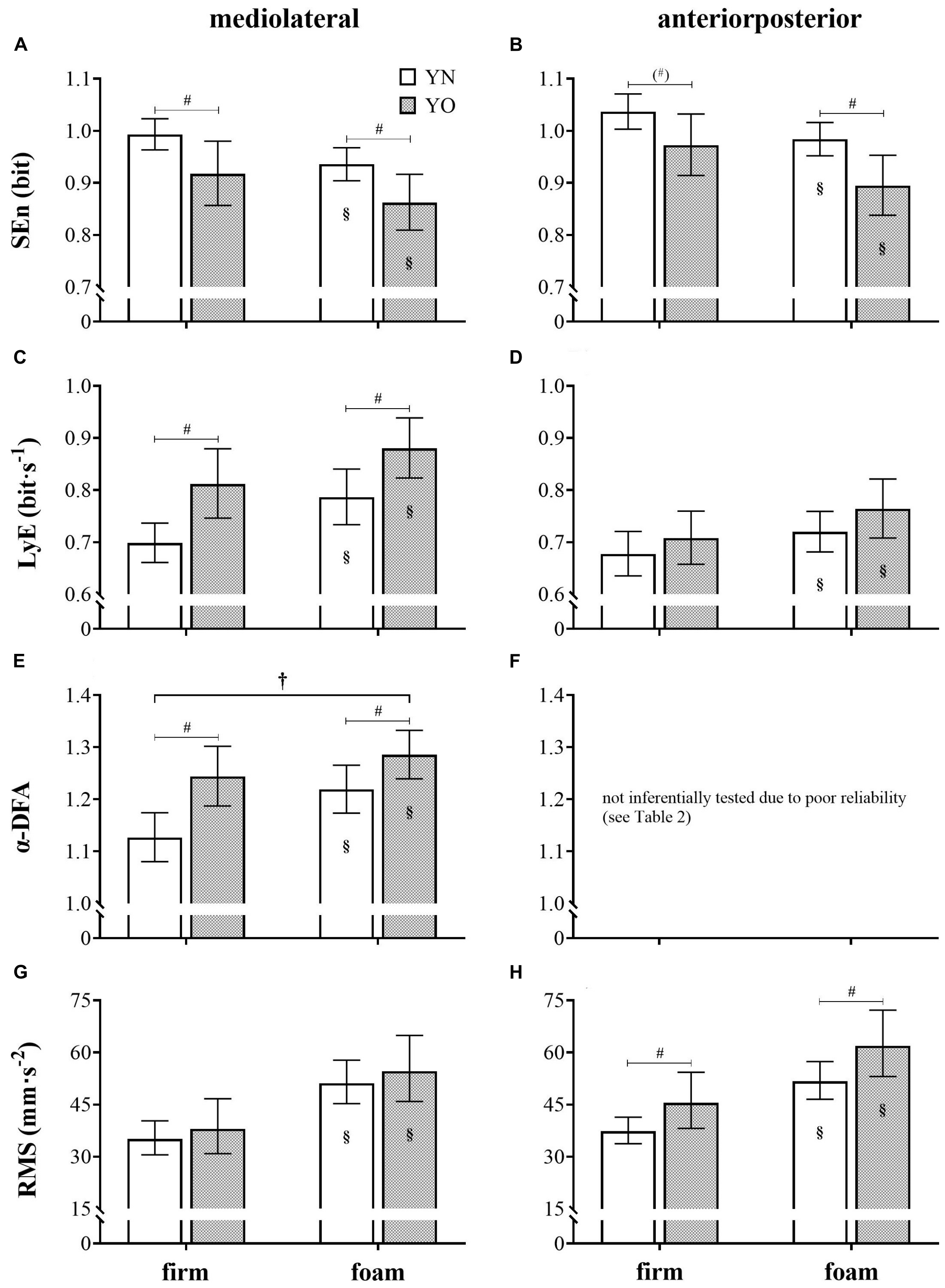

FIGURE 2 | Postural complexity and magnitude of postural fluctuations. Sample entropy (SEn) **(A,B)**, largest Lyapunov exponent (LyE) **(C,D)**, scaling exponent α derived from detrended fluctuation analysis (α-DFA) **(E,F)**, and root mean square (RMS) **(G,H)** values from acceleration time series in mediolateral and anteriorposterior direction for the young overweight/obese (YO) and matched young normal-weight (YN) controls during standing on firm and foam support surfaces pooled across the visit. Bars represent anti-logged means and 95% confidence intervals. †, significant ($p < 0.05$) group × condition interaction; #, significant ($p < 0.05$) difference between groups; ($^{\#}$), nearly significant ($p = 0.059$) difference between groups; §, significantly different ($p < 0.05$) to standing on a firm support surface.

in the underlying organization of postural control between YO and YN of the firm and foam bilateral static stance.

At a meta-level, these main outcomes of reduced balance control caused by a lower complexity in YO are broadly in line with the few previous cross-comparison studies in this cohort (Pau et al., 2012; Villarrasa-Sapina et al., 2016; Fink et al., 2019). However, past reports did not follow a comprehensive approach to reveal the temporally evolving dynamics of postural control by employing three prominent non-linear measures. Furthermore, non-neglectable methodological and sample-based differences (e.g., the number of included data points, different data pre-processing techniques) make it difficult to contextualize our structure-based outcomes in the available literature. Still, to classify our findings within this research framework, the increased magnitudes of fluctuations in the sagittal plane when being overweight or obese are consistent findings of cross-comparison studies (McGraw et al., 2000; Deforche et al., 2009; Pau et al., 2012). Likewise, the lack of between groups RMS disparity in the frontal plane is in line with our expectation of minor task challenge due to the high base of support of the hip-width double-legged stance. However, this appearance of more significant postural control deficits in the AP direction seems to emerge reversed when looking beyond the postural performance at an outcome level. That means, SEn, LyE, and the α-DFA values systematically provided group differences with pronounced decomplexification in the ML direction. These discrepancies in the outcome of magnitude (RMS) and structure-based variables (SEn, LyE, α-DFA) highlight that postural control is embedded in a complex network composed of many interacting heterogeneous constituents and underpins the value of monitoring postural control from the perspective of complexity.

Cumulatively, the used non-linear sway measures during bilateral stance on a firm or foam support surface show less complex balance strategies for YO than YN peers. Such patterns of decomplexification of postural regulation are thought to reduce the adaptability of YO to heterogeneous stressors of the natural environment but bring them into a state of increased postural challenge or even a state comparable to persons with different pathologies. Accordingly, CoP variabilities of lower SEn (Hansen et al., 2017; Lubetzky et al., 2018), higher LyE (Lamoth et al., 2009; Lamoth and van Heuvelen, 2012; Buchecker et al., 2018), and higher α-DFA values (Zhou et al., 2013) have systematically been observed across various cohorts during balance conditions like standing on unstable surfaces, perturbing the optical flow, or desensitizing the somatosensory system. Likewise, decomplexifications in postural control indicate pathologies affecting selective neuromuscular or sensory control systems (see Goldberger, 1996 for review) or age-related deteriorations (Sturnieks et al., 2008). Concerning task difficulty, the foam condition deteriorated postural performance in both groups of the present study. However, contrary to our second hypothesis and previous non-linear observations in a comparable cohort (Pau et al., 2012; Villarrasa-Sapina et al., 2016; Fink et al., 2019), standing under altered sensory condition exhibited no amplification of detrimental balance effects in the YO compared to their YN peers (**Figures 2A–H**). Instead, the contrary, group differences in α-DFA_{ML} values between YO and YN were even more significant in the firm than in the foam condition (**Figure 2E**). We speculate that this task independence reflects that standing on a foam support surface even pushed YN to their postural equilibrium limits and impeded better group discriminations.

Nevertheless, the present findings clearly show that the balance strategy of YO is less complex than the regulation processes of their YN peers. Based on models of optimal movement (Stergiou et al., 2006) or optimal coordination (Hamill et al., 2012) variability, it is strongly indicated to foster the degree of variability–particularly in YO–up a certain level where the balance control system develops a highly complex, chaotic structure. Such a system brings multiple degrees of freedom at different spatio-temporal scales into proper relations so that it flexibly compensates and quickly adapts to internal and external perturbations. At an optimal degree of temporal structuredness (Stergiou et al., 2006) and an optimal coupling variability (Hamill et al., 2012), the control system is said to become globally more stable with an improvement in task performance. From a practical perspective, state-of-the-art training concepts, like the constraints-led approach (Davids et al., 2008) or differential learning (Schöllhorn et al., 2012), aim to encourage movement complexity by prompting self-organization processes through the exploration of different movement solutions. In a recent quantitative meta-analysis (Tassignon et al., 2021), differential learning was more effective than at least traditional training methods. The constraint-led approach can be even more effective by adding specific task constraints that further stimulate exploration of the solution space (Gray, 2020). However, finding the optimal movement complexity and the optimal dose of practice variability for a given task and a certain individual in the time course of learning is still an open question, for which the assessment of the actual level of motion (postural) complexity with a toolbox of non-linear measures provides future relief.

Using a cost- and time-effective procedure in the form of one single IMU and one single 60 s balance test for judging the temporal structure of sway patterns of balance is a strength of the present paper. Intriguingly, the RMS measures and most non-linear tools provided ICC values comparable to the gold-standard force plate CoP evaluations (see Ruhe et al., 2010 for review). Data reproducibility was further confirmed by observing no relevant changes between test sessions means of any main factorial ANOVA model (*a priori* disregarding α-DFA_{AP}). It is reasonable to assume that the reliability can be further improved by using the mean sway patterns of multiple similar double-legged tests within a short break. This may also advance the reproducibility of DFA_{AP} up to an acceptable level. Besides being reliable, highly economical, and practical, IMU signals have frequently been connected to smartphone applications (e.g., Aqueveque et al., 2020). Moving forward, such applications could open new possibilities for enabling real-time feedback in self-organization routines such as Newell's Constraints Led Approaches. Therefore, the shown applicability can be crucial for opening the door to a vast array of new exercise-based therapeutic perspectives in postural control for clinicians and other practitioners.

CONCLUSION

Our findings replicated previous observations showing that YO are more disadvantaged in their coordination and balance ability than normal-weight peers. Our novel approach using a simple IMU by extending traditional magnitude-based methods through implementing the most valued non-linear tools revealed underlying control processes toward a less complex but more simplified balance regulation in this specific cohort. This increased system rigidity with decreased degrees of freedom and increased sensitivity to internal perturbations is unisonously described as a disability in motor development. Since balance/stability is an essential prerequisite of almost all movement skills, we strongly urge to implement postural challenges in the early years to avoid life-long physical inactivity and all the negative consequences.

Likewise, we hope future studies delve even more into the mechanisms underlying the harmful effects of being overweight and its reciprocal impact on physical activity. Besides studies to examine the usefulness of targeted exercises, prospective investigations are urgently required to quantify the long-term effects of improvements in postural control. Consequently, much research remains to be done, yet we are confident of the high potential of balance training to help to alleviate the global epidemic of inactivity and overall mortality.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Ethics Review Board of the University of Salzburg (EK-GZ: 17/2018). Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin.

AUTHOR CONTRIBUTIONS

H-PW, MB, EM, TS, and JB: conceptualization, writing—review and editing, and supervision. H-PW, MB, and TS: methodology. MB and H-PW: investigation, data analysis, and visualization. H-PW, MB, and JB: writing – original draft. All authors read and approved the content of the final manuscript.

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Effects of Physical and Mental Fatigue on Postural Sway and Cortical Activity in Healthy Young Adults

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Physical fatigue (PF) negatively affects postural control, resulting in impaired balance performance in young and older adults. Similar effects on postural control can be observed for mental fatigue (MF) mainly in older adults. Controversial results exist for young adults. There is a void in the literature on the effects of fatigue on balance and cortical activity. Therefore, this study aimed to examine the acute effects of PF and MF on postural sway and cortical activity. Fifteen healthy young adults aged 28 ± 3 years participated in this study. MF and PF protocols comprising of an all-out repeated sit-to-stand task and a computer-based attention network test, respectively, were applied in random order. Pre and post fatigue, cortical activity and postural sway (i.e., center of pressure displacements [CoP_d], velocity [CoP_v], and CoP variability [CV CoP_d, CV CoP_v]) were tested during a challenging bipedal balance board task. Absolute spectral power was calculated for theta (4–7.5 Hz), alpha-2 (10.5–12.5 Hz), beta-1 (13–18 Hz), and beta-2 (18.5–25 Hz) in frontal, central, and parietal regions of interest (ROI) and baseline-normalized. Inference statistics revealed a significant time-by-fatigue interaction for CoP_d ($p = 0.009$, $d = 0.39$, Δ 9.2%) and CoP_v ($p = 0.009$, $d = 0.36$, Δ 9.2%), and a significant main effect of time for CoP variability (CV CoP_d: $p = 0.001$, $d = 0.84$; CV CoP_v: $p = 0.05$, $d = 0.62$). *Post hoc* analyses showed a significant increase in CoP_d ($p = 0.002$, $d = 1.03$) and CoP_v ($p = 0.003$, $d = 1.03$) following PF but not MF. For cortical activity, a significant time-by-fatigue interaction was found for relative alpha-2 power in parietal ($p < 0.001$, $d = 0.06$) areas. *Post hoc* tests indicated larger alpha-2 power increases after PF ($p < 0.001$, $d = 1.69$, Δ 3.9%) compared to MF ($p = 0.001$, $d = 1.03$, Δ 2.5%). In addition, changes in parietal alpha-2 power and measures of postural sway did not correlate significantly, irrespective of the applied fatigue protocol. No significant changes were found for the other frequency bands, irrespective of the fatigue protocol and ROI under investigation. Thus, the applied PF protocol resulted in increased postural sway (CoP_d and CoP_v) and CoP variability accompanied by enhanced alpha-2 power in the parietal ROI while MF led to increased CoP variability and alpha-2 power in our sample of young adults. Potential underlying cortical mechanisms responsible for the greater increase in parietal alpha-2 power after PF were discussed but could not be clearly identified as cause. Therefore, further future research is needed to decipher alternative interpretations.

Keywords: balance, cognitive/muscular fatigue, EEG, theta, alpha-2

INTRODUCTION

Postural balance is an essential prerequisite to successfully perform everyday and sports-related activities (Horak, 2006; Kiers et al., 2013). For decades, it has been speculated that balance is controlled predominantly by spinal and subcortical systems (brain stem, basal ganglia, and cerebellum) (Dietz et al., 1991; Shumway-Cook and Woollacott, 2012). However, evidence from numerous neuroimaging studies supports the hypothesis that the maintenance of posture underlies complex interactions within specific regions of the somatosensory system (Jacobs and Horak, 2007). In fact, the performance of balance exercises with increasing task difficulty resulted in increases in theta band activity of fronto-central areas and decreases in alpha-2 frequency band power of centro-parietal regions using electroencephalography (EEG) (Hülsdünker et al., 2015a; Gebel et al., 2020). Solis-Escalante et al. (2019) and Varghese et al. (2019) reported that the application of postural perturbations resulted in multifocal transient changes in the frequency band power which could be indicative of cortical network activity involved in postural control. These study findings indicate modifications in balance performance are accompanied by changes in cortical activity (for a review see Wittenberg et al., 2017).

While there is evidence that exercise has positive effects on cortical activity during the performance of balance tasks, less is known on potential detrimental effects of physical (PF) and/or mental fatigue (MF). PF has been defined as exercise-induced declines in muscle force due repetitive single-joint (i.e., local fatigue) or multi-joint movement tasks (i.e., general fatigue) resulting in reduced sensory afferents from types Ia and II fibers of muscle spindles (Paillard, 2012). In contrast, MF impairs attentional control and working memory due to the performance after prolonged or strenuous cognitive task performed (e.g., attention network test, continuous performance test, and Stroop-task) of at least 30 min (Holtzer et al., 2011; Lew and Qu, 2014; Wascher et al., 2014; Van Cutsem et al., 2017a; Hachard et al., 2020; Verschueren et al., 2020). Additionally, there is preliminary evidence that PF and MF conditions modulate brain oscillatory activity in the theta, alpha, and beta frequency bands in frontoparietal areas (Bailey et al., 2008; Craig et al., 2012; Wascher et al., 2014; John et al., 2020). In this context, changes in frontal theta activity, presumably originating in the anterior cingulate cortex, have been observed in various cognitive and motor tasks requiring concentration, attention, working memory and performance monitoring (Smith et al., 1999; Slobounov et al., 2009, 2013; Shou et al., 2012; Baumeister et al., 2013; Sipp et al., 2013; Hülsdünker et al., 2015a,b; Gebel et al., 2020). Furthermore, changes in alpha-2 power in central and parietal brain areas involving the somatosensory cortex and sensory association areas seem to indicate task-specific sensory information processing (Leocani et al., 1997; Smith et al., 1999; Shou et al., 2012; Baumeister et al., 2013; Sipp et al., 2013; Gebel et al., 2020), while changes in beta-1 and beta-2 power in fronto-central areas have been associated with movement planning and execution (Leocani et al., 1997; Gwin et al., 2011; Shou et al., 2012; Sipp et al., 2013).

Several studies have shown that PF caused increased postural sway and sway velocity during the performance of balance tasks as well as altered spinal reflexes and increased co-contractions (Springer and Pincivero, 2009; Zech et al., 2012; Bryanton and Bilodeau, 2016; Ritzmann et al., 2016; Sadowska and Krzepota, 2016; Hamacher et al., 2018; Bedo et al., 2020). In a review article on the effects of PF on postural control, Paillard (2012) concluded that local and general PF deteriorate afferent sensory information processing and motor output during the performance of balance tasks which may partly be compensated by the allocation of additional cognitive resources.

Only few studies have examined the effects of MF on balance performance (Deschamps et al., 2013; Lew and Qu, 2014; Hachard et al., 2020; Morris and Christie, 2020; Verschueren et al., 2020). For instance, Deschamps et al. (2013) investigated the impact of a psychomotor vigilance test on static balance in healthy male students aged 22 years. Fatigue-related increases in static balance were found during bipedal stance on a foam surface. More recently, Hachard et al. (2020) evaluated the effects of a 90 min AX-continuous performance task on static balance in healthy young adults with a mean age of 21 years. Following the MF protocol, increased regularity of the center of pressure (CoP) indicated less automated processes of postural control with increased cognitive contributions during bipedal stance. The authors argued that in a state of MF, which led to impaired attentional processing, an increased activation of cognitive resources was necessary to control and monitor balance. Hence, these findings suggest that a fatigue related decline in balance performance might be compensated by the activation of additional cognitive resources (Hachard et al., 2020). In this context, Donker et al. (2007) reported that, in addition to CoP regularity, CoP variability is also well-suited to measure cognitive involvement in postural control. However, there is a paucity of data on the effects of PF and MF on neural correlates of balance performance. The available studies examined the effects of PF or MF on different measures of physical fitness (e.g., counter movement jumping height and gait velocity) and cognitive function (e.g., working memory performance) as well as their neural correlates.

Data on cortical activity provide further insight into the relationship between cortical activity and balance and how this relation is affected by fatigue. Therefore, the objective of this study was to examine the acute effects of MF and PF on measures of postural sway (i.e., CoP displacements [CoP_d], velocity [CoP_v], variability of CoP_d and CoP_v) and cortical activity during the performance of a demanding balance task in healthy young adults. Based on the relevant literature, we expected increases in measures of postural sway following both fatigue protocols (Paillard, 2012; Zech et al., 2012; Sadowska and Krzepota, 2016; Grobe et al., 2017; Hachard et al., 2020). More specifically, due to the specifics of the fatigue protocols, we expected stronger declines in balance performance after PF. Additionally, as cognitive processing (e.g., allocation of attentional resources) is involved in postural control, especially when balance tasks are more challenging, we expected that MF affects balance performance too but to a lesser degree. Furthermore, we hypothesized that PF and MF result in spectral power increases across fronto-parietal areas in the theta, alpha-2, beta-1, and beta-2 frequency band (Bailey et al., 2008; Craig et al., 2012; Wascher et al., 2014; John et al., 2020). More precisely, we expected alpha-2 frequency band power increases to be larger over areas involved in sensory information processing (i.e., parietal) following PF compared to MF due to reduced sensory feedback as a result of PF.

MATERIALS AND METHODS

Participants

With reference to the study of Sadowska and Krzepota (2016) and the reported large effect size ($\eta^2 = 0.23$) of PF on CoP_v, an *a priori* power analysis was performed in G $\times$ Power (Version 3.1.9.2, University of Kiel, Germany) using the *F* test family (ANOVA repeated measures, within between interaction). The analysis revealed that a total sample size of $N = 12$ would be sufficient to find significant large-sized pre-post effects of PF on CoP_v (effect size $f = 0.5$, $\alpha = 0.05$, power = 0.80), with an actual power of 0.88 (critical F-value = 4.96). Accordingly, 15 healthy sport science students (6 females) aged 20–33 years were enrolled in this study. The participants' self-reported physical activity level was assessed using the International Physical Activity Questionnaire-short (IPAQ-SF). **Table 1** shows the characteristics of the participants. Individuals were excluded from study participation if they had any neurological diseases, medications that may influence cortical activity, or lower limbs musculoskeletal injury (e.g., ankle sprain) 6 months prior to the start of the study. Written informed consent was obtained from all participants. The study was approved by the local ethics committee of the University of Potsdam (application no 12/2019) and followed the latest version of the Declaration of Helsinki.

Experimental Procedure

A single group cross-over design was used to examine the effects of PF and MF on postural sway and cortical activity. For this purpose, participants were invited to the biomechanics laboratory for two experimental sessions to test the effects of PF and MF separately. The two sessions were scheduled 1 week apart at the same time of day to account for potential fatigue-related confounds such as muscle soreness, intraday performance variability, and learning effects. The order of the fatigue protocol was randomized. Every test session started with the measurements of anthropometrics followed by EEG preparations. The electrode cap was fitted to the participant's head and the gel-electrodes were prepared. The experimental procedure continued with a standardized and short (3 min) familiarization period to introduce the multi-directional balance board which was used for balance testing. More information on the specifics of the balance task can be found in the section "Balance task." After the familiarization period, baseline EEG recording was performed for 3 min during quiet bipedal stand with eyes opened on stable surface. Baseline tests were realized at the beginning of both experimental sessions. Thereafter, participants performed pre-tests in unfatigued condition with 5 60-s trials on the multi-directional balance board. Subsequently, the participants completed the respective fatigue protocol. Immediately after the fatigue protocol, post-tests were scheduled. Time between the termination of the fatigue protocol and the start of the post-tests was approximately 30 s. The post-tests followed the same procedure as described above, i.e., 5 consecutive 60-s trials of the balance task. During all trials, measures of postural sway and EEG were synchronously recorded.

Physical Fatigue Protocol

To induce PF, a repeated sit-to-stand protocol was selected that resembled an everyday activity (Helbostad et al., 2007; Bryanton and Bilodeau, 2016). Due to the applied fatigue protocol together with the young age of the study participants, weighted vests with a load corresponding to 30% of the individuals' body mass were used to induce PF. During the PF task, participants kept their arms crossed in front of their chest, the back was erect, and the knees were in a 90° at the starting position and fully extended during erect stance. Participants were asked to stand up and sit down at a self-selected pace until task failure. A metronome was not used during the performance of the PF protocol to prevent task failure due to imposed pacing and not PF. The fatigue

TABLE 1 | Participants' characteristics.

	Total (*N* = 15)	Male (*n* = 9)	Female (*n* = 6)
	***M* (*SD*)**	***M* (*SD*)**	***M* (*SD*)**
Age (years)	28.8 (3.4)	29.7 (2.8)	27.5 (4.0)
Body height (cm)	173.3 (9.1)	178.5 (4.6)	165.4 (8.9)
Body mass (kg)	70.0 (9.8)	74.3 (8.4)	63.6 (8.9)
	Low	**Medium**	**High**
Physical activity level (IPAQ-SF)	0	4 (2f/2m)	11 (4f/7m)

IPAQ-SF, International Physical Activity Questionnaire Short Form; f, female; m, male.

protocol was completed if participants were unable to perform the sit-to-stand task anymore. Time until failure was recorded.

Mental Fatigue Protocol

To induce MF, participants completed a software-based attention network test with a total protocol duration of 30 min (Fan et al., 2002). According to Van Cutsem et al. (2017a), MF protocols should at least last 30 min to make sure that participants are actually fatigued. The attention network test is a demanding attentional task, which combines cued visual reaction time (RT) tasks and a flanker task to assess three attentional network components (i.e., alerting, orienting, and executive control). The attention network test protocol used in this study consisted of 480 trials divided in five experimental blocks with 96 trials (4 cue conditions * 2 target locations * 2 target directions * 3 flanker conditions * 2 repetitions) each and were preceded by 24-trial training block. During the MF protocol, participants were in seated position in front of a screen and were asked to quickly and accurately determine the pointing direction of an arrow on the screen. The left arrow key on a keyboard should be pressed with the left index finger for leftward pointing arrows and the right arrow key with the right index finger for rightward pointing arrows. The arrow could appear above or below a fixation cross and might or might not be accompanied by different cue conditions or flanker stimuli. Flankers consisted of neutral dashes (e.g., -- → --) or a sequence of congruent (e.g., →→→→→) or incongruent arrows (e.g., ←←→←←). The four cue conditions were (i) no cue, (ii) center cue, (iii) double cue (no directed spatial information) and (iv) spatial cue (Fan et al., 2002). Subtractions of performance (error rate, RTs) between the different stimulus conditions reflect specific attentional networks and were performed according to Holtzer et al. (2011): no cue – center cue for alerting, center cue – spatial cue for orienting and incongruent vs. congruent flanker trials for executive attention. The mean error rates and reaction times across all blocks were taken to assess performance during the MF protocol. Additionally, mean attention network scores (i.e., alerting, orienting, and executive) were calculated for each of the five blocks and analyzed to identify a potential fatiguing effect on specific attentional networks (Holtzer et al., 2011).

Subjective Level of Physical and Mental Fatigue

In order to evaluate the subjective levels of PF/MF, participants were asked to rate their perceived levels of fatigue on a visual analogue scale (VAS) from 0 cm (not physically/mentally fatigued at all) to 10 cm (extremely physically/mentally fatigued) as previously reported by Van Cutsem et al. (2017b) and Verschueren et al. (2020). Subjective levels of fatigue were assessed immediately before and after the respective fatigue protocol.

Balance Task

The balance tests required participants to stand as still as possible in bipedal and barefooted stance for 60 s on a commercially available multi-directional balance board (Wobblesmart©, Artzt GmbH, Dornburg, Germany). Test trials were started with the balance board in quiet and horizontal position. While performing the balance task, participants were instructed to place their hands akimbo and to fix a cross on a nearby wall (3 m distance) with their eyes. Participants were kindly instructed to avoid tilting movements as well as ground contacts of the board. In other words, participants' task was to keep the board in horizontal position (Gebel et al., 2019). In brief, the pivot of the balance board has 6 difficulty levels. While level 1 is characterized by the largest base of support, level 6 has the smallest. During all tests, level 4 was selected. Two sensor mats (Pedar©, novel GmbH, Posturo S2094, novel GmbH, Munich, Germany) were placed on top of the balance board to measure CoP_d and CoP_v at a sampling frequency of 40 Hz. Postural data were analyzed using the manufacturer software (Posturo 32 Expert software, version 25.3.6, novel GmbH, Munich, Germany). Biomechanical and neurophysiological data were synchronized by sending a continuous 5 V signal from the Pedar© system (Posturo Sync Box, novel GmbH, München, Germany) to the EEG system from the start to the end of each test trial. For additional analyses of CoP variability, the respective coefficients of variation (CV) were calculated for CoP_d and CoP_v.

Electroencephalography Recordings and Analysis

Cortical activity was continuously recorded during each balance test trial. Therefore, a mobile EEG system (eego™ sports, Advanced Neuro Technology B.V., Enschede, Netherlands) with 64 Ag/AgCl passive gel-electrodes implemented in an elastic cap (Waveguard classic, Advanced Neuro Technology B.V., Enschede, Netherlands) was used. Electrodes inside the cap were positioned in conformance with the extended 10–20 system. All channels were re-referenced to the CPz electrode. To obtain a high signal-to-noise ratio, electrode impedances were kept below 5 kΩ. EEG signals were amplified, digitized with a 24-bit analog-to-digital converter (eego™, Advanced Neuro Technology B.V., Enschede, Netherlands) and recorded with the eego™ software (Version 1.2.1, Advanced Neuro Technology B.V. Enschede, Netherlands) at a sampling frequency at 1,024 Hz. Offline EEG data processing was performed using the EEGLAB 13.5.4b toolbox (Delorme and Makeig, 2004) implemented in MATLAB (MathWorks Inc., Natick, MA, United States). Sinusoidal line noise was removed by means of the CleanLine plugin (Mullen, 2012). EEG signals were band pass filtered with a finite impulse response filter ranging from 3 to 50 Hz and finally down-sampled to 256 Hz. Channels contaminated by high-frequency noise, electrode movement, and non-stereotypical electromyographic activity were manually removed. EEG data were then re-referenced to common average. Continuous data were visually inspected, and the identified non-stereotypical artifacts were removed from the data set. An adaptive mixture independent component analysis (Palmer et al., 2011) was performed on the remaining data to identify and remove independent components representing stereotypical artifacts like electro-oculographic (i.e., eye blinks) sources, muscle electromyographic activities (Onton and Makeig, 2006). For frequency specific analyses, EEG data were merged for all five trials recorded before (pre) and after (post) the respective fatigue protocol. This was done for each fatigue protocol and participant separately. According to previous studies (Slobounov et al., 2008, 2013; Hülsdünker et al., 2015a,b; Edwards et al., 2018; Gebel et al., 2020), three regions of interest (ROI) were built at frontal (F3, F1, Fz, F2, and F4), central (C3, C1, Cz, C2, and C4), and parietal (P3, P1, Pz, P2, and P4) areas of the cortex associated with processing attention (Shou et al., 2012; Baumeister et al., 2013; Hülsdünker et al., 2015a,b; Gebel et al., 2020), motor planning and execution (Leocani et al., 1997; Gwin et al., 2011; Shou et al., 2012; Sipp et al., 2013), and sensory information (Leocani et al., 1997; Shou et al., 2012; Baumeister et al., 2013; Gebel et al., 2020). Absolute spectral power was calculated for theta (4–7.5 Hz), alpha-2 (10.5–12.5 Hz), beta-1 (13–18 Hz), and beta-2 (18.5–25 Hz) in the respective ROIs using a fast Fourier transformation with a spectral resolution of 1 Hz and a 10% Hanning window. For further analyses, spectral power values were then normalized to a baseline recording taken prior to the tests during unfatigued bipedal standing on stable surface.

Statistical Analyses

Data are presented as mean and standard deviation. Distribution of the data for normality was checked by the Shapiro–Wilk test. To analyze fatigue-related pre-post differences in measures of balance and fatigue, five separate two-way repeated measures analysis of variance (rmANOVA) were performed for CoP_v, total CoP_d, CV CoP_v, CV CoP_d and VAS scores. Moreover, three rmANOVAs were calculated to evaluate attention network test performance (i.e., error rates, reaction times, and attention network scores) across all participants and blocks. Finally, twelve two-way rmANOVAs were calculated to examine the potential effects of fatigue on frequency-specific cortical activity in the four frequency bands within the three predefined ROIs (frontal, central, and parietal). If significant time-by-fatigue interactions were found, *post hoc* Holm–Bonferroni adjusted paired *t*-tests were computed. In addition, Pearson correlation coefficients were calculated between measures of postural sway and spectral power for frequencies with significant time-by-fatigue interactions. All effect sizes are presented as Cohens *d*. If necessary, effect estimates (η^2) were converted accordingly. As proposed by Cohen (1988), an effect was considered small with an effect size of $d \geq 0.2$, medium $d \geq 0.5$ and large $d \geq 0.8$. The statistical analyses were calculated using the JASP statistical software (Version 0.14.1.0; JASP Team, 2020).

RESULTS

Subjective Level of Physical/Mental Fatigue

Averaged VAS scores are shown in **Figure 1**. A significant main effect for time was observed for the VAS scores [$F_{(1,14)} = 31.431$, $p < 0.001$, $d = 0.91$]. No main effect for fatigue [$F_{(1,14)} = 0.943$, $p = 0.35$] and no interaction effect for time and fatigue [$F_{(1,14)} = 1.934$, $p = 0.186$] was found, indicating comparable changes in subjective fatigue after the PF and the MF protocol.

Physical and Mental Fatigue Protocol

During the PF protocol, the time until exhaustion was clocked for each participant individually. The average time to exhaustion was 10 min (±5.5) with a range from 2 to 21 min. Regarding the MF protocol, the average error rates [$F_{(2.2,10.1)} = 1.069$, $p = 0.361$] and reaction times [$F_{(3.9,5608.7)} = 1.805$, $p = 0.126$] across all participants between blocks showed no significant differences (**Table 2**). Analyses of the mean attention network scores (i.e., alerting, orienting, and executive) across blocks revealed significant main effect for network [$F_{(2,28)} = 18.053$, $p < 0.001$, $d = 2.27$]. Scores for the executive attention network were significantly higher than for the alerting [$t_{(74)} = 6.59$, $p < 0.001$, $d = 1.70$] and orienting network [$t_{(74)} = 4.01$, $p = 0.003$, $d = 1.04$] (**Figure 2**). However, no main effect for block [$F_{(1.4,8828.7)} = 1.010$, $p = 0.356$] and no interaction effect for block and network [$F_{(2.7,8345.1)} = 0.957$, $p = 0.416$] was found. Thus, while an increase in subjective fatigue was reported in the VAS, behavioral performance in the attention network test did not deteriorate significantly over time.

Balance Performance

The two-way rmANOVA revealed a significant main effect for time on CoP_v [$F_{(1,14)} = 6.789$, $p = 0.021$, $d = 0.3$]. Moreover, a significant small-sized time-by-fatigue interaction was indicated [$F_{(1,14)} = 9.062$, $p = 0.009$, $d = 0.36$]. *Post hoc* tests showed a significant increase of sway velocity following PF [$t_{(29)} = 3.98$, $p = 0.003$, $d = 1.03$, Δ 9.2%] while sway velocity remained on pre-fatigue level after MF [$t_{(29)} = 0.4$, $p = 1.00$, Δ −1.0%] (**Figure 3A**). The statistical analyses for CoP_d yielded a significant main effect for fatigue [$F_{(1,14)} = 8.677$, $p = 0.011$, $d = 0.55$] and time [$F_{(1,14)} = 6.683$, $p = 0.022$, $d = 0.28$]. Additionally, the analyses revealed a small-sized time-by-fatigue interaction [$F_{(1,14)} = 9.197$, $p = 0.009$, $d = 0.39$] for postural sway. *Post hoc* analyses showed a significant increase of postural sway following the PF [$t_{(29)} = 3.98$, $p = 0.002$, $d = 1.03$, Δ 9.2%] but not the MF protocol [$t_{(29)} = 0.62$, $p = 0.97$, Δ −1.5%] (**Figure 3B**). In terms of CoP variability, the analyses revealed a significant large-sized main effect of time for CV CoP_d [$F_{(1,14)} = 16.342$, $p = 0.001$, $d = 0.84$] (**Figure 4A**) as well as a significant medium-sized main effect of time for CV CoP_v [$F_{(1,14)} = 4.617$, $p = 0.05$, $d = 0.62$] (**Figure 4B**). No main effects of fatigue or interactions were found.

Cortical Activity

The two-way rmANOVAs with factors time and fatigue indicated a significant main effect of time for relative theta power in the central [$F_{(1,14)} = 8.648$, $p = 0.011$, $d = 0.09$] and parietal ROI [$F_{(1,14)} = 6.646$, $p = 0.022$, $d = 0.09$], for relative alpha-2 power in the frontal [$F_{(1,14)} = 27.812$, $p < 0.001$, $d = 0.19$], central [$F_{(1,14)} = 28.144$, $p < 0.001$, $d = 0.14$] and parietal ROI [$F_{(1,14)} = 35.879$, $p < 0.001$, $d = 0.26$], for relative beta-1 power in the frontal [$F_{(1,14)} = 14.005$, $p = 0.002$, $d = 0.09$], central [$F_{(1,14)} = 20.543$, $p < 0.001$, $d = 0.11$] and parietal ROI [$F_{(1,14)} = 17.785$, $p < 0.001$, $d = 0.13$], and for relative beta-2 power in the parietal ROI [$F_{(1,14)} = 5.207$,

TABLE 2 | Error rates and reaction times for the different blocks of the attention network test presented as mean with standard deviation across all participants.

	Block 1	Block 2	Block 3	Block 4	Block 5
	M (SD)	***M (SD)***	***M (SD)***	***M (SD)***	***M (SD)***
Error rates (%)	3.7 (3.4)	2.4 (2.1)	2.8 (1.6)	2.9 (1.8)	2.1 (1.9)
Reaction time (ms)	582.5 (166.9)	577.5 (158.0)	577.5 (161.0)	570.5 (178.9)	569.7 (161.4)

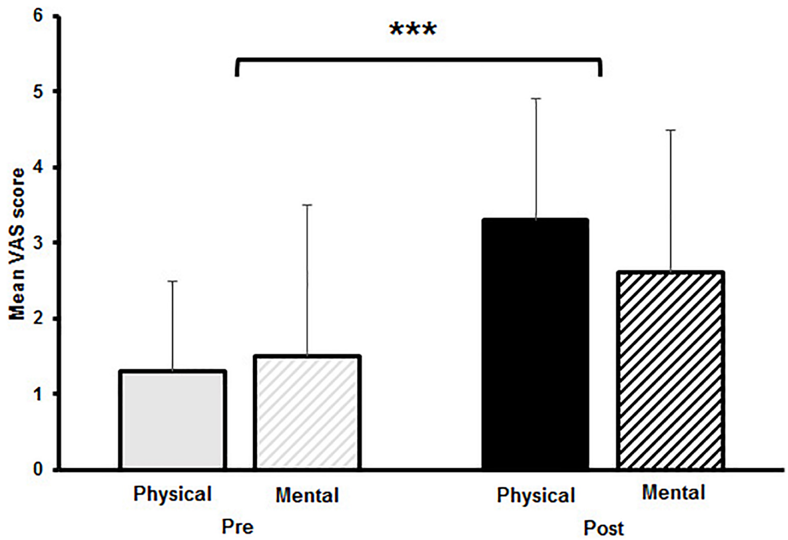

FIGURE 1 | Mean visual analogue scale (VAS) scores with standard deviation of subjective levels of fatigue pre and post to physical and mental fatigue. The significant main effect for time is indicated by the bracket with asterisk; ***$p < 0.001$.

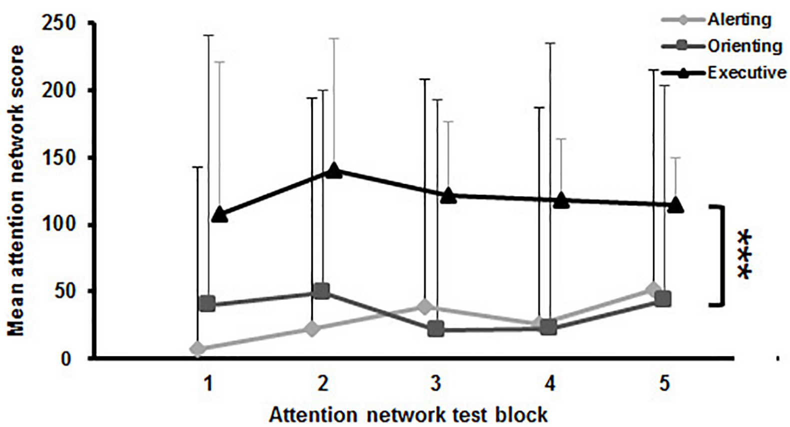

FIGURE 2 | Mean attention network scores with standard deviation of the attention network test across blocks. The significant main effect for network is indicated by the bracket with asterisk; ***$p < 0.001$.

$p = 0.039$, $d = 0.11$] (**Tables 3–6**). Moreover, a significant time-by-fatigue interaction [$F_{(1,14)} = 7.149$, $p = 0.018$, $d = 0.06$] was found for the parietal ROI. Applied *post hoc t*-tests indicated significant increases in relative alpha-2 power (**Figure 5**) during balance task performance after both fatigue protocols [physical: $t_{(29)} = 6.54$, $p < 0.001$, $d = 1.69$; mental: $t_{(29)} = 3.98$, $p = 0.003$, $d = 1.03$]. The relative power increase was more pronounced after the PF (Δ 3.9%) than the MF protocol (Δ 2.5%). Further, we checked whether the interaction effect was confounded by a potential retest effect. The dependent t-test between pre-test values of parietal alpha-2 indicated no significant difference [$t_{(29)} = -0.174$, $p = 0.864$]. Calculation of the intraclass correlation coefficient (ICC; 2-way mixed model, single measurement) according to Koo and Li (2016) also yielded good test-retest reliability with an ICC of 0.82, 95% confidence interval (CI) [0.54, 0.94] for relative alpha-2 frequency band

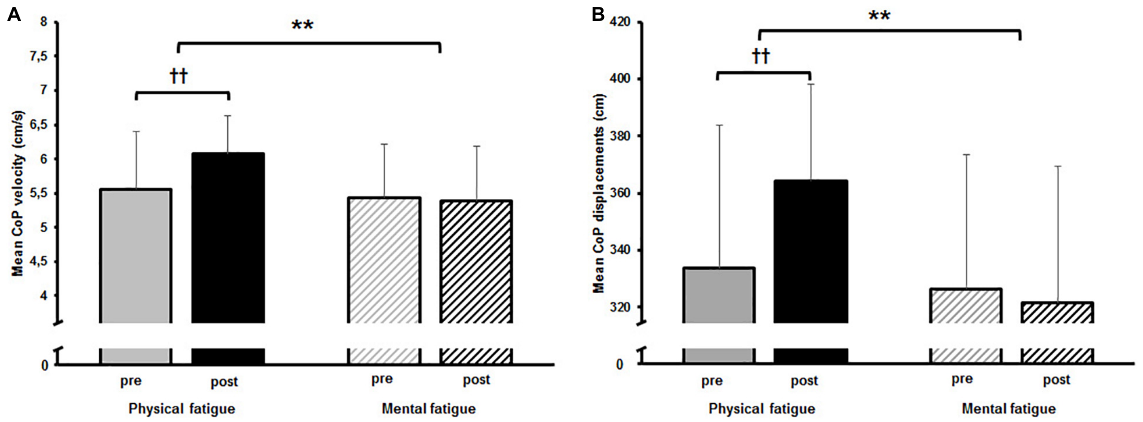

FIGURE 3 | (A) Mean center of pressure (CoP) velocity and **(B)** mean CoP displacement pre and post to both fatigue protocols as well as the results of the *post hoc* comparisons. The significant interaction effect is indicated by; $^{**}p < 0.01$. *Post hoc* results are indicated by; $^{††}p < 0.01$.

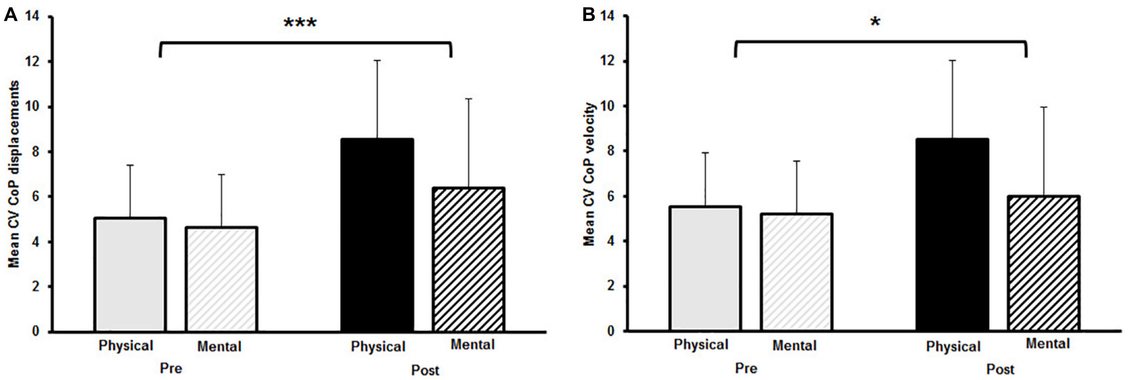

FIGURE 4 | (A) Mean coefficient of variation (CV) for center of pressure (CoP) displacement and **(B)** mean CV CoP velocity pre and post to physical and mental fatigue. The significant main effect for time is indicated by the bracket with asterisk; $^{*}p < 0.05$, $^{***}p < 0.001$.

power. Additionally, the analyses yielded a tendency toward significance for a time-by-fatigue interaction [$F_{(1,14)} = 4.260$, $p = 0.058$] within the central ROI for relative alpha-2 frequency power. No main effects for fatigue or other interactions were found (**Tables 3–6**).

Finally, Pearson correlation coefficients were calculated between pre/post changes (deltas) for measures of postural sway (i.e., CoP_d, CoP_v, CV CoP_d, and CV CoP_v) and relative alpha-2 power of the parietal ROI. No statistically significant associations were found between balance performance and cortical activity, irrespective of the fatigue protocol. The respective correlation coefficients for PF between alpha-2 power and measures of postural sway were $r_{(13)} = 0.191$, ($p = 0.496$, CI −0.37, 0.64) for CoP_d and CoP_v, $r_{(13)} = 0.004$ ($p = 0.989$, CI −0.51, 0.51) for CV CoP_d, and $r_{(13)} = -0.152$ ($p = 0.588$, CI −0.62, 0.39) for CV CoP_v. Regarding MF, the correlation analyses for alpha-2 power with both measures of postural sway resulted in $r_{(13)} = -0.267$ ($p = 0.335$, CI −0.69, 0.28) for CoP_d, $r_{(13)} = -0.351$ ($p = 0.200$, CI −0.73, 0.20) for CoP_v, $r_{(13)} = -0.134$ ($p = 0.633$, CI −0.60, 0.41) for CV CoP_d, and $r_{(13)} = -0.403$ ($p = 0.136$, CI −0.76, 0.14) for CV CoP_v.

DISCUSSION

This study is the first to compare the effects of a PF and MF on cortical activity in the theta, alpha, and beta frequency bands as well as on postural sway while performing a challenging balance task in healthy young adults. The main findings of this study were that only the PF protocol affected postural sway (CoP displacements) and sway velocity while sway variability and cortical activity were affected by both, mental and physical fatigue. In terms of postural sway and sway velocity, CoP_d and CoP_v increased significantly only after the physical fatigue protocol using repeated sit-to-stand tasks to failure. Both fatigue protocols had an impact on variability of postural sway and sway velocity (i.e., CV CoP_d and CV CoP_v). No statistically significant time-by-fatigue interaction was observed for relative theta, beta-1, and beta-2 frequency band power in the three ROIs (i.e., frontal, central, and parietal). However, the relative alpha-2 frequency band power increased significantly in the parietal ROI after both fatigue protocols with a significantly larger increase after the PF protocol. This could reflect, amongst others, deteriorated sensory

TABLE 3 | Averaged theta frequency power relative to baseline values pre and post to PF and MF within the respective ROIs as well as rmANOVA results and *post hoc* tests.

	Relative theta power		Change (%)	RmANOVA			*Post hoc*	
	Pre	Post		Time	Fatigue	Time*fatigue	*p*-value	Effect size (*d*)
	M (*SD*)	*M* (*SD*)						
Frontal								
Physical	101.5 (13.2)	101.5 (12.8)	0.0	*p* = 0.228	*p* = 0.789	*p* = 0.184	–	–
Mental	100.7 (12.7)	101.8 (13.9)	1.1				–	–
Central								
Physical	101.0 (11.2)	101.8 (11.0)	0.8	***p* = 0.011**	*p* = 0.782	*p* = 0.579	–	–
Mental	101.4 (14.0)	102.6 (14.6)	1.2				–	–
Parietal								
Physical	100.4 (11.8)	101.6 (11.9)	1.2	***p* = 0.022**	*p* = 0.656	*p* = 0.685	–	–
Mental	99.9 (11.2)	100.9 (12.6)	1.0				–	–

Data are presented as M with SD in % from baseline. Bold p-values indicate statistical significancy.
Post hoc tests were computed if the omnibus test (i.e., time-by-fatigue interaction) turned out to be significant.

TABLE 4 | Averaged alpha-2 frequency power relative to baseline values pre and post to PF and MF within the respective ROIs as well as rmANOVA results and *post hoc* tests.

	Relative alpha-2 power		Change (%)	RmANOVA			*Post hoc*	
	Pre	Post		Time	Fatigue	Time*fatigue	*p*-value	Effect size (*d*)
	M (*SD*)	*M* (*SD*)						
Frontal								
Physical	97.2 (17.1)	100.2 (13.0)	3.0	***p* < 0.001**	*p* = 0.475	*p* = 0.379	–	–
Mental	96.1 (15.6)	98.5 (16.7)	2.4				–	–
Central								
Physical	96.6 (13.5)	99.6 (13.9)	3.0	***p* < 0.001**	*p* = 0.746	*p* = 0.058	**–**	–
Mental	96.0 (18.7)	100.8 (18.8)	4.8				–	–
Parietal								
Physical	95.5 (10.6)	99.4 (11.3)	3.9	***p* < 0.001**	*p* = 0.512	***p* = 0.018**	**<0.001**	1.69
Mental	95.3 (12.8)	97.7 (14.1)	2.5				**0.003**	1.03

Data are presented as M with SD in % from baseline. Bold p-values indicate statistical significancy.
Post hoc tests were computed if the omnibus test (i.e., time-by-fatigue interaction) turned out to be significant.

information processing related to impaired balance performance caused by PF.

Effects of Fatigue on Balance Performance

The participants reported to have reached a state of fatigue after both, the PF and MF protocol, as indicated by the significant pre/post increase in the VAS scores. Additionally, analyses of the attentional network scores of the attention network test indicated a higher attentional load within the executive attention network without an observable fatiguing effect. Even though participants stated to be physically and mentally fatigued, the observed effects on postural sway diverged from these results. In accordance with our first hypothesis, we observed increased postural sway (CoP_d), sway velocity (CoP_v), and sway variability (CV CoP_d and CV CoP_v) during quite bipedal standing on a wobble board after performing a modified sit-to-stand task till exhaustion in healthy young adults. These results indicate the negative effects of PF on balance performance, although Santos et al. (2019) point out that fatigue-related changes after multi joint exercise are the result of combined physiological and mental fatiguing effects.

Nonetheless, our results are in line with previous research, which reported fatigue-related impairments for balance performance and, thus, postural control resulting in significant increases in postural sway (Sadowska and Krzepota, 2016;

TABLE 5 | Averaged beta-1 frequency power relative to baseline values pre and post to PF and MF within the respective ROIs as well as rmANOVA results and *post hoc* tests.

	Relative beta-1 power		Change (%)	RmANOVA			*Post hoc*	
	Pre	Post		Time	Fatigue	Time*fatigue	*p*-value	Effect size (*d*)
	M (*SD*)	*M* (*SD*)						
Frontal								
Physical	101.0 (12.8)	101.9 (12.6)	0.9	***p* = 0.002**	*p* = 0.810	*p* = 0.591	–	–
Mental	100.3 (14.3)	101.8 (15.8)	1.5				–	–
Central								
Physical	99.9 (12.1)	101.5 (12.1)	1.6	***p* < 0.001**	*p* = 0.807	*p* = 0.881	–	–
Mental	100.6 (17.4)	102.1 (17.4)	1.5				–	–
Parietal								
Physical	99.2 (11.4)	101.2 (11.1)	2.0	***p* < 0.001**	*p* = 0.824	*p* = 0.246	–	–
Mental	99.4 (12.0)	100.4 (13.5)	1.0				–	–

Data are presented as M with SD in % from baseline. Bold p-values indicate statistical significancy.
Post hoc tests were computed if the omnibus test (i.e., time-by-fatigue interaction) turned out to be significant.

TABLE 6 | Averaged beta-2 frequency power relative to baseline values pre and post to PF and MF within the respective ROIs as well as rmANOVA results and *post hoc* tests.

	Relative beta-2 power		Change (%)	RmANOVA			*Post hoc*	
	Pre	Post		Time	Fatigue	Time*fatigue	*p*-value	Effect size (*d*)
	M (*SD*)	*M* (*SD*)						
Frontal								
Physical	102.0 (12.6)	101.3 (12.1)	–0.7	*p* = 0.437	*p* = 0.703	*p* = 0.194	–	–
Mental	100.6 (13.0)	101.8 (13.8)	1.2				–	–
Central								
Physical	99.2 (9.9)	99.9 (9.8)	0.7	*p* = 0.09	*p* = 0.743	*p* = 0.653	–	–
Mental	99.8 (16.2)	100.9 (14.3)	1.1				–	–
Parietal								
Physical	100.0 (9.4)	101.2 (8.6)	1.2	***p* = 0.039**	*p* = 0.601	*p* = 0.843	–	–
Mental	99.5 (9.4)	100.4 (10.6)	0.9				–	–

Data are presented as M with SD in % from baseline. Bold p-values indicate statistical significancy.
Post hoc tests were computed if the omnibus test (i.e., time-by-fatigue interaction) turned out to be significant.

Penedo et al., 2021), sway velocity (Zech et al., 2012), sway area (Bedo et al., 2020; Moon et al., 2020; Penedo et al., 2021), as well as decreases of the stability index (Cooper et al., 2020) during stable and unstable bipedal and unipedal stance in healthy young adults. For instance, Zech et al. (2012) investigated the effects of a localized fatigue versus a general fatigue protocol on static and dynamic balance in male handball athletes. The authors reported significant increases in sway velocity during single leg stance after both fatigue protocols while no changes were observed in star excursion balance test performance. Accordingly, they assumed that there might exist different sensorimotor control mechanisms within the postural control system responsible for static and dynamic balance which are affected differently by the applied fatigue protocols. Moreover, Penedo et al. (2021) examined the effects of local PF on postural sway and lower limb muscle activation in healthy young adults. While postural sway increased, no changes in muscle activation were observed after PF. The authors suggested that impairments in postural control emerge from deteriorated proprioception as well as from changes within the peripheral and central systems (Penedo et al., 2021).

In this context, Gandevia (2001) and Paillard (2012) described in their extensive literature reviews potential spinal (e.g., altered sensory input) and supraspinal factors (e.g., changes in cortical excitability and inhibitory processes) contributing to the fatigue-related declines in afferent sensory information processing and motor output of postural control. The potential role of these

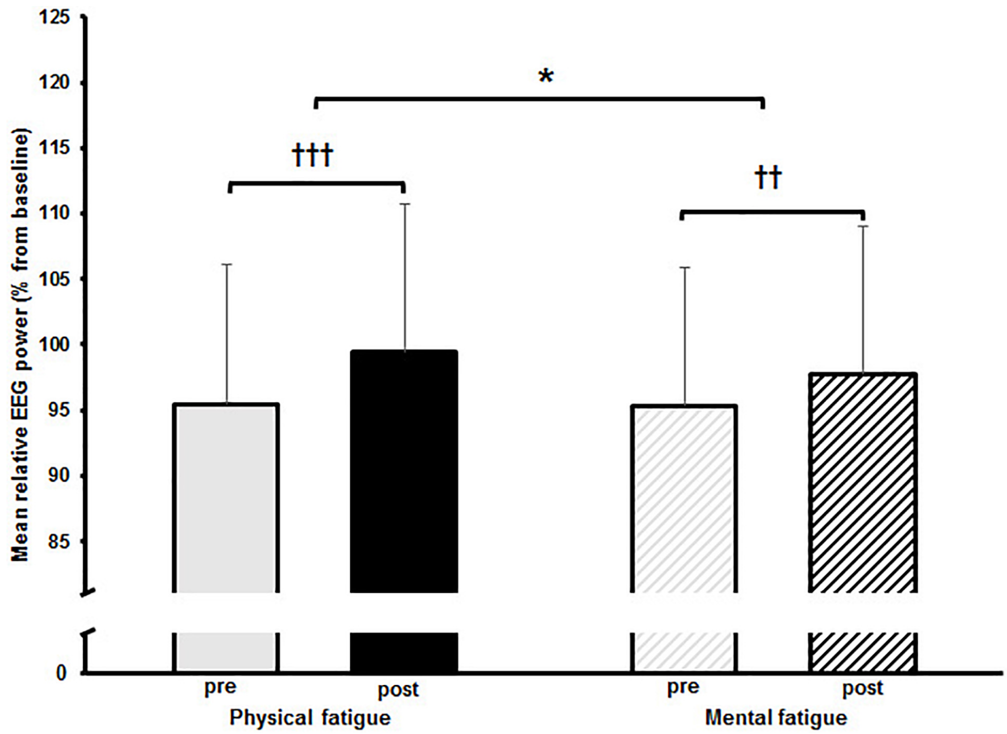

FIGURE 5 | Mean relative alpha-2 frequency band power with standard deviation pre and post to physical and mental fatigue calculated from the electroencephalographic (EEG) electrodes located within the parietal region of interest. The significant interaction effect is indicated by; *$p < 0.05$. *Post hoc* results are indicated by; ††$p < 0.01$, †††$p < 0.001$.

factors is discussed together with the changes in cortical activity related to balance performance decreases.

With respect to the effects of MF on postural control, we did not observe significant increases in postural sway (CoP_d) and sway velocity (CoP_v). At first glance, these results seem to be contradicting previously reported findings. Morris and Christie (2020), for instance, found increased sway velocity during a bipedal reactive balance task in young women after performing a psychomotor vigilance task for 20 min. Similarly, Deschamps et al. (2013) showed significantly increased body sway during bipedal stance on a foam surface after inducing MF. However, and despite the absence of a decrease in balance performance, we observed increases in sway variability (CV CoP_d and CV CoP_v) after the MF protocol suggesting that postural control was influenced by mental fatigue. These results are consistent with findings by Hachard et al. (2020) who investigated the effect of MF on static balance in healthy young adults. After performing a 90 min continuous performance task, the authors found no changes in sway velocity and sway area whereas dynamical features of postural sway variations (i.e., CoP sample entropy) decreased during quiet bipedal stance. Decreases in sample entropy were interpreted as increased cognitive contributions to postural control to compensate for the impact of fatigue and to maintain balance performance. Moreover, Noé et al. (2021) reported that the completion of the same prolonged demanding cognitive task (i.e., continuous performance task) induces a strong heterogeneity in subjects' responses, which affect the individual's postural control system differently according to the sensory context. Thus, it seems that the effects of MF on postural control are initially not reflected by crude CoP measures such as displacement or velocity but by CoP variability or non-linear measures like sample entropy. In addition, the individual's responses to the MF protocol and consequently its different effects on postural control might have confounded our results regarding postural sway and sway velocity. The short duration of the attention network test protocol might also have a confounding effect, but this is discussed in the limitation section.

Although we did not observe a clear deterioration in balance performance after a prolonged cognitive task, the increase in CoP variability can be interpreted as increasing cognitive involvement in postural control (Donker et al., 2007). Moreover, our results consolidate previous findings on the fatigue-inducing effects of multi-joint exercises resulting in impairments of postural control. Future studies should include analyses of CoP parameters such as CoP variability or CoP regularity (Hachard et al., 2020) as traditional CoP parameters (e.g., CoP_d) tend to be less sensitive to detect initial changes in postural control due to MF.

Effects of Fatigue on Cortical Activity

In terms of cortical activity, our results reveal widespread increases in relative alpha-2 and beta-1 frequency band power in combination with more restricted increases in relative theta (i.e., central and parietal) and beta-2 power (i.e., parietal) after both fatigue protocols. Previous studies examining the effects of

PF on cortical activity reported increases in theta, alpha, and beta frequency band power at frontal, central, and parietal brain areas (Crabbe and Dishman, 2004; Bailey et al., 2008; Dishman et al., 2010; John et al., 2020). Moreover, MF studies reported increases in fronto-central theta power (Craig et al., 2012; Tanaka et al., 2012; Wascher et al., 2014; Fan et al., 2015) as well as increases in the alpha (Craig et al., 2012; Fan et al., 2015) and beta power (Craig et al., 2012) in fronto-central and parieto-occipital areas after mental fatiguing tasks in healthy adults.

For example, Bailey et al. (2008) investigated the influence of a graded exercise test until exhaustion on cortical activity at frontal, central, and parietal electrode sites. Their results indicated increases in theta, alpha-1, alpha-2, beta-1, and beta-2 frequency band power relative to baseline across selected frontal, central, and parietal areas. Similar findings were obtained by John et al. (2020), who also observed increases in theta, alpha, and beta frequency band power at frontal, central, and parietal brain areas during and after performing a graded exercise test until exhaustion. The authors suggested that stronger cortical involvement might reflect adaptive mechanisms to maintain information processing and to cope with fatigue-induced impairments by increasing attentional processes.

Regarding MF, Wascher et al. (2014) examined the effects of a prolonged spatial stimulus-response-compatibility task on cortical activity in healthy young adults. Results showed a continuous increase of theta power at electrodes over fronto-central areas of the cortex as well as an increase of alpha-1 and alpha-2 power at fronto-central electrode sites with longer duration of time on task. Moreover, Craig et al. (2012) examined changes in EEG measures that occurred during a driving simulator task in healthy adults. With proceeding MF, spectral power within the theta, alpha-1, alpha-2, beta-1, and beta-2 frequency band increased in frontal, central, and parietal areas and were accompanied by decreasing task performance. Thus, Craig et al. (2012) assumed that a fatigue-induced decline in cognitive capacity associated with impaired task performance might be related to global increase in theta, alpha 1, and alpha 2 wave power.

However, when comparing the present results with previous findings, it should be borne in mind that previous studies investigated fatigue-induced changes on cortical activity in sedentary participants at rest and not during performance of a challenging balance task. Widespread increases in relative alpha-2 and beta-1 frequency band power in combination with the more localized increases in relative theta (i.e., central and parietal) and beta-2 power (i.e., parietal), therefore, might reflect fatigue-related changes in cortical network activity involved in postural control (Solis-Escalante et al., 2019; Varghese et al., 2019). More specifically, changes in alpha-2 power over centro-parietal areas could be linked to balance performance decreases after PF. However, it should be noted that pre-post changes (deltas) in CoP variables and alpha-2 power did not correlate significantly with each other. Accordingly, the interpretation of the cortical mechanisms behind the observed fatigue-related effects on balance performance remain speculative and further research is needed to elucidate this issue. According to the literature, declines of alpha-2 power in centro-parietal areas, where the sensorimotor cortex and precuneus are localized, represent task specific sensory and movement-related information processing (Leocani et al., 1997; Neuper and Pfurtscheller, 2001). Consequently, the observed pre to post increases of relative alpha-2 power in these areas with concomitant decreases in balance performance might indicate deterioration of those processes. This deterioration might result in impaired sensory information processing related to movement planning and execution during balance control (Robertson and Marino, 2015).

Another explanation for the large increases of parietal alpha-2 power might be increased activity of somatosensory afferents after exercise (Dishman et al., 2010). Altered sensory input from muscle spindles and tendon organs (Paillard, 2012) as well as from muscle afferents (i.e., groups III and IV) that innervate the fatigued muscles (Gandevia, 2001) during the postural task, thus, might explain changed activity at central and parietal electrode sites overlaying portions of the somatosensory cortex and the precuneus, a structure known to be involved in dynamic balance control revealed by imaging data (Papegaaij et al., 2017).

Moreover, Benedek et al. (2014) suggested that parietal alpha power increases might reflect an inhibition of the ventral attention network which is thought to prevent reorienting to irrelevant stimulation during goal-driven, top-down behavior. This in turn might be indicative for a shielding function during cognitive demanding tasks to avoid/minimize distractor interference.

Thus, parietal alpha-2 power increases following the MF and PF protocol might represent shielding of specific cortical areas in an effort to maintain relevant task-specific information processing. It might also be possible that all, a shielding function, increased sensory afferent input and impaired task specific sensory, and movement-related information processing, are held responsible for the observed increments in alpha-2 power. As brain dynamics are inherently multi-scale, the underlying cortical mechanisms for the observed findings on balance decrements with fatigue cannot be elucidated with this study. Therefore, future studies should further investigate the role of centro-parietal alpha-2 activity in fatigue condition and if it is linked to changes in balance performance utilizing high-density EEG and source localization. Moreover, these studies should address other fatigue-related spectral EEG measures such as individual alpha peak frequency (Ng and Raveendran, 2007; Mierau et al., 2017). In this regard, also lateralization effects are of particular interest together with the specific effects of physical versus mental fatigue protocols on spectral EEG measures (e.g., alpha peak frequency).

Limitations

First, PF was assessed by means of a VAS reflecting only the subjective feeling of PF. The additional assessment of heart rate or blood lactate would have provided an objective measure to determine whether participants reached exhaustion. However, the PF protocol can be rated as effective as observed performance impairments were in line with the literature. Further, one could argue that the range from 2 to 21 min until task failure is large and might have confounded our results. As almost all participants had a high level of physical activity and participants were selected from a rather homogeneous cohort of sport science students, we consider that strength and fitness did not have an impact on the outcomes of this study.

Second, we had no control condition in which participants had to rest for a certain time (e.g., 30 min). Therefore, even though we controlled for a re-test effect, we cannot completely rule out that increases in relative frequency band power during performance of challenging balance task are solely attributable to the fatigue protocols. As previous controlled studies reported similar increases, it can be argued that the observed changes in cortical activity are very likely a result of PF and MF. Moreover, the time (∼30 s) between the termination of the fatigue protocol and the starting of post-tests could have been too long so that maybe fatigue-related effects were mitigated. We also did not have a seated control test measurement of EEG activity hence we cannot tell if the changes we report here are specific to the balance task or could have been present in sitting as well. There was little specificity between the fatigue tasks and the balance board test task. We cannot tell if fatigue would have been induced by a balance task and tested with the same balance task, the effects of PF would have been more profound on EEG spectral power.

Third, one can argue that the duration of the attention network test might have been too short to induce MF. In this context, Holtzer et al. (2011) were able to show that even a 30-min attention network test has a fatiguing effect on the executive attention network. Since the authors examined old adults in their study, it seems possible that the test duration would have to be slightly longer to induce MF in young adults. However, even though balance performance (i.e., CoP_d and CoP_v) was not impaired, CoP variability increased after both fatigue protocols. The observed changes in cortical activity, especially increases in alpha-2 power, therefore, seem to be indicative for a compensatory mechanism (Benedek et al., 2014) to maintain relevant task-specific information processing and thus performance during fatigue. Similarly, Hachard et al. (2020) observed increased cognitive contributions (i.e., decreased sample entropy) to postural control while balance performance remained unchanged. Consequently, balance performance must not be necessarily impaired in a state of MF if sufficient cognitive resources can be allocated to compensate for the negative impact of fatigue on postural control.

Finally, it has to be noted that causality between changes in alpha-2 power over centro-parietal areas with measures of balance after PF remains speculative due to the absence of a significant correlation between changes in alpha-2 power and CoP variables. However, Hedge et al. (2018) argued that the reliability of the single behavioral and neurophysiological measures might be a problem with such correlative analyses. Furthermore, inferences from electrode signals to the origin of changes in the oscillatory activity of the brain are speculative, as the brain works as a volume conductor. Therefore, more studies utilizing high-density EEG, source localization, and co-registration are needed in the future to investigate a possible relation between the effects of PF and MF on balance performance and cortical activity.

CONCLUSION

In summary, the present study revealed decreased balance performance indicated by increased CoP_d and CoP_v after performing a PF but not MF protocol. In addition, MF and PF resulted in increased CoP variability. Cortical activity in the form of relative frequency band power increased across almost all ROIs and frequency bands, irrespective of the fatigue protocol under investigation. Notably, increases in relative alpha-2 power in the parietal ROI were significantly larger after PF. These increases in parietal alpha-2 power could reflect a fatigue-induced impairment of sensory information processing related to movement planning and execution within the somatosensory cortex and precuneus resulting in decreased balance performance and impaired postural control. However, other underlying cortical mechanisms including a shielding function and/or increased sensory afferent input might be held responsible for the observed increments in alpha-2 power. Thus, further research is required to disentangle the alternative interpretations.

DATA AVAILABILITY STATEMENT

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

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Local Ethics Committee of the University of Potsdam (application no. 12/2019). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

AG, AB, CS, TH, and UG conceived and designed the research. AG and AB conducted the experiments and analyzed the data. All authors contributed to the writing of this manuscript, read, and approved the manuscript.

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A comparison of placebo and nocebo effects on objective and subjective postural stability: a double-edged sword?

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Background: Positive expectations (i.e., placebo effect) can improve postural control during quiet standing. This raises an important question: if postural control is susceptible to positive expectations, is it possible to elicit the opposite, a decline in postural stability, simply by suggesting a performance impairment (i.e., nocebo) will take place? Yet no studies have examined the nocebo effect on balance performance. To better understand both phenomena, comparative studies, which include both placebo and nocebo conditions, are needed.

Method: Forty-two healthy adults were initially assessed for objective (center of pressure movement) and subjective (perceived) postural stability and performance expectations. Participants were then randomly assigned in equal numbers to a placebo (positive expectation), nocebo (negative expectation) or control (no suggestion) group. Participants in the placebo/nocebo groups were deceptively administered an inert capsule described as a potent supplement which would either positively or negatively influence their balance performance. Objective and subjective postural stability, and performance expectations were reassessed 20 min later.

Results: The nocebo procedure evoked an increase in COP sway movements and reduced perceived stability compared to a control group. The placebo group presented with reductions COP sway movements and increased perceived stability following expectation manipulation. Compared to the control group, the placebo group showed a significantly higher performance expectation whilst the nocebo group showed a significantly lower performance expectation. Regression analyses also revealed that performance expectations following the placebo/nocebo procedure significantly predicted perceptions of postural instability (i.e., perceived performance), accounting for around 50% of the variance. These results remained even when controlling for *actual* performance (i.e., objective postural stability).

Conclusion: Our findings indicate that positive and negative performance expectations evoked by instructional manipulation can profoundly influence both objective and subjective postural stability. Postural control—and perceptions regarding such—are clearly susceptible to expectation manipulation, which could have important practical implications and repercussions on testing, training interventions and rehabilitation programs. Positive and negative expectancies are a double-edged sword for postural control.

KEYWORDS

placebo, nocebo, expectation, postural control, subjective stability, belief

Introduction

Placebos and nocebos are physiologically inert substances (i.e., pharmacological/nutritional) or sham interventions (i.e., psychological, physical or mechanical), which produce complex psychobiological responses independent of any direct therapeutic effects (Price et al., 2008; Turi et al., 2018). The *placebo* effect (originating from the Latin phrase "I will please") refers to a desirable outcome attributable to a purported beneficial treatment (Hurst et al., 2020). In contrast, the *nocebo* effect (originating from the Latin phrase "I will harm") refers to an undesirable outcome to a purported harmful treatment, administered with or without deliberate damage intention (Beedie et al., 2018). Placebo and nocebo effects are often explained on the basis of the recipient's expectancies about the received substance or intervention. These expectancies may be the result of conscious (e.g., expectations) and non-conscious (e.g., conditioning) cognitive or affective processes (Wager and Atlas, 2015).

Although placebo (analgesia) and nocebo (hyperalgesia) phenomena have been largely confined to the study of pain tolerance (Frisaldi et al., 2015), there is emerging evidence elucidating the application of these effects on motor and cognitive performance. In this regard, it is now well established that placebo effects can positively influence muscle force production (Fiorio et al., 2014; Villa-Sánchez et al., 2019), increase fatigue resistance (Piedimonte et al., 2015) and improve attention/vigilance (Colagiuri and Boakes, 2010). Despite convincing evidence that placebo related expectations can induce positive changes in several cognitive/motor functions, the potential effects of placebos on other important cognitive-motor functions that are essential to normal everyday functioning, such as balance performance, are less well understood.

It is well established that older adults frequently hold inappropriate expectations related to their balance abilities, with many individuals holding either over- or under-confident beliefs (Delbaere et al., 2010). Despite this, there has been little focus in the literature on how such performance expectations affect balance, as well as the efficacy of procedures (e.g., placebos) that can be used to modify these beliefs. Indeed, only one study has examined changes in postural control following a placebo procedure. Young adults who were made to believe that a placebo treatment was effective (application of an inert electrical device over the leg muscle) presented with reduced postural sway and perceived their balance control to be subsequently more stable when compared to a control group (Villa-Sánchez et al., 2019). These findings indicate that instead of being regarded as a bias to control for in randomized control trials, placebos could be deliberately utilized and combined with established approaches to increase therapeutic efficacy of balance interventions (Enck et al., 2013; Schwarz and Büchel, 2015). Although the initial work of Villa-Sánchez and colleagues offers valuable insight into the effects of placebo effects on objective and subjective balance performance, to our knowledge, no studies have examined the nocebo effect on balance performance. In order to better understand both phenomena, comparative studies, which include both placebo and nocebo conditions, are needed.

Previous research has shown that information disclosure about potential side effects of a treatment (i.e., nocebo), may create expectancies which contribute to adverse effects and prevention of improvement. For example, the nocebo effect can negatively influence muscle force production (Pollo et al., 2012; Emadi Andani et al., 2015; Corsi et al., 2019), endurance performance (McLemore et al., 2020), vigilance (Harrell and Juliano, 2009), response accuracy (Turi et al., 2018), and reaction time (Colagiuri et al., 2011). However, the interaction between balance performance and the nocebo effect are unknown. This raises an important question: if postural control is susceptible to positive expectations, is it possible to elicit the opposite, a decline in postural stability, simply by suggesting a performance impairment will take place? If so, this would have far-reaching implications for applied practice, given that such negative expectations could potentially elicit profound repercussions by interfering with training adaptations. Nocebos have also been shown to affect perceptual processes, resulting in individuals perceiving stimuli as being more painful and fatiguing (Reicherts et al., 2016; Wolters et al., 2019; Feldhaus et al., 2021). Given the clear dissociation between *actual* and *perceived* postural instability in a range of clinical balance disorders (Kaski, 2020; Castro et al., 2022), is it possible to induce such similarly distorted perceptions of instability *via* negative performance expectations elicited *via* a nocebo? Research is needed to fill these knowledge gaps.

The aim of the present study is to directly compare the influence of placebo and nocebo instructions on objective and subjective postural stability and performance expectancies compared to a no-treatment control group. The utilization of a no-treatment control will enable us to accurately estimate the relative magnitude of placebo and nocebo effects in response to treatments. Our hypotheses are as follows: (1) performance expectations would be high in the placebo group and low in the nocebo group; (2) positive performance expectancies (placebo) would result in improved objective (i.e., center of pressure movements) and subjective (i.e., perceived stability) postural stability; and (3) negative performance expectancies (nocebo) would result in reduced objective and subjective postural stability. Finally, we also predicted that performance expectations would predict perceptions of stability (i.e., perceived performance) irrespective of *actual* stability (i.e., objective performance).

Methods

Participants

Effect size (Cohen's d) were calculated from a similar study from mean changes in postural sway (large effect size, d = 1.20) in a placebo group (Villa-Sánchez et al., 2019). Power analysis (G*Power, v3.1.9.4) showed that for a repeated measures ANOVA analyses a minimum of 42 participants (n = 14 per group) would be required to be able to detect a significant within-between interaction of medium effect size [assuming 1-β = 80%, α = 0.05, Cohen's f = 0.25 (standardized medium effect size), three groups, and two within-subject conditions]. Whilst previous research has reported a large effect size of placebo effects on balance performance (Villa-Sánchez et al., 2019), we chose a more conservative medium effect size estimate because the relatively low number of investigations will inherently increase the uncertainty of the true population estimate. All participants initially completed a health screening questionnaire to assess eligibility for the study. Inclusion criteria were age between 18 and 35 years. Exclusion criteria were self-reported history of psychiatric, neurological, cardiovascular or pulmonary diseases, orthopedic pathology or musculoskeletal dysfunctions. Additionally, none of the participants reported any allergies, prior adverse responses to medication, or current health problems requiring medication were excluded from the study to minimize the possibility of adverse responses to the belief that an active treatment had been received (Beedie, 2007). Following baseline assessment, all participants were randomly assigned to one of three groups: (1) placebo (positive belief); (2) nocebo (negative belief); and (3) control (no suggestion; **Table 1**). There were no statistically significant differences between the three groups for age, sex or self-reported physical activity levels ($p > 0.05$). Participants provided written, informed consent prior to data collection. The experimental procedures were carried out in accordance with the standards outlined in the declaration of Helsinki (1964) and the study received approval by the institutional ethics committee (Application ID: P126096).

TABLE 1 Mean and SD participant characteristics.

	Placebo (n = 14)	**Nocebo (n = 14)**	**Control (n = 14)**
Sex (female)	7	6	7
Age (years)	21.0 ± 1.7	20.1 ± 0.9	23.0 ± 3.3
Height (m)	1.80 ± 0.09	1.76 ± 0.08	1.79 ± 0.08
Mass (kg)	74.3 ± 13.1	74.2 ± 15.9	75.3 ± 16.3
BMI (kg/m^2)	22.9 ± 2.2	23.8 ± 4.3	24.7 ± 4.1
Physical activity (h·wk^{-1})	2.6 ± 1.2	2.6 ± 0.9	2.7 ± 1.4

Design

To minimize cross-contamination between experimental and control treatments, this study was conducted as a randomized controlled parallel trial (**Figure 1**). The importance of a no-treatment control group alongside placebo and/or nocebo group has recently been highlighted in the literature (Beedie et al., 2018; Colloca and Barsky, 2020). Primary outcome measures were objective (posturography) and subjective (perceived postural stability) balance performance. Secondary outcomes were subjective performance expectation, perceived change in performance, and adverse symptoms. Following baseline assessments, participants were allocated to a placebo, nocebo or control group. The randomization process was done using Research Randomizer, a program published on a publicly accessible official website[1]. During baseline assessments, the principle investigator was blind to treatment allocation. Objective and subjective postural stability and performance expectations were reassessed 20 min later. Subjective performance expectation was measured before baseline and experimental conditions. Perceived change in performance was rated after the experimental trial. Adverse symptoms were assessed before the baseline condition and after the experimental trial. In accordance with recent recommendation (Horváth et al., 2021) participants in the no-treatment control group underwent the same procedure but did not receive any capsules and were not further instructed regarding positive or negative suggestions (**Figure 1**).

1 www.randomizer.org

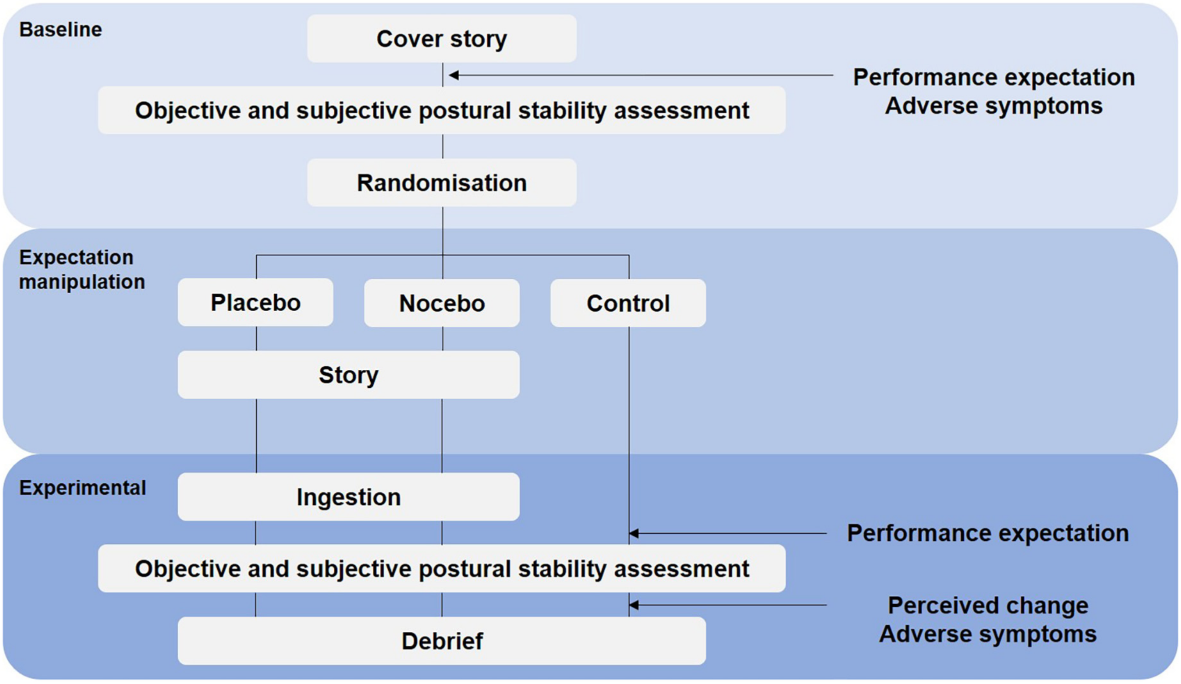

FIGURE 1
Schematic of the experimental design.

Belief manipulation

In accordance with previous experiments (i.e., Beedie, 2007; Hurst et al., 2017), during the 20-min interval between baseline and experimental conditions, participants in the placebo and nocebo groups were administered a capsule described as a potent sports supplement, "inorganic nitrate". Participants in the placebo group (i.e., positive belief treatment) were asked to ingest two, size 0 (volume; 0.68 ml, size; 21.5 mm) clear gelatine capsules containing 200 mg of corn flour (Morrison's, Bradford, UK) and informed that inorganic nitrate would "improve mental alertness and enhance muscle force production", which are important determinants of balance performance. Participants in the nocebo group (i.e., negative belief treatment) were also asked to ingest two, size 0 (volume; 0.68 ml, size; 21.5 mm) clear gelatine capsules containing 200 mg of corn flour, but were informed that inorganic nitrate can "dampen the activity of the central nervous system, reduce alertness and causes sensations of tiredness, fatigue and lethargy" which would decrease balance performance. Similar to previous studies (Hurst et al., 2017), the effectiveness of the belief manipulation (placebo and nocebo groups) was assessed during a debrief immediately after the experimental trials, at which point the true nature of the study was revealed. Participants were asked to respond on a visual analogue scale (VAS) "how much did you believe the treatment influenced your performance" (from 0, no influence to 10, high influence). We also asked participants to respond on a VAS, "how much did you believe the information you received" (from 0, did not believe at all to 10, completely believed).

Postural stability

Quantitative posturography

During baseline and experimental assessments, participants performed three 30-s quiet standing trials on a force platform (AMTI, AccuGait, Watertown, MA) with 15-s rest between each trial. To further explore the possible mediating influence of task difficulty, participants completed trials under both bipedal and unipedal conditions, in a randomized order. To ensure continuity between trials, participants were unshod and instructed to stand quietly with the hands clasped together in front of the body. In the bipedal stance, participants stood with the feet together (Romberg stance; Objero et al., 2019). For the unipedal trials, participants maintained a single-leg stance with the dominant limb (defined as the foot used to kick a ball). Participants were instructed that the unloaded leg should not touch the supporting leg and the knee should be flexed to 90°. During all quiet standing trials, participants were asked stand quietly on the force platform while and minimizing any extraneous movements and gazing at a target 1.5 m from the force platform. Participants could *step off* the plate and rest between trials (±15 s). Ground reaction force data were sampled at 100 Hz (AMTI, Netforce, Watertown, MA) and filtered using a fourth-order low-pass (6 Hz) Butterworth filter (BioAnalysis, V2.2, AMTI) prior to calculation of center of pressure (COP) metrics. The maximal displacement (i.e., distance between the maximum and minimum COP displacement) of center of pressure (COP) in the anteroposterior (AP) and mediolateral (ML) directions (both cm) were subsequently calculated (AMTI, BioAnalysis, Version 2.2, Watertown, MA). The validity and reliability of these parameters have previously been established for this sampling duration (Pinsault and Vuillerme, 2009). An average of the three trial trials (total 90 s) recorded during baseline and experimental conditions were used in the subsequent analyses (Ruhe et al., 2010).

Perceived postural (in)stability

Immediately following each 30-s quiet standing trial, participants were asked to rate their degree of instability during the trial ("how stable did you feel during the trial?") using a 0–10 VAS, where 0 corresponded to being "completely steady" and 10 "so unsteady that I would fall" (Castro et al., 2019). This was based on the subjective stability scoring system originally proposed by Schieppati et al. (1999) and has been shown to be valid and reliable for bipedal and unipedal balance tasks in healthy young adults (Hauck et al., 2008). As with posturographic data, an average of the three trials recorded during baseline and experimental conditions were used in subsequent analyses.

Questionnaires and self-report

Subjective performance expectation

To assess participants subjective performance expectation (also referred to as self-efficacy), we used the item "I will perform well in the task" (Winkler and Hermann, 2019) to be rated on a VAS ranging from 0 to 10 (0 being "do not agree at all", 5 "neutral" and 10 being "totally agree"). Performance expectation was measured before each block of baseline and experimental trials (**Figure 1**).

Perceived change in performance

Immediately following the experimental trials, participants were asked to rate the perceived change in balance performance between the baseline condition (i.e., before performance expectation manipulation) and the experimental assessment (i.e., after performance expectation manipulation; **Figure 1**). Participants were asked "how do you rate your balance performance now in comparison to the first assessment?" on a VAS ranging from 0 (worse) to 10 (better).

Adverse symptoms

Perceived adverse symptoms (or side effects) of the administered capsules were assessing using an adapted version of the Generic Assessment of Side Effects Scale (GASE; Rief et al., 2011). Although the original GASE consists of 36 symptoms, for the purpose of our study, 12 adverse symptoms were selected to match the potential side effects described in the participant information sheet (headache, fatigue, irritability, dizziness, nausea, feeling of weakness, drowsiness, tremor, muscle pain, anxiety/fear). As recommended (Rheker et al., 2018; Winkler and Hermann, 2019), we assessed adverse symptoms before the baseline and after the experimental conditions.

Statistical analysis

Data were analyzed using SPSS version 25.0 (IBM Inc., Chicago, IL). For all analyses, normality (Shapiro–Wilk Test) and homogeneity of variance/sphericity (Mauchly Test) were performed and confirmed prior to parametric analyses. To examine differences in objective and subjective postural stability, subjective performance expectation and adverse symptoms, a series of two-way mixed model analysis of variance (ANOVA) were undertaken (with Bonferroni correction) to test for the within-subject effects of time [× 2 (baseline vs. experimental)] and between subject effects of group [× 3 (placebo vs. nocebo vs. control)]. Therefore, where VAS data was normally distributed, parametric tests were employed. A one-way ANOVA was undertaken to assess differences in perceived change in performance between the three groups (placebo vs. nocebo vs. control). The effectiveness of belief manipulation was assessed using an independent t-test (placebo vs. nocebo). The Mann-Whitney *U* test was used for adverse symptoms (GASE: 0 = no symptoms, 1 = mild symptoms, 2 = moderate symptoms, 3 = severe symptoms) for pairwise comparisons. Where significant interactions or main effects were detected, *post hoc* analyses using Bonferroni-adjusted α determined the location of any differences. For ANOVA, effect sizes are reported as partial eta-squared value (η^2). Cohen's *d* effect sizes (ES) are reported for *post hoc* comparisons with an effect size of 0.2, 0.6, 1.2 and 2.0 indicating small, medium, large and very large effects, respectively. The alpha value was *a priori* set at $p < 0.05$ for all tests. Given that placebo effects can demonstrate considerable variability (Beedie and Foad, 2009), we reported inter-individual responses to treatments.

We also performed regressional analyses to examine whether performance expectations influence perceptions of postural stability (i.e., perceived performance) independent of *actual* stability (i.e., objective performance). We conducted a separate regression for each task (bipedal vs unipedal) and condition (pre- and post-belief manipulation). For each regression, perceived stability for that given task/condition was entered as the outcome variable, whilst the predictor variables were: performance expectation (for that given condition) and objective task performance for that given task/condition (AP and ML COP displacement). The assumptions of homoscedasticity (inspecting the standardized residuals by standardized predicted values plot), error-independence (Durbin–Watson values = 1.67–1.83), lack of multicollinearity (variance inflation factors <2.05, tolerances >0.49), and normal distribution of errors (determined with Kolmogorov–Smirnov tests and inspection of histogram of residuals) were verified.

Results

Performance expectations

The mixed-model ANOVA revealed a statistically significant group × time interaction for performance expectation ($F_{(2,78)} = 27.462$, $p < 0.001$, $\eta_p^2 = 0.413$). Follow up *post hoc* analyses revealed that the groups differed significantly in their performance expectations after expectation manipulation, but there were no differences at baseline (**Figure 2**). Participants in the placebo group ($M = 8.9$, $SD = 0.7$) reported a statistically significantly higher performance expectation than participants in the nocebo ($M = 5.9$, $SD = 0.9$, $p < 0.001$, $M_{diff} = 3.02$, $d = 3.71$) and control ($M = 7.6$, $SD = 0.7$, $p < 0.001$, $M_{diff} = 1.29$, $d = 1.88$) group, after expectation manipulation. The nocebo group reported a statistically significantly lower performance expectation that the control group ($p < 0.001$, $M_{diff} = 1.73$, $d = 2.09$), after expectation manipulation. Moreover, performance expectation increased significantly in the placebo group following the expectation manipulation ($p < 0.001$, $M_{diff} = 1.34$, $d = 1.79$), and it decreased significantly in the nocebo group ($p < 0.001$, $M_{diff} = 1.86$, $d = 2.03$) but not the control group ($p > 0.05$, $M_{diff} = 0.06$, $d = 0.08$).

Perceived change in performance and effectiveness of belief manipulation

The mixed-model ANOVA revealed a statistically significant group main effect ($F_{(2,41)} = 92.248$, $p < 0.001$) for perceived change in balance performance (**Figure 3A**). Follow up *post hoc* analyses revealed that the placebo group ($M = 6.8$, $SD = 0.9$) reported a statistically significantly greater improvement in performance than the nocebo ($M = 3.2$, $SD = 0.7$, $p < 0.001$, $M_{diff} = 3.6$, $d = 4.48$) and control group ($M = 5.2$, $SD = 0.4$, $p < 0.001$, $M_{diff} = 1.6$, $d = 2.31$), whilst the nocebo group reported a statistically significantly lower balance performance compared to the control group ($p < 0.001$, $M_{diff} = 2.0$, $d = 3.34$). There was no statistically significant difference in perceived belief that the treatment influenced participants performance between the placebo ($M = 6.7$, $SD = 1.0$) and nocebo ($M = 6.5$, $SD = 1.3$, $p > 0.05$, $M_{diff} = 0.2$, $d = 0.18$) groups (**Figure 3B**). All participants in the placebo ($M = 8.8$, $SD = 1.1$) and nocebo ($M = 8.3$, $SD = 1.2$) group believed the information that they

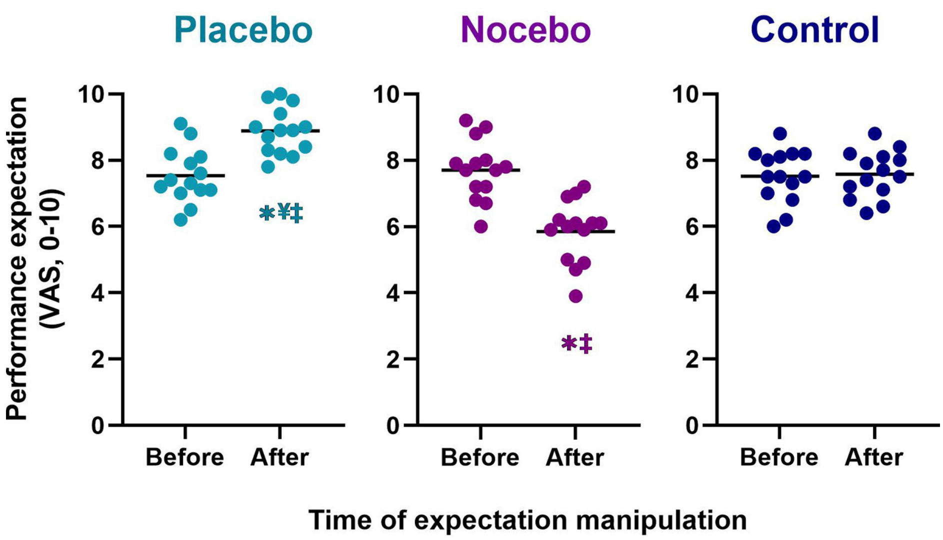

FIGURE 2
Performance expectation before the first balance assessment and after the expectation manipulation for placebo, nocebo, and control groups. *Statistically significantly different to before the first balance assessment. ¥Statistically significantly different to nocebo group after expectation manipulation. ‡Statistically significantly different to control group after expectation manipulation.

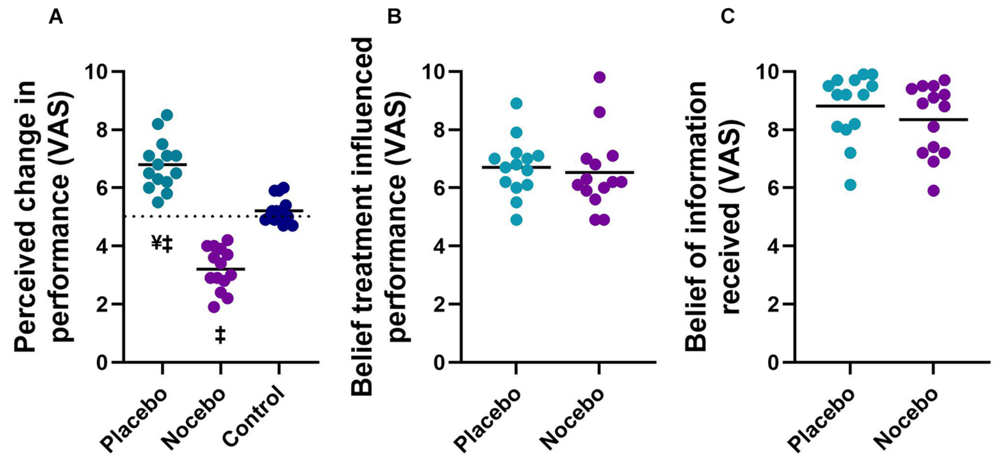

FIGURE 3
Perceived change in balance performance **(A)**, the degree to which participants believed the treatment influenced their performance **(B)**, and the degree to which participants believed the information they received **(C)** in placebo, nocebo, and control groups. [¥]Statistically significantly different to nocebo group. [‡]Statistically significantly different to control group after expectation manipulation.

received (scored > than 5; neutral; Figure 3C). There was no difference in how much participants believed the information that they received between the placebo and nocebo groups ($p > 0.05$, $M_{diff} = 0.5$, $d = 0.40$).

Objective and subjective postural stability during bipedal stance

Mixed-model ANOVA revealed a statistically significant group × time interaction for the anteroposterior COP range ($F_{(2,78)} = 11.471$, $p < 0.001$, $\eta_p^2 = 0.227$), mediolateral COP range ($F_{(2,78)} = 5.338$, $p = 0.007$, $\eta_p^2 = 0.120$) and subjective postural stability ($F_{(2,78)} = 21.780$, $p < 0.001$, $\eta_p^2 = 0.358$; Figure 4).

Anteriorposterior COP range

Although there were no differences at baseline, participants in the placebo group ($M = 1.92$, $SD = 0.35$) demonstrated a statistically significantly smaller anteroposterior COP range than participants in the nocebo ($M = 2.81$, $SD = 0.87$, $p < 0.001$, $M_{diff} = 0.89$, $d = 1.35$) but not the control ($M = 2.08$, $SD = 0.35$, $p > 0.05$, $M_{diff} = 0.16$, $d = 0.46$) group, after expectation manipulation (Figures 4A–C). The nocebo group reported a statistically significantly greater anteroposterior COP range than the control group ($p = 0.001$, $M_{diff} = 0.73$, $d = 1.11$), after expectation manipulation. Moreover, anteroposterior COP range decreased significantly in the placebo group ($p = 0.012$, $M_{diff} = 0.51$, $d = 1.16$), and it increased significantly in the nocebo group ($p < 0.001$, $M_{diff} = 0.82$, $d = 1.21$) but not the control group ($p > 0.05$, $M_{diff} = 0.06$, $d = 0.14$), with respect to baseline assessments.

Mediolateral COP range

After expectation manipulation, there were no statistically significant differences in the mediolateral COP range between the placebo ($M = 2.15$, $SD = 0.46$) with nocebo ($M = 2.44$, $SD = 0.71$, $M_{diff} = 0.30$, $d = 0.50$) or control ($M = 2.18$, $SD = 0.35$, $M_{diff} = 0.04$, $d = 0.09$) groups, or between the nocebo and control ($M_{diff} = 0.26$, $d = 0.47$) groups (all $p > 0.05$; Figures 4D–F). However, the mediolateral COP range decreased significantly in the placebo group ($p = 0.021$, $M_{diff} = 0.39$, $d = 1.0$), and it increased significantly in the nocebo group ($p = 0.026$, $M_{diff} = 0.38$, $d = 0.65$) but not the control group ($p > 0.05$, $M_{diff} = 0.01$, $d = 0.04$), with respect to baseline assessments.

Subjective postural (in)stability

Although there were no differences at baseline, participants in the placebo group ($M = 1.02$, $SD = 0.48$) reported statistically significantly lower subjective instability than participants in the nocebo ($M = 3.54$, $SD = 1.32$, $p < 0.001$, $M_{diff} = 2.51$, $d = 2.54$) and the control ($M = 2.06$, $SD = 0.81$, $p = 0.006$ $M_{diff} = 1.04$, $d = 1.56$) group, after expectation manipulation (Figures 4G–I). The nocebo group reported a statistically significantly greater subjective instability than the control group ($p < 0.001$, $M_{diff} = 1.48$, $d = 1.35$), after expectation manipulation. Moreover, subjective postural instability decreased significantly in the placebo group ($p = 0.002$, $M_{diff} = 1.06$, $d = 1.59$), and it increased significantly in the nocebo group ($p < 0.001$, $M_{diff} = 1.91$, $d = 1.81$) but not the control group ($p > 0.05$, $M_{diff} = 0.04$, $d = 0.05$), with respect to baseline assessments.

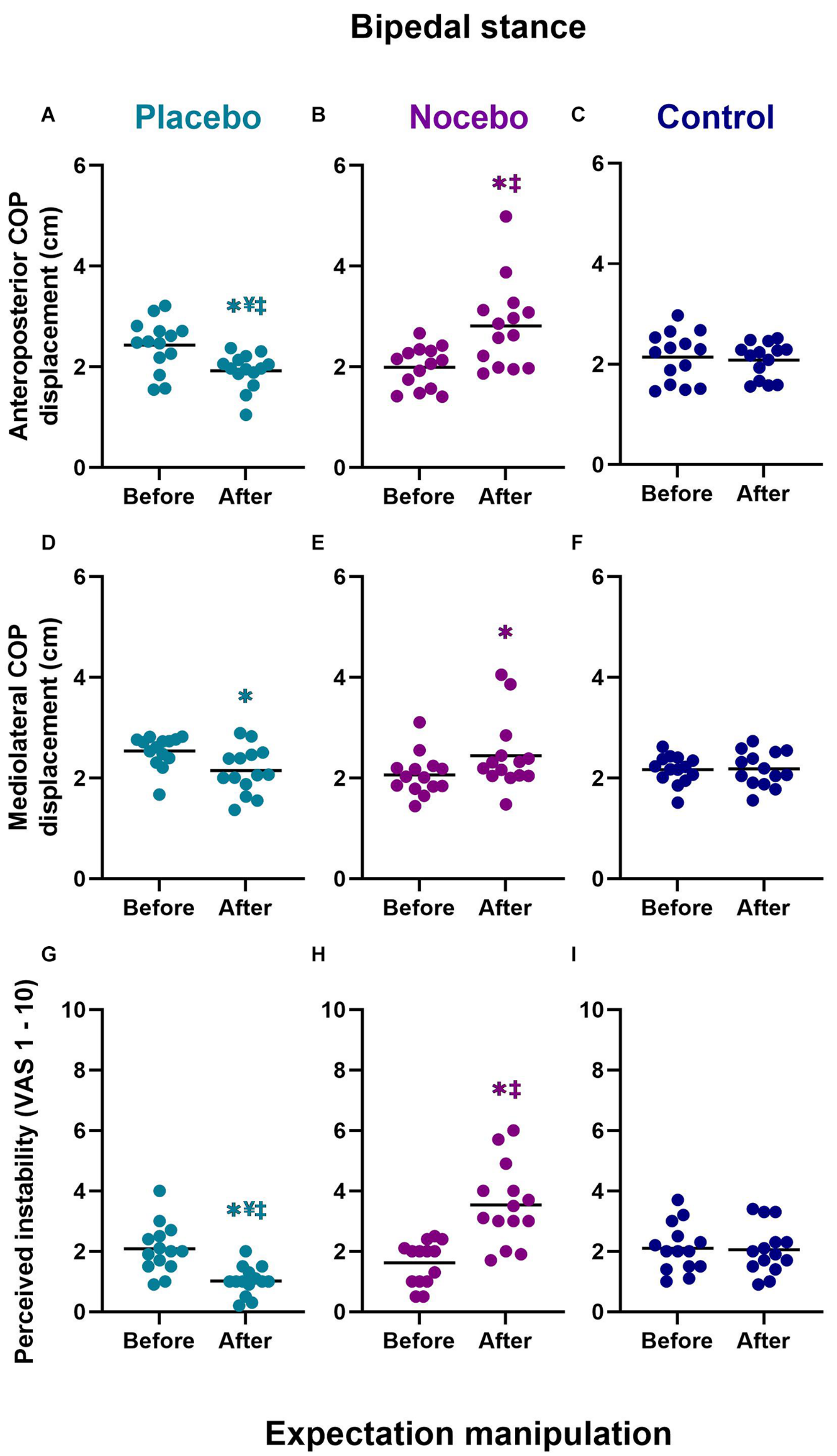

FIGURE 4
Objective (A–F) and subjective (G–I) bipedal balance performance before and after expectation manipulation for placebo, nocebo, and control groups. *Statistically significantly different to before the first balance assessment. ¥Statistically significantly different to nocebo group after expectation manipulation. ‡Statistically significantly different to control group after expectation manipulation.

Objective and subjective postural stability during unipedal stance

For unipedal stance, the mixed-model ANOVA revealed a statistically significant group × time interaction for the anteroposterior COP range ($F_{(2,78)}$ = 14.244, p < 0.001, η_p^2 = 0.268), mediolateral COP range ($F_{(2,78)}$ = 13.613, p < 0.001, η_p^2 = 0.259) and perceived postural stability ($F_{(2,78)}$ = 26.553, p < 0.001, η_p^2 = 0.405; **Figure 5**).

Anteriorposterior COP range

Although there were no differences at baseline, participants in the placebo group (M = 3.14, SD = 0.67) demonstrated a statistically significantly smaller anteroposterior COP range than participants in the nocebo (M = 4.48, SD = 1.25, p < 0.001, M_{diff} = 1.34, d = 1.34) but not the control (M = 3.36, SD = 0.68, p > 0.05 M_{diff} = 0.22, d = 0.33) group, after expectation manipulation (**Figures 5A–C**). The nocebo group reported a statistically significantly greater anteroposterior COP range the control group (p = 0.001, M_{diff} = 1.12, d = 1.11), after expectation manipulation. Moreover, anteroposterior COP range decreased significantly in the placebo group (p = 0.011, M_{diff} = 0.75, d = 1.24), and it increased significantly in the nocebo group (p < 0.001, M_{diff} = 1.38, d = 1.43) but not the control group (p > 0.05, M_{diff} = 0.06, d = 0.09), with respect to baseline assessments.

Mediolateral COP range

Although there were no differences at baseline, participants in the placebo group (M = 2.62, SD = 0.43) demonstrated a statistically significantly smaller mediolateral COP range than participants in the nocebo (M = 4.05, SD = 0.86, p < 0.001, M_{diff} = 1.44, d = 2.11) but not the control (M = 2.89, SD = 0.55, p > 0.05 M_{diff} = 0.27, d = 0.55) group, after expectation manipulation (**Figures 5D–F**). The nocebo group reported a statistically significantly greater mediolateral COP range than the control group (p < 0.001, M_{diff} = 1.16, d = 1.61), after expectation manipulation. Moreover, mediolateral COP range increased significantly in the nocebo group (p < 0.001, M_{diff} = 1.12, d = 1.71) but not the placebo (p > 0.05, M_{diff} = 0.53, d = 1.21) or control group (p > 0.05, M_{diff} = 0.01, d = 0.02), with respect to baseline assessments.

Subjective postural (in)stability

Although there were no differences at baseline, participants in the placebo group (M = 4.09, SD = 1.07) demonstrated a statistically significantly lower subjective postural instability than participants in the nocebo (M = 7.29 SD = 1.13, p < 0.001, M_{diff} = 3.19, d = 2.90) and the control (M = 5.25, SD = 0.96, p = 0.013 M_{diff} = 1.16, d = 1.14) group, after expectation manipulation (**Figures 5G–I**). The nocebo group reported a statistically significantly greater subjective instability than the control group (p < 0.001, M_{diff} = 2.04, d = 1.94), after expectation manipulation. Moreover, subjective postural instability decreased significantly in the placebo group (p < 0.001, M_{diff} = 2.26, d = 1.97), and increased significantly in the nocebo group (p < 0.001, M_{diff} = 1.79, d = 1.66) but not the control group (p > 0.05, M_{diff} = 0.02, d = 0.02), with respect to baseline assessments.

Adverse symptoms

Mann-Whitney U test showed a statistically significant difference in the level of perceived fatigue reported in the nocebo group compared to the placebo (U = 23.0, p < 0.001) and control (U = 32.0, p = 0.001) group after expectation manipulation (**Figure 6**). Similarly, there was statistically significant difference in the level of perceived weakness reported in the nocebo group compared to the placebo (U = 32.5, p = 0.001) and control (U = 55.5, p = 0.031) group after expectation manipulation (**Figure 6**). Finally, there was a statistically significant difference in the level of perceived drowsiness reported in the nocebo group compared to the placebo and control (both; U = 60.0, p = 0.034) group after expectation manipulation (**Figure 6**).

Regression analyses

The regression models did not significantly predict perceived instability during the pre-manipulation phase, for either bipedal (R^2 = 0.079, $F_{(3,38)}$ = 1.08, p = 0.369) or unipedal task performance (R^2 = 0.055, $F_{(3,38)}$ = 0.73, p = 0.539). However, post-manipulation, the regression model for both bipedal (R^2 = 0.577, $F_{(3,38)}$ = 17.30, p < 0.001) and unipedal (R^2 = 0.560, $F_{(3,38)}$ = 16.13, p < 0.001) task performance was significant. In both models, the only significant predictor of perceived instability was performance expectations (bipedal: β = −0.73, p < 0.001, unipedal: β = −0.57, p = 0.003); with lower performance expectations predicting greater perceived instability. In contrast, neither *actual* COP displacement in either the AP (bipedal: β = −0.05, p = 0.860, unipedal: β = 0.30, p = 0.175) or ML direction (bipedal: β = −0.03, p = 0.926, unipedal: β = 0.31, p = 0.272) significantly predicted perceptions of instability.

Discussion

In this study, we investigated whether placebo and nocebo effects in postural stability can be elicited by evoking positive and negative expectations about task performance.

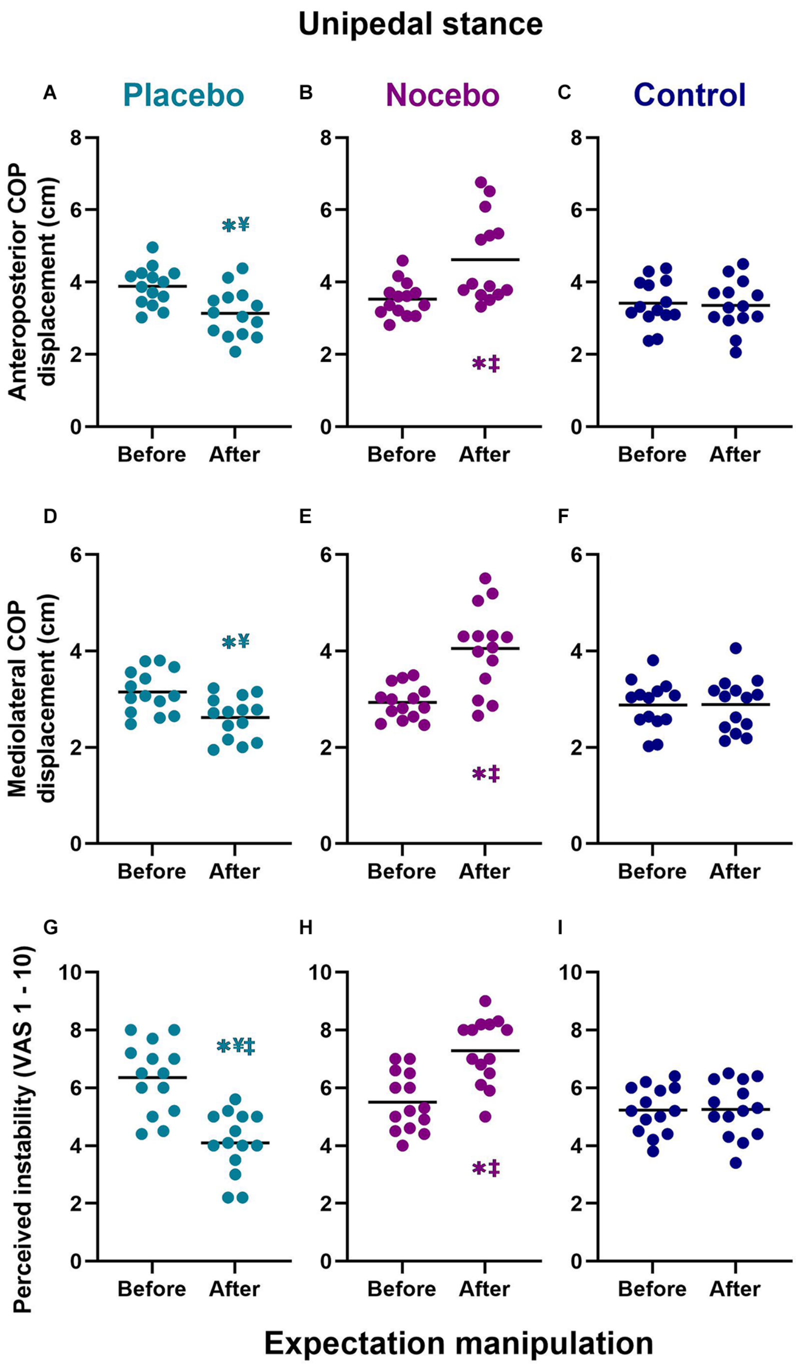

FIGURE 5
Objective **(A–F)** and subjective **(G–I)** unipedal balance performance before and after expectation manipulation for placebo, nocebo, and control groups. *Statistically significantly different to before the first balance assessment. ¥Statistically significantly different to nocebo group after expectation manipulation. ‡Statistically significantly different to control group after expectation manipulation.

Performance expectations were manipulated by deceptively instructing participants about alleged beneficial (placebo group) or detrimental (nocebo group) effects of a dietary supplement. To our knowledge, the present study is the first on postural control to include both placebo and nocebo instructions in a single experimental design. By directly comparing placebo and nocebo effects with a group that received no-treatment (to control for repeated testing), we are able to demonstrate a bidirectional postural response to positive and negative verbal suggestions.

Specifically, our data indicate that a nocebo procedure can negatively modulate objective and subjective postural stability compared to a control group. In contrast, our placebo group presented with a marked improvement in postural control and increased perceived stability following expectation manipulation. Our findings indicate that positive and negative performance expectations, evoked by instructional manipulation, can profoundly influence performance expectation and both objective and subjective postural stability. Postural control is clearly susceptible to expectation manipulation, which could have important practical implications and repercussions on training interventions and rehabilitation programs. For example, physical therapists or practitioners who are not fully aware of the power of words may inadvertently elicit placebo or nocebo effects, which may substantially modulate the efficacy of evidence-based interventions. There may also be important practical implications for practitioners in other sports-related professions such as coaches and teachers in the importance of psychological factors (i.e., expectations) in shaping performance.

Performance expectations, belief manipulation and perceived change in performance

We initially hypothesized that administration of a placebo procedure would induce enhanced expectations about balance performance. We anticipated the opposite effects for the nocebo procedure. Consistent with our hypothesis and previous literature (Winkler and Hermann, 2019), the placebo group showed a significantly higher performance expectation post-manipulation compared to the nocebo and control group (**Figure 2**). Similarly, the nocebo group showed a significantly lower performance expectation post-manipulation compared to the placebo and control group. These findings confirm that placebo and nocebo procedures were highly effective in inducing positive and negative performance expectations, respectively. We also found that participants in the placebo and nocebo group strongly believed that the treatment influenced their performance (**Figure 3B**) and believed the information they were given (**Figure 3C**), further supporting the credibility of our belief-manipulation procedures. Furthermore, as predicted, participants in the placebo group rated their perceived change in balance performance significantly better than the control group (d = 2.31). Similarly, participants in the nocebo group rated their perceived change in balance performance significantly poorer than the control group (d = 3.34). Taken together our findings clearly indicate that manipulation of performance expectations strongly affects the perceived change in balance performance and that participants strongly believed the efficacy of the treatments.

Reduced postural sway following a placebo procedure

Reduced COP movements during quiet standing have previously been described in healthy young adults following a placebo procedure (Villa-Sánchez et al., 2019). In agreement with our hypothesis and previous literature, our findings confirm

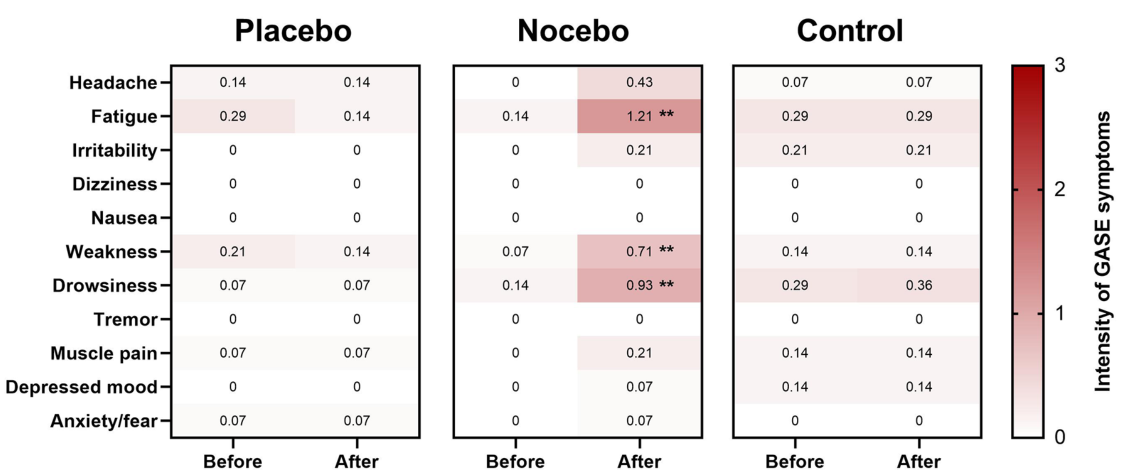

FIGURE 6
Heat map of adverse symptoms (or side effects) for the placebo, nocebo and control group. Shading indicates strength of the adverse effect [white = 0 (no symptoms), light pink = 1 (mild symptoms), dark pink = 2 (moderate symptoms), 3 = red (severe symptoms)]. Value in each box represents the mean response. **Statistically significantly different to both placebo and control groups (p < 0.05; adjusted by Bonferroni correction).

that participants in the placebo group exhibited a reduced range (greater precision achieved by the postural control system) of the COP and concomitantly increased perceived stability after expectation manipulation. Since postural sway in the control group did not decline, the smaller range of COP movements during quiet standing in the placebo group is unlikely to be caused by a learning effect. This raises an important question: what are the mechanisms that allow for a placebo procedure to reduce postural sway during quiet standing? There are at least two possibilities. One explanation could be related to an increased corticospinal excitability. There is strong evidence for a crucial contribution of several cortical structures (e.g., cerebral cortex) in maintaining upright stance (Jacobs and Horak, 2007; Bolton et al., 2012; Mierau et al., 2017). These brain structures could potentially be exploited by placebo procedures. For example, a previous study showed that a placebo procedure increased the activity of the primary motor cortex, increasing the excitability of the corticospinal system (Fiorio et al., 2014). Therefore, we speculate that the reduced postural sway reported here could be partly due to increased voluntary drive to the motor cortex, which would logically elicit favorable adaptations in balance performance.

Another possible important factor facilitating the placebo effects reported in the present study could be related to increased motivation (i.e., choice, effort and persistence) or self-efficacy (i.e., the belief that one can successfully execute a task in a specific context; Bandura et al., 1997) following expectation manipulation. For example, there is clear evidence showing that psychological factors such as motivation, self-efficacy, and attentional focus can impact balance performance (Wulf and Lewthwaite, 2009; Lewthwaite and Wulf, 2010; Wulf et al., 2012; Chua et al., 2020). Additionally, our placebo procedure may have enhanced positive affect and resulted in increased effort. However, in a previous study, positive verbal suggestions elicited a reduction in postural sway without any changes in balance effort (measured using the Borg 0–10 scale; Borg, 1982; Villa-Sánchez et al., 2019). Therefore, the relative contribution of increased effort and motivation on postural stability following a placebo procedure remain unclear. In the present study we did not assess balance effort or motivation because we assumed that all participants would approach balance testing with reasonable effort and would be motivated to do their best. Future research should measure motivation and effort to help elucidate the relative contribution of these factors on postural control in response to enhanced performance expectancies.

Increased postural sway following a nocebo procedure

The most striking finding presented here was a pronounced increase in postural sway in the nocebo group compared to the control group. Of course, an interesting question concerns the mechanism by which the nocebo procedure resulted in changes in COP movements during quiet standing. From a neurobiological perspective, nocebo procedures can inhibit dopaminergic neurological systems (resulting in decreased motivation/effort and self-efficacy; Beedie et al., 2018; Horváth et al., 2021), which may in turn hinder balance performance. It is also plausible that reduced performance expectations impaired balance performance through individuals directing attentional focus internally, in an attempt to consciously control movement. For example, research has described how consciously processing movements leads to distorted perceptions of instability (Ellmers et al., 2021), in addition to increased postural stability (Chow et al., 2019). However, we did not measure conscious movement processing in the present study and are therefore unable to determine the extent to which negative performance expectancies influenced attentional focus and subsequent balance performance. This issue should be explored in future studies.

An interesting feature of the data presented here was that participants in the nocebo group presented with feelings of fatigue, weakness and drowsiness after expectation manipulation. In this study, we gave detailed information about the (sham) mechanisms and subsequent effects on balance performance and explicitly pointed out the potential adverse symptoms that may be experienced in response to the nocebo procedure. In agreement with previous studies (Winkler and Hermann, 2019), fatigue was described as a side effect in the nocebo group. Unique to the present investigation, participants in the nocebo group also reported increased levels of weakness and drowsiness. Anecdotally, several participants in the nocebo group reported to postural tasks feeling "harder to perform" after expectation manipulation. However, we did not directly measure the amount of effort that was required to maintain balance in this study. Future studies should assess balance effort to help elucidate the mechanisms responsible for the nocebo effect on balance performance reported here.

Subjective postural stability

Another important finding from the present work is that performance expectations exert a strong influence over perceptions of balance performance (i.e., perceived stability), even when accounting for *actual* task performance (i.e., objective postural sway outcomes). Interestingly, this relationship was only present post-manipulation; with performance expectations failing to predict perceived stability pre-manipulation. These findings fit with Predictive Processing Frameworks of perception (e.g., Clark, 2015). These frameworks contend that perception is the consequence of an interaction between top-down expectations ("priors") and bottom-up sensory input ("prediction errors"). The extent to which each of these ultimately influence perception is determined by their predicted reliability ("precision"). For instance, if walking along a familiar street when it is dark and visibility is poor, incoming sensory input would lack precision, and perception would therefore be more strongly influenced by prior knowledge of what one would *expect* to see. Highly precise expectations are argued to explain visual illusions and hallucinations, whereby the objective sensory input does not match the subjective perception (Clark, 2015). We similarly contend that the false feedback that participants received in the present study was viewed as a highly precise source of information. This had the effect of disproportionately influencing perceptions of instability and balance performance. Participants who expected that they would perform poorly perceived themselves to be more unstable than those who expected that they would perform well—irrespective of *actual* performance (i.e., objective postural stability). Given that numerous clinical balance disorders are characterized by distorted perceptions of instability [whereby actual and perceived stability are decoupled (Ellmers et al., 2021)], these novel findings have high levels of applied relevance. Future work should explore if such distortions of perceived instability can be altered through targeting faulty expectancies about balance performance.

Implications

There are several important implications to be gleaned from the present study. We provide the first direct evidence that negative verbal suggestions can elicit marked increases in postural sway and reductions in perceived stability in healthy young adults. These findings are disconcerting and point towards negative suggestions and expectancies potentially interfering or even preventing balance adaptations during chronic training. Physical therapists, researchers, coaches, teachers, and practitioners who are not fully aware of the power of words may unknowingly and unintendedly introduce negative expectations that could have detrimental repercussions on testing and training. Although we manipulated performance expectations by asking participants to ingest a pill, even subtle expectancy manipulation, such as providing positive and negative performance feedback, can negatively influence balance performance (Wulf and Lewthwaite, 2009; Lewthwaite and Wulf, 2010; Wulf et al., 2012; Chua et al., 2020). Although participants should clearly be informed about potential risks associated with training (i.e., exercise-induced falls), care should be taken to avoid inadvertently eliciting iatrogeny (see Evers et al., 2021 for expert consensus).

Contrary to negative expectations, placebo effects facilitate optimal balance performance. The practical significance of these findings, as evidenced by the large effect sizes, appear to be quite meaningful. From an applied perspective, these findings may suggest that placebo procedures could be exploited in physical training to improve balance performance, and could also be used to increase beliefs/expectations about balance performance in those older adults with disproportionately low levels of balance confidence (e.g., Delbaere et al., 2010). On one hand, overselling the placebo effect might prove ethically problematical; trust between client and health professionals, or between participant and scientist, should be paramount (Beedie and Foad, 2009). Instead, through the use of positive verbal suggestions about the efficacy of an intervention, practitioners can use the information presented here to increase the likelihood of performance improvements (Evers et al., 2021). For example, the physical therapist might provide the following disclosure to their patient: "*I recommend we add volitional step training in your program. It has been shown that this type of training can be highly effective in reducing the risk of falls. The use of volitional step training is very beneficial for your training and performance*". Such information is honest, evidence based (see Okubo et al., 2017) and aims to engender a positive belief in the effectiveness of the treatment. Another important implication is that expectations appear to be relatively easy to manipulate and are therefore an effective target to induce changes in balance outcomes. From a theoretical standpoint, Self-Efficacy Theory suggests that social persuasion (most commonly verbal persuasion) can enhance self-efficacy, as long as it comes from a reputable source (Bandura et al., 1997). Collectively, there are considerations pertaining to the delivery and potential manipulation of beliefs/expectations that must be taken into account.

Limitations

This study provides a novel contribution to the literature, and a solid platform for future investigations to examine the effects of positive and negative expectancies on postural control. However, the findings of the present study should be interpreted in light of the study limitations. Firstly, our study was limited to healthy young adults. There is a reasonable basis for expectation that verbal suggestions might be even more potent in older adults and clinical patients. Thus, more studies are warranted to provide a more definitive view of placebo and nocebo effects on postural stability, controlling for age, sex and inclusion of more diverse groups. Secondly, the quiet standing balance task used here may not adequately stress the postural control system and represent a relatively small subset of our balance repertoire (Hill et al., 2020). Although we increased the level of difficulty of balance tasks by manipulation of stance (bipedal vs. unipedal), we did not measure other components of balance, including dynamic steady-state (i.e., walking), proactive (i.e., reaching), and reactive (i.e., responding to an unpredicted perturbation, such as slip or trip) abilities. Therefore, future studies should seek to use a more comprehensive battery of balance assessments to determine which balance functions are susceptible to positive and negative verbal suggestions. Although, the 36-item GASE has good internal consistency (Cronbach's α = 0.89; Rief et al., 2011), we only presented participants with the symptoms that we thought would occur relatively quickly after the pills were consumed. Therefore, we are unable to confirm the validity and reliability of the GASE as used here. Finally, the mechanistic basis of the increase (nocebo) and decrease (placebo) in postural sway remains to be elusive (and likely complicated). Neurophysiological investigations are needed to elucidate potential mechanisms underpinning the placebo- and nocebo-induced changes in balance. Additional examination of task specific anxiety, balance confidence, effort and motivation would also be quite valuable given their relevance to postural control.

Conclusion

The present investigation represents the first study to directly compare the influence of placebo and nocebo instructions on objective and subjective postural stability. To summarize, the present study shows that opposite verbal suggestions (i.e., placebo vs. nocebo) result in distinct postural outcomes. Specifically, placebo procedures result in a pronounced reduction in postural sway and increase in subjective ratings of postural stability, whilst nocebo procedures elicited a marked increase in postural sway and reduction in subjective stability. Positive and negative expectancies are a double-edged sword for postural control, and clinicians should consider how their instructions and feedback during therapeutic intervention may influence patients' expectation judgments.

Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Coventry University Ethics Committee. The patients/participants provided their written informed consent to participate in this study.

Author contributions

MH conceived and designed research. MH and KR conducted experiments. MH performed the analyses and wrote the manuscript. MH, AM, MP, MD, and TE revised the manuscript. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Sport-specific training induced adaptations in postural control and their relationship with athletic performance

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Effects of various exercise programs on postural balance control in athletes and their underlying physiological mechanisms have been extensively investigated. However, little is known regarding how challenging sport-specific conditions contribute to the improvement of body balance and to what extent these changes may be explained by sensorimotor and/or neuromuscular function adaptations. Analysis of the literature could provide useful information on the interpretation of changes in postural sway variables in response to long-term sport-specific training and their association with performance measures. Therefore, the aim of this scoping review was (1) to analyze the literature investigating postural control adaptations induced by sport-specific training and their relationship with measures of athletic performance, and (2) to identify gaps in the existing research and to propose suggestions for future studies. A literature search conducted with Scopus, Web of Science, MEDLINE and Cochrane Library was completed by Elsevier, SpringerLink and Google Scholar with no date restrictions. Overall, 126 articles were eligible for inclusion. However, the association between variables of postural balance control and measures of sport-specific performance was investigated in only 14 of the articles. A relationship between static and/or dynamic balance and criterion measures of athletic performance was revealed in shooting, archery, golf, baseball, ice-hockey, tennis, and snowboarding. This may be ascribed to improved ability of athletes to perform postural adjustments in highly balanced task demands. However, the extent to which sport-specific exercises contribute to their superior postural stability is unknown. Although there is a good deal of evidence supporting neurophysiological adaptations in postural balance control induced by body conditioning exercises, little effort has been made to explain balance adaptations induced by sport-specific exercises and their effects on athletic performance. While an enhancement in athletic performance is often attributed to an improvement of neuromuscular functions induced by sport-specific balance exercises, it can be equally well ascribed to their improvement by general body conditioning exercises. Therefore, the relevant experiments have yet to be conducted to investigate the relative contributions of each of these exercises to improving athletic performance.

KEYWORDS

athletes, neuromuscular functions, performance measures, postural sway variables, static and dynamic balance

1. Introduction

Long-term adaptations in postural balance control following a variety of exercises have been extensively researched (Behm et al., 2015; Hammami et al., 2016; Granacher and Behm, 2022). In particular, balance exercises lead to neurophysiological adaptations, which is beneficial for improvement of physical performance including posture, strength and jumping (Taube et al., 2008). The enhancement in motor skills following balance training can often be observed in increasing the rate of force development (Hrysomallis, 2011). Supraspinal adaptations within the central nervous system (CNS) are mainly responsible for improving functional parameters like balance skills, explosive strength or coordinative movement control (Taube, 2012). This suggests plasticity of the sensorimotor system, particularly the spinal and supraspinal structures (Taube et al., 2008). This plasticity of the spinal, corticospinal and cortical pathways is highly task specific (Taube et al., 2008).

Postural adaptation may be ascribed to improved ability to regulate the center of mass (CoM) movement more precisely with less effort and to perceive movement of the CoM more accurately via the use of proprioceptors. This is mostly true for task-oriented balance exercises based on visual feedback control of the CoM position (Zemková, 2014a). Although it is not possible to separate sensory and motor components of balance ability, practice can improve mainly proprioceptive functions. This is because the same receptors contribute to discriminating the position of ankle joints while regulating postural sway in the anteroposterior direction and transmissing the weight from one leg to the other while regulating postural sway in the mediolateral direction. This contributes to our understanding of physiological mechanisms underlying improvements of postural stability after body conditioning exercises (Zemková, 2010).

However, research to date has only marginally addressed their relevance with respect to performance in sports where static and/or dynamic balance plays an essential role. Postural adaptations are specific to the context in which the physical activity is practiced, so there is only slight transfer to non-experienced motor tasks (Paillard, 2017). However, the adaptation may occur as part of the interlimb relationship, particularly when the two legs do not display the same motor experience (Paillard, 2017). The most successful competitive athletes have more elaborate postural strategies compared with athletes at lower competition level (Paillard, 2017). They have the best postural performance both in ecological (specific postural conditions related to the sport practiced) and non-ecological (decontextualized postural conditions in relation to the sport practiced) postural conditions (Paillard, 2017). However, in non-ecological conditions, the postural tasks should be preferentially challenging or relatively close to the sport practice stance (Paillard, 2017). Though balance training improves performance of sport-related and postural control measures, it is unclear whether the effect of training would transfer to general functional enhancement (Yaggie and Campbell, 2006).

A novel approach is necessary to provide the basis for transfer of underlying sensorimotor processes of postural control to specific sport environments. There is a need to analyze the existing literature and elucidate whether the environment in which exercises are performed plays a role in an improvement of postural stability relevant to sport-specific skills. Provided that body balance is a complex adaptive system interacting with the environment in a functionally integrated manner, the interrelation between the sensorimotor processes of postural control and sport-specific tasks may be assumed. What is lacking is a review of the evidence investigating how these components are modified with training and subsequently how they influence performance in sports with high demands on body balance.

Therefore, the aim of this scoping review was to analyze literature investigating postural control adaptations induced by sport-specific training and their relationship with measures of athletic performance. This provided a basis for identifying gaps in the existing research and suggesting recommendations for future studies.

2. Methods

This paper is presented in a form of a scoping review (Arksey and O'Malley, 2005; Armstrong et al., 2011; Sucharew and Macaluso, 2019). It addresses two research questions: Do sport-specific balance exercises contribute to the enhancement of athletic performance in sports with high demands on postural stability? Is there a connection between improvements in postural control induced by sport-specific training and athletic performance?

A literature review was made to analyze existing research related to the relationship between variables of postural balance control and athletic performance measures, and related neurophysiological adaptations induced by sport-specific balance exercises. The search conducted with Scopus, Web of Science, MEDLINE and Cochrane Library was completed by Elsevier, SpringerLink and Google Scholar with no date restrictions. Articles in peer-reviewed journals were analyzed. However, references included in reviews were also manually searched to identify other relevant studies. If overlapping data were found in multiple articles, resulting from similar or the same research, those with the most recent publication date were considered for analysis. Books, theses, case reports, abstracts or articles published in conference proceedings were excluded. Incomplete articles and studies that did not include original research were also excluded. The inclusion criteria included research papers that sufficiently described participants, study design, and relevant measures. Studies written in English were preferred. Papers that failed to meet the eligibility criteria were excluded.

The search was focused on studies close to the main aim of this review. The key inclusion criterion was that (a) the training included specific balance exercises performed within a given sport, (b) variables of postural balance control were related to athletic performance measures, and (c) postural adaptations induced by sport-specific training were analyzed. However, only a small number of studies was revealed using this approach. Therefore, the search was widened to studies investigating adaptations in postural balance control and athletic performance induced by sport-specific as well as general body conditioning exercises. This helped us to identify gaps in the existing research and propose recommendations for further studies on this topic.

The search and appraisal of studies selected by inclusion and exclusion criteria was conducted by both authors. Some concerns were about the representativeness of samples, missing information related to balance exercises included in sport-specific training programs, imprecisely described variables of postural balance control and related performance measures and/or non-controlled compliance of experiments. Athletes of individual and team sports where postural stability plays an important role in their performance were considered a target population.

In the search strategy were included suggested sports combined with these terms: "balance exercises" AND "sport-specific exercise" AND "balance" AND "postural control" AND "athletes" AND "athletic performance" AND "postural sway variables" AND "performance measures" AND "neurophysiological adaptations," AND "neuromuscular functions," AND "sensorimotor functions," AND "physiological mechanisms." Further searches were performed using words from subheadings that specified other balance related exercises within particular sports, and other variables related to performance except for body balance (e.g., core strength and stability, muscle power and strength, etc.). Altogether 193 papers were found through database searching and other sources when these keywords were connected with particular sports. After an initial screening and assessing for eligibility, studies that did not meet the inclusion criteria were removed. Out of 126 articles included in this scoping review, only 14 investigated the association between variables of postural balance control and measures of sport-specific performance. **Figure 1** shows phases of the search process.

3. Results and discussion

In summary, studies in following sports were reviewed:

(a) shooting and air-rifle shooting (Niinimaa and McAvoy, 1983; Larue et al., 1989; Aalto et al., 1990; Era et al., 1996; Ball et al., 2003; Mononen et al., 2007; Ihalainen et al., 2016a,b; Spancken et al., 2021; Lang and Zhou, 2022a,b), air-pistol shooting (Ko et al., 2017, 2018), sharpshooting (Konttinen et al., 1998, 1999), small-bore shooting (Spancken et al., 2021);
(b) biathlon (Niinimaa and McAvoy, 1983; Larue et al., 1989; Ihalainen et al., 2018; Michalska et al., 2022);
(c) archery (Mason and Pelgrim, 1986; Stuart and Atha, 1990; Mohamed and Azhar, 2012; Spratford and Campbell, 2017; Musa et al., 2018; Serrien et al., 2018; Simsek et al., 2019; Wada and Takeda, 2020; Sarro et al., 2021; Vendrame et al., 2022);
(d) gymnastics (Bringoux et al., 2000; Vuillerme et al., 2001a,b; Aydin et al., 2002; Asseman et al., 2004, 2005, 2008; Davlin, 2004; Bressel et al., 2007; Carrick et al., 2007; Croix et al., 2010; Omorczyk et al., 2018; Marcolin et al., 2019; Busquets et al., 2021), rhythmic gymnastics (Kioumourtzoglou et al., 1997; Calavalle et al., 2008), acrobatic gymnastics (Sobera et al., 2019; Gómez-Landero et al., 2021; Opala-Berdzik et al., 2021), and artistic gymnastics (Puszczałowska-Lizis and Omorczyk, 2019);
(e) dancing (Perrin et al., 2002; Gerbino et al., 2007; Stins et al., 2009; Munzert et al., 2019; Nikolaidou et al., 2021), and ballet dancing (Schmit et al., 2005; Michalska et al., 2018; Thalassinos et al., 2018);
(f) golf (Lephart et al., 2007; Sell et al., 2007; Wells et al., 2009; Glofcheskie and Brown, 2017);
(g) baseball (Marsh et al., 2004; Butler et al., 2016; Liang et al., 2019);
(h) basketball (Perrin et al., 1991; Kioumourtzoglou et al., 1998; Bressel et al., 2007; Matsuda et al., 2008; Verhoeven and Newell, 2016; Halabchi et al., 2020; Glass and Ross, 2021; Makaracı et al., 2021);
(i) handball (Caballero et al., 2020);
(j) ice hockey (Behm et al., 2005; Kim et al., 2018; Rosker et al., 2021);
(k) soccer (Davlin, 2004; Paillard and Noé, 2006; Paillard et al., 2006; Bressel et al., 2007; Gerbino et al., 2007; Matsuda et al., 2008; Thorpe and Ebersole, 2008; Ben Moussa et al., 2012; Pau et al., 2015, 2018; Edis et al., 2016, 2017; Ozmen, 2016; Thalassinos et al., 2018; Jadczak et al., 2019a,b; Liang et al., 2019; Halabchi et al., 2020; Snyder and Cinelli, 2020; Zago et al., 2020; Glass and Ross, 2021; Śliwowski et al., 2021; González-Fernández et al., 2022; Scinicarelli et al., 2022);
(l) tennis (Caballero et al., 2021; Glass and Ross, 2021; Kozinc and Šarabon, 2021);
(m) volleyball (Agostini et al., 2013; Borzucka et al., 2020a,b; Makaracı et al., 2021);
(n) judo (Paillard et al., 2002, 2007b; Perrin et al., 2002);
(o) alpine skiing (Noé and Paillard, 2005; Kiers et al., 2022), cross country skiing (Glass and Ross, 2021), and snowboarding (Platzer et al., 2009a,b);
(p) surfing (Chapman et al., 2008; Paillard et al., 2011), canoeing and kayaking (Stambolieva et al., 2012), and paddle boarding (Schram et al., 2016);
(r) swimming (Davlin, 2004; Matsuda et al., 2008);
(s) others, such as track and field (Schmit et al., 2005), cascade ball juggling (Rodrigues et al., 2016), long distance running (Glofcheskie and Brown, 2017), running (Leightley et al., 2017), horseback riding (Olivier et al., 2019), pentathlon (Sadowska et al., 2019), and slacklining (Kodama et al., 2021).

3.1. Sport-specific training induced adaptations in postural balance control and performance measures

A review of the literature revealed that training under sport-specific conditions that include various balance exercises can lead to an improvement of postural stability. A recent analysis of postural sway variables in 936 athletes ranging from 6 to 47 years (shooters, football players, boxers, cross-country skiers, gymnasts, runners, team sport players, wrestlers, tennis players, alpine skiers, rowers, speed skaters and figure skaters) identified that practicing any kind of sport improves bipedal balance (Andreeva et al., 2021). However, it mostly depends on their age, and partly on their level of performance, sex, and shoe features (Andreeva et al., 2020). Usually, balance performance is associated with the level of competition, with better balance in more proficient athletes (Hrysomallis, 2011).

Most studies compared postural sway variables in athletes of different sports and levels of expertise, or physically active individuals with a control group of sedentary individuals. In general, athletes are superior to non-athletes in balance performance (Davlin, 2004). More specifically, athletes of both team and individual sports demonstrate better body balance over the group of non-athletes (Mocanu et al., 2022). Lateral CoP deviation is also lower in individuals with moderate than low physical activities (Onofrei and Amaricai, 2022). However, static balance in collegiate athletes with a sport background does not differ from their multisport counterparts (Chou et al., 2022).

Further studies have been conducted to investigate acute (Zemková and Hamar, 2014; Zemková, 2022) and adaptive changes in postural balance control. A recent scoping review revealed that neuromuscular control of core and postural stability contributes to more effective functional movements in a given sport (Zemková and Zapletalová, 2022). Including sport-specific and general core

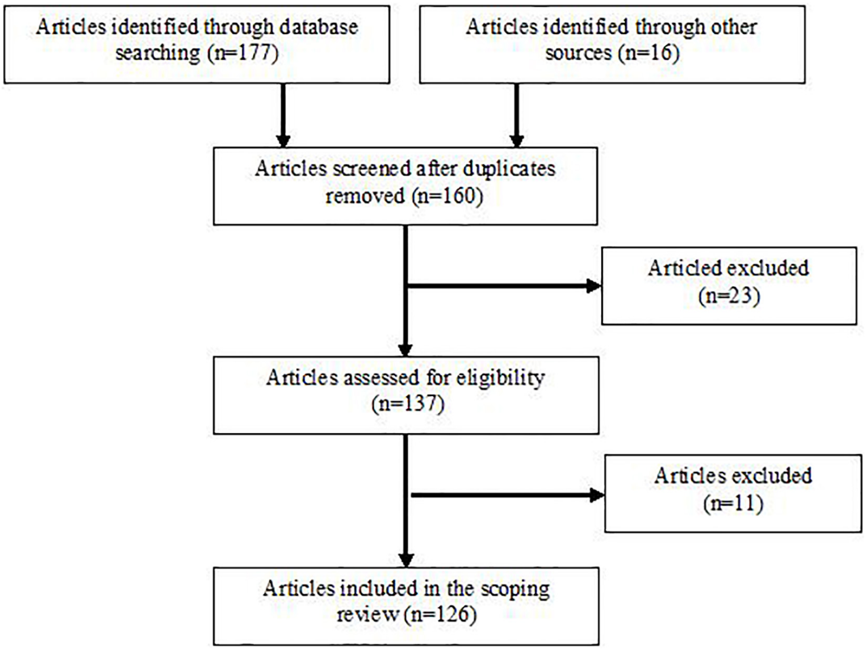

FIGURE 1
Flow chart illustrating phases of the literature search and study selection.

and balance exercises into training programs improves postural stability, strength and endurance of the back muscles (Zemková and Zapletalová, 2022). Although the ankle is the most significant predictor of the magnitude of body sway, the trunk is the second most important element during specific postural tasks (Duchene et al., 2021). Postural control of the trunk expressed by lower CoP movement in a seated position is better in long distance runners and golfers than in controls (Glofcheskie and Brown, 2017). Lumbar spine angular displacement is lower, muscle activation amplitudes is higher, and trunk muscle activation onset is faster in response to unexpected perturbations of the trunk in athletes than in controls (Glofcheskie and Brown, 2017). Variable and absolute error in the trunk repositioning task is lower in golfers than in runners and controls, indicating higher proprioceptive ability (Glofcheskie and Brown, 2017). This assumes that an association exits between proprioception, postural and neuromuscular control in athletes, and that it discriminates between those with different training background (Glofcheskie and Brown, 2017).

Balance training can improve some sport-related performance measures, however its transfer to general functional enhancement is unclear (Yaggie and Campbell, 2006). Balance adaptations are specific to practiced physical activities (Paillard, 2017). Successful competitive athletes have the best postural stability in both specific and decontextualized balance conditions in relation to sports practiced (Paillard, 2019). There is only small transfer to a non-experienced task (Paillard, 2017). The level of performance is often associated with proprioceptive acuity in elite athletes (Han et al., 2015). However, the relationship between sport-specific training and improvements in proprioceptive acuity is limited by biologically determined factors (Han et al., 2015). An investigation of the effects of sport-specific training on CoP velocity and displacement in collegiate cross country, basketball, and soccer athletes indicates that between-group differences may be related to sensorimotor adaptations in a given sport, including proprioceptive acuity, strength and power of lower limbs, and efficiency of vestibular system (Glass and Ross, 2021). The involvement of muscle synergies or cutaneous feedback are further potential mechanisms (Glass and Ross, 2021). For instance, the use of vestibular and proprioceptive cues in contact sports athletes leads to better postural stability (Liang et al., 2019). Furthermore, postural strategies are better in athletes at higher than lower competition level (Paillard, 2019). Dynamic balance strategies may be influenced by neurocognitive performance in athletes (Porter et al., 2022). Different approaches may be used when performing difficult balance tasks with strategies related to higher anteroposterior and lower vertical acceleration in higher than lower neurocognitive performers (Porter et al., 2022).

In comparison with a number of studies dealing with neurophysiological adaptations in postural control induced by general body conditioning exercises, scarce research has been conducted to investigate adaptive changes in postural balance control in response to sport-specific balance exercises and their effects on athletic performance.

3.1.1. Shooting

Static balance is important for performance in shooters. Both postural sway and rifle stability are different in national- and elite-level air-rifle and small-bore shooters (Spancken et al., 2021). While aiming accuracy, rifle stability, and aiming time influence the shot score in national-level air-rifle athletes, postural sway does not affect the shot score in these athletes (Spancken et al., 2021). Sway velocity in shooters is reduced under both visual and non-visual conditions when using competition clothing (Aalto et al., 1990). Shooters use a higher amount of proprioceptive and vestibular cues for stabilizing their posture (Aalto et al., 1990). This posture stabilization is better in top-level male shooters compared to national-level male and top-level female shooters, whose stability is better than in naive shooters (Era et al., 1996). In particular, the ability to stabilize posture prior to the shot is better in experienced shooters (Era et al., 1996). However, worse stabilization of posture does not contribute to bad results among the highly trained shooters (Era et al., 1996).

In addition to the CoP movement, also arm joints and the pistol end point motions are lower in expert than in novice shooters (Ko et al., 2017, 2018). Kinematic variables are reduced in a lower dimensional functional unit by skill acquisition, so pistol-aiming is characterized by upper limb and posture performance (Ko et al., 2017). The coordination of pistol and posture motion is more variable in novice shooters, whereas it is more consistent in the skilled group (Ko et al., 2018). Dispersion and complexity is reduced in the skilled arm-pistol motion (Ko et al., 2018). Thus, rifle motion is an important factor of advanced performance in shooting (Konttinen et al., 1998). While pre-elite athletes rely more on the visual-spatial processing, elite athletes focus more on stabilization of rifle position by using psychomotor regulation (Konttinen et al., 1998). There is a relationship of postural sway velocity and amplitude with changes in the concomitant brain slow potential, however it depends on the expertise of shooters (Konttinen et al., 1999). Reduced postural sway amplitude coincides with decreased frontal positivity in elite shooters, whereas postural anteroposterior sway velocity and amplitude are characterized by the central negativity lateralization in the non-elite shooters (Konttinen et al., 1999).

3.1.2. Biathlon

The strategy of biathletes is different from the one used by rifle shooters due to adaptive changes of their respective specializations (Larue et al., 1989). Postural stability is important during standing and aiming an air rifle at both rest and during a cross-country ski racing. Anteroposterior sway movement during standing at rest is about twice that of movement in a mediolateral direction (Niinimaa and McAvoy, 1983). However, lateral sway movement increases with aiming, and both sway movements are similar after exercise (Niinimaa and McAvoy, 1983). Vertical hold stability and cleanness of triggering are the most important factors influencing shooting performance at rest as well as during competition shots (Ihalainen et al., 2018). Postural stability in shooting direction is associated with vertical hold stability and cleanness of triggering (Ihalainen et al., 2018). Cleanness of triggering, aiming accuracy, vertical and horizontal hold stability decrease from resting to competition simulation shots, accompanied by a decrease in postural balance (Ihalainen et al., 2018). Therefore, biathletes should focus on vertical hold stability and cleanness of triggering to improve their shooting performance (Ihalainen et al., 2018). Better postural stability in shooting direction can contribute to the improvement of these technical components (Ihalainen et al., 2018). National- and junior-team biathletes differ only in the percentage of hits during resting shots and postural stability on the left leg in shooting direction during competition shots, but intensive exercise affects these technical components in both groups of shooters (Ihalainen et al., 2018). Different motor control strategy is used in experienced biathletes, beginners and controls, characterized by better postural stability during aiming at the target and shooting (Michalska et al., 2022). Body sway is significantly smaller in the position shooters than in those with less than four months of training (Niinimaa and McAvoy, 1983).

3.1.3. Archery

Maximum sway speed, draw force, and clicker reaction time are important factors of shot performance in elite recurve archers (Spratford and Campbell, 2017). More specifically, reducing postural sway during the release phase increases shooting performance of skilled archery athletes (Mohamed and Azhar, 2012). In addition to reduced postural sway speed, also greater bow draw force and reduced clicker reaction time post-arrow release contribute to better scoring shots (Spratford and Campbell, 2017). The ability to control the bow and postural sway movements together with higher activation of muscle extensor digitorum is better in medalist and elite archers (Musa et al., 2018). Postural sway during arrow shooting is lower in elite archers compared to those of mid-level and beginners (Simsek et al., 2019). Expert archers tend to maximize postural stability and develop personal strategies of muscular activation and time management (Vendrame et al., 2022). There is a considerable variability in the precision with which the positions of head, elbow and bow at the moment of loose are replicated by archers of similar skills level (Stuart and Atha, 1990). It seems that precise postural consistency is not the primary feature distinguishing between the performance of archers at the higher skill levels (Stuart and Atha, 1990). Serrien et al. (2018) suggets that it is not necessary for elite archers to minimize the movements of all degrees of freedom during aiming, but rather that the structure of variability of the redundant kinematic chain is exploited so that the relevant performance variable (orientation of the arrow) is stabilized. Taking into account better stability against visual disturbance in archers than ball game players and untrained subjects, one may assume that they rely on proprioceptive inputs to maintain balance (Wada and Takeda, 2020). Their training re-weights sensorimotor dominance from vision to proprioception for posture regulation to increase shooting accuracy (Wada and Takeda, 2020).

3.1.4. Gymnastics

Expertise in gymnastics enhances postural stability only in situations where it is practised, i.e., while standing on one leg with eyes open (Asseman et al., 2008). The CoP excursion during one-legged stance is lower in base than top gymnasts, whereas their values are lower in mid- than early adolescent gymnasts, regardless of the task (Gómez-Landero et al., 2021). However, the level of expertise does not have effect on bipedal postural stability (Marcolin et al., 2019). Therefore, sport-specific tasks are more selective in representing the expertise level in gymnasts (Marcolin et al., 2019).

Gymnasts have greater postural stability than non-gymnasts (Carrick et al., 2007). Specifically, decreasing CoP displacements by reinsertion of proprioceptive information is better in gymnasts than non-gymnasts (Vuillerme et al., 2001b). Greater gymnastic skills are characterized by more stable, less regular, and less variable acceleration time-series (Lamoth et al., 2009). Gymnastic training improves the ability to change the frame of reference (Croix et al., 2010). The rod-and-frame test results correlate with postural stability, and experts are less field dependent than non-experts (Croix et al., 2010). The remaining sensory modalities are used more efficiently under eyes closed conditions in expert gymnasts (Croix et al., 2010). They use remaining sensory modalities mainly when standing on unstable surfaces with eyes closed (Vuillerme et al., 2001a). Specific training in gymnasts improves the efficiency of the integration process leading to the perception of the body orientation in space (Bringoux et al., 2000). The relevance of otolithic and/or interoceptive inputs increases with their expertise (Bringoux et al., 2000).

Age influences static balance in acrobatic gymnastics (Gómez-Landero et al., 2021). Proprioceptive reweighting processes can be improved by gymnastic training during childhood, leading to similar control and coordination of posture as adults (Busquets et al., 2021).

More specifically, better anteroposterior postural stability with eyes open is related to greater age, body mass and height, biological maturity, and training experience in artistic gymnasts, whereas better postural stability regardless of visual conditions is related to higher body mass and BMI percentiles in acrobatic gymnasts (Opala-Berdzik et al., 2021).

Static and dynamic balance, and whole-body coordination are better in elite rhythmic gymnasts than in controls (Kioumourtzoglou et al., 1997). Static balance, hand-eye coordination, and an anticipation of coincidence (hand-eye coordination and its anticipation on visual accuracy) are better in older than younger (13–15, 11–12, 9–10 years) elite rhythmic gymnasts (Kioumourtzoglou et al., 1997). Moreover, strategies in lateral directions during simple postural tasks are better in rhythmic gymnasts than in students, indicating that their training has a direct impact on mediolateral bipedal balance (Calavalle et al., 2008). Their training also improves postural stability, sense of ankle joint position, and increases muscle tone (Aydin et al., 2002). Gymnasts learn and perform better in new motor routines compared to those with high fatigability ratios and lower adaptability scores (Carrick et al., 2007).

The stability of standing and handstand is differentiated by the sport advancement level so that the control of both body positions is better in seniors than in juniors (Omorczyk et al., 2018). In particular, more experienced gymnasts have better postural performance during the handstand than non-experts (Croix et al., 2010). Exerting force on a floor surface helps to minimize body sway, however less experienced athletes are not able to do that even after long-term training (Sobera et al., 2019). More skilled performance is demonstrated by adaptation of reactive rather than anticipatory strategies. Mechanical advantages in mediolateral balance while standing on wider base of support are observed in more skilled athletes than in less (Wyatt et al., 2021). Controlling the mediolateral CoP movement with eyes open is not essential for body stability in the frontal plane in seniors practicing gymnastics (Puszczałowska-Lizis and Omorczyk, 2019). The ratios of CoP velocity with eyes closed on eyes open are similar in the bipedal and handstand postures, suggesting that the specificity or difficulty of the posture is not directly related to the effect of vision removal (Asseman et al., 2005). However, stability indices in standing are not related to those of the handstand, indicating that stability in a standing position does not predict handstand performance (Omorczyk et al., 2018). On the contrary, handstand stability is not transferable to upright standing postures among elite gymnasts (Asseman et al., 2004).

3.1.5. Dancing

Postural sway is more stationary (lower absolute trend), less stable (lower maxline), less regular (lower recurrence), and less complex (lower entropy) in ballet dancers than in track athletes (Schmit et al., 2005). Dancers' stability is also better than in soccer players (Gerbino et al., 2007). Both professional dancers and high-level judoists perform better than controls, which indicates that their training has positive effects on sensorimotor adaptabilities (Perrin et al., 2002). Higher trembling component during both quiet and inclined standing indicates better ability to maintain balance under unstable conditions in professional ballet dancers than in non-trainees (Michalska et al., 2018). An expert advantage on postural stability is observed in specific dance tasks but not in static everyday tasks (Munzert et al., 2019).

However, postural stability may also play a role in performance of less trained individuals, such as entry level university dancers (Misegades et al., 2020) or older dancers (Nikolaidou et al., 2021). The superior balance performance in dancers over 70 years old most likely results from coordinated intersegmental movements, including between-leg alternations during dancing (Nikolaidou et al., 2021).

3.1.6. Baseball, basketball, handball, hockey, tennis, volleyball

Sport-specific balance in team sports athletes may depend on their expertise and level of performance. For example, dynamic balance depends on the competition level of baseball players (Butler et al., 2016).

Dynamic balance is lower and hand coordination is higher in elite male basketball players than in controls. They are also better on prediction measures, selective attention, and memory-retention (Kioumourtzoglou et al., 1998). The coordination of ball release and postural stability is important for success in shooting changes, reflecting the level of athletic skills (Verhoeven and Newell, 2016). Interestingly, the best basketball player has significant results in static balance with no differences in eyes open and eyes closed conditions (Perrin et al., 1991).

Better balance in experienced handball players is associated with the maturation of the motor system rather than performance level (Caballero et al., 2020). Slower CoP velocity during balance tasks and less irregular movements are in players who throw with less accuracy, whereas less auto-correlated and more irregular CoP movements are in players who throw faster (Caballero et al., 2020). Less dependence on previous behavior (lower regularity and long-range auto-correlation, respectively) has been considered as higher flexibility to perform motion adjustments to reduce motor output error (Wang and Yang, 2012). This means that expert players display a more exploratory behavior when performing balance tasks, thus have a greater adaptive capacity of the CNS over longer time scales (Barbado Murillo et al., 2017). Similarly, high postural strategies used for readjustments of unexpected perturbations are found in elite ice hockey players (Kim et al., 2018). Their muscle synergies display low co-activation strategy of antagonists and agonists in the neck and ankle (Kim et al., 2018). Upper body loading and sport-specific posture in elite ice hockey players induce adaptations in neck proprioception (Rosker et al., 2021). Cervical spine afferent input is esential for maintaining unilateral balance in hockey players (Rosker et al., 2021). Therefore, neck kinaesthesia, in addition to postural control, may influence oculomotor performance, which is important for initiating eye movement changes (Rosker et al., 2021).

Static balance is only slightly different on the non-preferred and preferred leg determined by the manipulation task in the form of kicking a ball in highly-trained tennis players, whereas there are no differences when the classification is based on the preference for performing a single-leg jump (Kozinc and Šarabon, 2021). Similarly, there are no between-leg differences in dynamic balance for the landing task (Kozinc and Šarabon, 2021).

The model of sensory integration in postural balance control is different in volleyball players and non-athlete controls (Agostini et al., 2013). Between-group postural stability is different with eyes open but not with eyes closed, which may be ascribed to better dynamic visual acuity in athletes because static refractive errors are corrected in both groups (Agostini et al., 2013).

Parallel stance and dominant leg postural sway is better in deaf volleyball than basketball players, whereas there are no significant between-group differences in non-dominant leg postural sway (Makaracı et al., 2021). Postural regulation in top level male volleyball players is more precise and less vulnerable to external disturbances than in non-athletes, which support optimal timing and precision of actions (Borzucka et al., 2020a). They have better capacity to use postural strategies for maintenance of balance and reduce the use of proprioception for performing challenging postural and motor tasks (Borzucka et al., 2020a). Volleyball players develop a unique posture control resulting from the motor demands of this sport (Borzucka et al., 2020b).

3.1.7. Soccer

Dynamic balance is better in gymnasts than in soccer players and swimmers, whereas there are no differences between two remaining groups of athletes (Davlin, 2004). However, Bressel et al. (2007) found that dynamic balance is not different in soccer players and gymnasts. Differences in these findings may be mainly ascribed to different tests used. While in the first case, a stabilometer, which requires participants to continuously adjust posture to maintain an unstable platform in the horizontal position, was used (Davlin, 2004), in the second case, participants performed multidirectional maximal single-leg reaches from a unilateral base of support using the Star Excursion Balance Test (Bressel et al., 2007). Furthermore, soccer players have superior postural control compared to baseball players and untrained students (Liang et al., 2019). Their ability to maintain one-legged balance is also better than in swimmers, basketball players, and non-athletes (Matsuda et al., 2008). Basketball players present inferior dynamic balance in comparison with soccer players and inferior static balance in comparison with gymnasts (Bressel et al., 2007). However, dancers perform better than soccer players (Gerbino et al., 2007).

Postural parameters improve with age until zero maturity offset is achieved (Zago et al., 2020). It seems that these parameters are most stable in developing soccer players (Zago et al., 2020). Variance in multidirectional speed performance in young soccer players may be predicted from dynamic balance performance and chronological age (Scinicarelli et al., 2022). There is a strong association of multidirectional speed performance with dynamic balance performance of the dominant side, whereas there is a small relationship with limb symmetry index (Scinicarelli et al., 2022).

Postural strategy and performance is influenced by the competition level in soccer players. Playing experience also affects postural measures and strategies in test conditions specific to playing soccer (Paillard et al., 2006). Moreover, static balance varies in elite soccer players playing at different positions, i.e., it is better in midfield players than those in other positions (Jadczak et al., 2019b), whereas centre-backs are worse than wingers and forwards (González-Fernández et al., 2022). Postural stability, postural strategy, and the use proprioception and vision information is different in national and regional soccer players (Paillard et al., 2006). Postural stability is less disturbed when manipulating sensory information in the high-level than in the regional-level players (Paillard et al., 2007a). The internal model of verticality is better in the high-level than in the regional-level players. Those with better postural stability are less disturbed by sensorial manipulation than the others (Paillard et al., 2007a). Proprioceptive executive control is improved by soccer-specific training, resulting in better correlation between CoP and single-support balance during a dynamic visuomotor reaching task of lower limbs (Snyder and Cinelli, 2020). Visual contribution is lower in professional than amateur players (Ben Moussa et al., 2012). Professional soccer players are less dependent on vision when controlling their posture, so vision can be dedicated to treat with information during the game (Paillard and Noé, 2006).

Professional soccer players have also greater postural stability on the non-dominant leg (Jadczak et al., 2019b). Their balance control is widely influenced by concentric isokinetic strength (peak torque of quadriceps and peak torque of hamstrings at high angular velocity), particularly in the supporting, non-dominant leg (Śliwowski et al., 2021). Better balance in young national-level soccer players is characterized by more efficient and faster stabilization after a forward jump, whereas the unipedal stance test is not able to reveal differences in postural control associated with a combination of physical and technical skills (Pau et al., 2018). Postural balance control is among the most important factors that influence performance of technical skills under the pressure and unexpected changing situations in trained amateur soccer players (Edis et al., 2016). Therefore, a combined training involving soccer-specific and balance exercises can significantly contribute to their performance (Edis et al., 2017). The higher their sport level, the better their balance. This may indirectly contribute to successful performance in any game situations (Jadczak et al., 2019a).

In practice, both static and dynamic balance tests should be performed in soccer players because balance variables in these two conditions are not related (Pau et al., 2015). For instance, Balance Error Scoring System including static postures reflects deficits in postural control better than dynamic balance tests in professional football and basketball players (Halabchi et al., 2020). The measures from the Star Excursion Balance Test may not reflect the balance performance in well-trained football and basketball players who have a better balance when performing sport-related skills (Halabchi et al., 2020). However, soccer players reach significantly farther than the non-soccer athletes, suggesting that the Star Excursion Balance Test may be sensitive to training status and/or sport-related adaptations (Thorpe and Ebersole, 2008). This a unilateral, functional joint-stability task may reflect their adaptation to single-leg exercises and other sport-related skills, such as standing on one leg while kicking the ball. Values of this test are not associated with those of the side bridge, trunk extension and flexion tests, which indicates that core stability does not contribute significantly to dynamic balance (Ozmen, 2016).

3.1.8. Golf

Balance, core strength and stability, peripheral muscle strength, and flexibility correlate with performance in golf (Wells et al., 2009). These abilities including balance, flexibility and strength are improved through a specific exercise program in golf, which results in higher upper-torso axial rotational velocity, and consequently also in higher club head and ball velocity, as well as driving distance (Lephart et al., 2007). Balance, torso, shoulder and hip strength and flexibility are better in golfers with handicap (HCP) of < 0 than in those with HCP of 10–20 (Sell et al., 2007).

3.1.9. Combat sports (judo)

Practice of high-skill activities that include proprioceptive afferences enhances both postural stability and performance (Perrin et al., 2002). Dancers and judoists perform better than controls because their training improves sensorimotor functions (Perrin et al., 2002). Specific postural adaptations are also induced by different movements performed on one or two legs (a tokui-waza in monopodal and bipodal stance) in competition-level judoists (Paillard et al., 2007b). However, static balance does not differ significantly between regional and national and international level judoists (Paillard et al., 2002).

3.1.10. Water sports (canoeing, kayaking, paddle boarding, surfing)

The model of sensory integration is different in young kayakers and canoeists than in non-athletes as a result of their sport specializations, which may be attributed to re-adaptation deficit after disembarking to stable surface with diminished sensitivity of the vision and vestibular systems (Stambolieva et al., 2012).

Stand-up paddle boarding athletes have increased static and dynamic balance, aerobic and anaerobic fitness, and isometric trunk endurance (Schram et al., 2016).

There is an association of postural stability with the competition level of surfers (Paillard et al., 2011). The sensorimotor dominance for maintenance of balance can be shifted from vision to proprioception in expert surfers (Paillard et al., 2011). However, standard postural sway variables are not able to indicate whether surfing expertise facilitates balance adaptations (Chapman et al., 2008).

3.1.11. Winter sports (alpine skiing)

Relative dynamic postural stability index improves annually in competitive youth skiers (Kiers et al., 2022). However, age and biological maturation correlate with absolute but not with relative values of dynamic postural stability index (Kiers et al., 2022). Furthermore, postural stability is similar when tested in ski boots and it is similarly influenced by the absence of visual information in regional and national level skiers (Noé and Paillard, 2005). However, postural stability without ski boots is better in regional than national level skiers (Noé and Paillard, 2005). Such an inferior postural stability may be attributed to repetitive wearing of ski boots during a long-term training, which affects balance by restricting the ROM of the ankle-foot complex (Noé and Paillard, 2005).

3.1.12. Other sports

Practicing many other sports may contribute to the improvement of balance, however a significant relationship with athletic performance has been rarely documented. For instance, proprioceptive functions of posture and postural muscle tone during bipodal dynamic perturbations are developed by horseback riding (Olivier et al., 2019). Refined postural stability is also associated with expertise in cascade juggling (Rodrigues et al., 2016). Stance stability is better in pentathletes than in untrained individuals and they are also less dependent on vision (Sadowska et al., 2019). Experts in slacklining tend have a more antiphase coordination pattern and coordinate their hands more sustainably than novices (Kodama et al., 2021). Postural control declines in master runners, however they may benefit from balance exercises (Leightley et al., 2017).

3.2. The relationship between postural balance control and athletic performance

Analysis of the literature revealed that postural balance control is a key determinant of performance in several sports (Zemková, 2014b). However, postural sway variables have been found to be associated with only a few measures of sport-specific performance (**Table 1**).

Static bipedal balance is associated with shooting accuracy in rifle shooters (Ball et al., 2003) but only at the inter-individual level (Mononen et al., 2007). While inter-individual analyses revealed that postural stability negatively correlates with aiming accuracy and shooting score and positively correlates with triggering and hold stability, intra-individual analyses showed the relationship between postural stability and performance, hold and triggering stability, and aiming accuracy (Lang and Zhou, 2022a). Thus, aiming and holding are the essential factors of shooting performance, followed by the stability of triggering (Lang and Zhou, 2022b). The aiming accuracy, stability of hold, cleanness and timing of triggering are key predictors of shooting performance, whereas postural stability plays a very small role (Ihalainen et al., 2016a). The effect is higher through stable holding, which correlates with postural stability (Ihalainen et al., 2016a). Cleanness of triggering and stability of hold are also associated with competition performances in elite air-rifle shooters (Ihalainen et al., 2016b). The stability of hold is also associated with postural balance control in cross-shooting direction, whilst in shooting direction it is related to cleanness of triggering (Ihalainen et al., 2016b). A relationship between body sway and aim point fluctuation means that aim point fluctuation increases and performance decreases when body sway increases (Ball et al., 2003). Specifically, the CoP velocity in mediolateral direction and the aiming point deviation are independent variables explaining the shooting score (Mononen et al., 2007).

Similarly, postural sway displacement in the X direction prior to the arrow release is associated with performance score in archers (Mason and Pelgrim, 1986). There is a significant but low inverse relationship between the total excursion of the archer's CoP in the interval one second prior to arrow release and shooting performance (Mason and Pelgrim, 1986). This association of the total CoP excursion with the criterion variable of shooting performance is stronger in junior than senior archers (Mason and Pelgrim, 1986). That is, shooting performance in less experienced archers is better when their postural sway movement decreases (Mason and Pelgrim, 1986). Therefore, the synchronization of body and bow sway is important for shot accuracy in recurve archers, which may be corroborated by a significant correlation between the CoP and bow displacement (Sarro et al., 2021).

Furthermore, static unipedal balance correlates with putt distance after a chip shot and greens in regulation in elite golfers, suggesting that standing on uneven ground and weight shift during the golf swing may require good postural balance control (Wells et al., 2009). Unilateral balance also correlates with pitch velocity but not with pitching error in baseball players (Marsh et al., 2004).

An investigation of associations of balance with tennis expertise and performance revealed a lack of correlations, suggesting that postural stability measured in non-specific conditions is not a key factor of performance in the tennis serve (Caballero et al., 2021). However, sport experience in expert tennis players leads to better ability to perform postural adjustments (Caballero et al., 2021). A non-linear analysis is able to identify small postural adaptations induced by sport practice while the CoP dynamics discriminates sport expertise (Caballero et al., 2021).

With regard to dynamic balance, it is associated with maximum skating speed in young ice hockey players (Behm et al., 2005). Unipedal dynamic balance also correlates with starting speed during a simulated luge start in snowboarders (Platzer et al., 2009b), but not, however, with their ranking points (Platzer et al., 2009a).

TABLE 1 An overview of studies dealing with the relationship between postural sway variables and measures of athletic performance.

Authors (year)	Study objective	Participants	Postural sway variables	Athletic performance measures	The relationship between postural sway variables and measures of athletic performance
Mason and Pelgrim (1986)	To quantify body movements of the archers and to identify their relationships with shooting accuracy	Austrian senior and junior archers	CoP movement in Y and X directions before and after arrow release	Arrow shooting accuracy	A significant correlation (-0.30) between shooting performance and the total excursion of the archer's CoP in the interval one second prior to arrow release; The total CoP excursion correlates with the criterion variable of shooting performance in junior (-0.51) but not in senior archers (-0.24)
Ball et al. (2003)	To examine the relationships between body sway, aim point fluctuation and performance in rifle shooting on an inter- and intra-individual basis	Six elite shooters	Body sway parameters quantified for the time periods 5 s to shot, 3 s to shot and 1 s to shot	Four aim point fluctuation parameters quantified for the time periods 5 s to shot, 3 s to shot and 1 s to shot	Body sway is related to aim point fluctuation; These relationships are specific to the individual, with the strength of association, parameters of importance and time period of importance different for different shooters
Marsh et al. (2004)	To examine the relationship between balance and pitching error in college baseball pitchers	Sixteen college baseball pitchers, 9 National Association of Intercollegiate Athletics and 7 National Collegiate Athletic Association, Division III	Average sway velocity during dominant leg unilateral stance with eyes open and eyes closed using the Balance Master System 7.04; Sensory organization testing on the SMART EquiTest System providing information on the use of the somatosensory, visual, and vestibular inputs	Pitching error assessed with a high-speed video camera recorder; Pitch velocity measured using a JUGS radar gun	A significant negative correlation between sensory organization test 5 and pitching error (-0.50) and between sensory organization test 5/1 and pitching error (-0.50); A positive correlation between unilateral stance eyes closed and a pitch velocity (0.52); No significant correlation between unilateral stance eyes open and pitching error (-0.24) or unilateral stance eyes closed and pitching error (-0.29)
Behm et al. (2005)	To determine the relationship between specific performance measures and hockey skating speed	Thirty competitive secondary school and junior hockey players	Balance ratio (wobble board test)	Maximum skating speed (a 40-yd (36.9-m) sprint)	Significant correlations between skating performance and the sprint and balance tests; Significant correlations between balance and players under the age of 19 years (-0.65) but not those over 19 years old (-0.28)
Mononen et al. (2007)	To examine the relationships between shooting accuracy and shooters' behavioral performance, i.e., postural balance and gun barrel stability, among novice rifle shooters in intra- and inter-individual levels	Fifty-eight shooters	Postural balance assessed in terms of anteroposterior [VEL(AP)]and mediolateral [VEL(ML)] sway velocity of the CoP movement	Rifle stability assessed in terms of horizontal [DEV(H)] and vertical [DEV(V)] deviation of the aiming point	The shooting accuracy is related to postural balance and rifle stability, but only at the inter-individual level; The correlation coefficients between shooting score and behavioral performance variables range from -0.29 to -0.45; The VEL(ML) and the DEV(H) as independent variables account for 26% of the variance in the shooting score; Postural balance is related to the shooting accuracy both directly and indirectly through rifle stability
Platzer et al. (2009a)	To assemble and evaluate a battery of tests for the snowboard disciplines parallel, snowboard cross (SBX), big air, and half-pipe (HP)	Thirty-seven competitive snowboarders	Dynamic unipedal balance measured by the Biodex Balance System	World Cup & International Federation of Skiing points	Dynamic unipedal balance is not associated with snowboarders' ranking points
Platzer et al. (2009b)	To evaluate the influence of different physiological factors on the luge start and identify an appropriate physiological test battery	Thirteen male members of the Austrian national luge team	Dynamic unipedal balance measured by the Biodex Balance System	Starting speed measured by the luge start simulator	Dynamic unipedal balance is associated with end speed (0.590) but not with maximal speed
Wells et al. (2009)	To identify physiological correlates of golf performance in elite golfers under laboratory and tournament conditions	Elite golfers	Timed unipedal stance	Ball speed and distance, average score, greens in regulation, short game measures, and putting accuracy	Static balance is associated with greens in regulation (-0.43) and average putt distance after a chip shot (0.50)

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TABLE 1 (Continued)

Authors (year)	Study objective	Participants	Postural sway variables	Athletic performance measures	The relationship between postural sway variables and measures of athletic performance
Ihalainen et al. (2016a)	To identify the most important factors determining performance in elite-level air rifle shooting technique	Elite-level air rifle shooters		Six components in the air rifle shooting technique: aiming time, stability of hold, measurement time, cleanness of triggering, aiming accuracy, and timing of triggering	Stability of hold, cleanness of triggering, aiming accuracy, and timing of triggering are the most important predictors of shooting performance, accounting for 81% of the variance in shooting score; The direct effect of postural balance on performance is small, accounting for less than 1% of the variance in shooting score; The effect can be greater through a more stable holding ability, to which postural balance is significantly correlated
Ihalainen et al. (2016b)	To describe the long-term changes in shooting technique in relation to competition performances in elite air-rifle shooters	Seventeen elite air rifle shooters	Postural-balance variables measured with force platform	Shooting score and aiming-point-trajectory variables obtained with an optoelectronic shooting device; Shooters' competition results collected from all international and national competitions during the 3-y period	Seasonal mean test results in stability of hold (-0.70) and cleanness of triggering (-0.75) are related to competition performances; Changes in stability of hold (-0.61) and cleanness of triggering (-0.39) are related to the changes in competition performances; Postural balance in shooting direction is more related to cleanness of triggering (0.57), whereas balance in cross-shooting direction is more related to stability of hold (0.70)
Caballero et al. (2021)	To assess the relationship between balance and tennis performance using linear and non-linear parameters through 1) the comparison of tennis players of different ages and levels of expertise, and 2) analyzing the relationship between balance and tennis serving speed and accuracy	One hundred and six recreational and expert male tennis players	Temporal dynamics of postural control during a balance task on an unstable surface analyzed through the mean velocity and the detrended fluctuation analysis (DFAV) of the CoP	Tennis serve performance quantified by measuring accuracy and speed	The CoP showed a reduction of auto-correlated variability (reflected by DFAV) with age but mainly in expert players; The CoP dynamics is the only balance parameter discriminating sport expertise and it is related to age; Sport experience in expert tennis players induces balance adaptations characterized by a higher ability to perform postural adjustments; The lack of correlations suggests that balance, measured with scattering variables, in a non-specific task is not a main determinant of sport performance in tennis serve
Lang and Zhou (2022a)	To examine the relationships between postural balance, aiming technique and shooting score among elite rifle shooters at an intra- and inter-individual level	Twelve elite athletes belonging to China national team	Postural balance variables measured using footscan 1.0 force platform	Aiming technique parameters measured using a SCATT MX-02 optoelectronic training device	Inter-individual analyses: Postural balance is negatively correlated with shooting score (-0.697) and aiming accuracy (-0.810); A positive correlation of postural balance with the stability of hold (0.923) and stability of triggering (0.564); Intra-individual analyses: A significant correlation between postural balance and performance, aiming accuracy and stability of hold; Postural balance is related to the stability of triggering; Postural balance is not significant with aiming time on an intra- and inter-individual basis
Sarro et al. (2021)	To investigate the relationship between bow stability and postural control in recurve archery according to shooting performance	Eight archers	The CoP position of the archer (the point of application of the resultant ground reaction force on a force plate) measured during the aiming phase, representing archer displacement	The three-dimensional position of one marker attached to the bow measured during the aiming phase, representing bow displacement	A significant correlation between CoP and bow displacement in the direction toward/away from the target (COP_X and D_X) and between COP_X and vertical displacement of the bow (D_Z) during the highest scoring shot
Lang and Zhou (2022b)	To identify the determinants of shooting performance in elite 10 m air rifle shooters	Twelve international-level 10 m air rifle shooters belonging to China's national team	A footscan 1.0 force platform used to collect postural balance parameters (A-P and M-L balance)	A SCATT MX-02 optoelectronic shooting test system used to collect shooting score and shooting technical variables (holding and aiming, stability of triggering, time)	The holding and aiming ability is the most important component, which could explain the 36.3% variance of shooting performance; The stability of triggering is the second important component, which could explain the 24.5% variance of shooting performance

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3.3. A summary of studies published so far, their gaps and proposals for future research

While most of the analyzed studies reported an improvement of static and/or dynamic balance as a result of training including a variety of sport-specific balance exercises, a direct relationship between postural sway variables and athletic performance measures has been investigated and demonstrated only in few of them.

A relationship between static bipedal balance and shooting accuracy in rifle shooters was found (Ball et al., 2003) but only at the inter-individual level (Mononen et al., 2007). Postural stability correlates negatively with aiming accuracy and shooting score and positively with triggering and holding stability (Lang and Zhou, 2022a). In particular, aiming and holding are essential factors of shooting performance, followed by stability of triggering (Lang and Zhou, 2022b). Similarly, Ihalainen et al. (2016a) identified the aiming accuracy, cleanness of triggering, timing of triggering, and stability of hold as key predictors of shooting performance, whereas the effect of postural balance on performance is small (Ihalainen et al., 2016a). While postural balance control in cross-shooting direction is related to stability of hold, in shooting direction it is related to cleanness of triggering (Ihalainen et al., 2016b). Postural sway displacement prior to the arrow release is also associated with shooting performance in juniors but not in more experienced senior archers (Mason and Pelgrim, 1986). A significant correlation also exists between postural and bow stability in recurve archers (Sarro et al., 2021). Static unipedal balance correlates with criterion variables of performance in elite golfers (Wells et al., 2009). Unilateral stance is associated with pitch velocity but not with pitching error in baseball players (Marsh et al., 2004). Dynamic balance is related to maximum skating speed in young ice-hockey athletes (Behm et al., 2005). Dynamic unipedal balance is also associated with maximum starting speed during a luge start (Platzer et al., 2009b), however not with ranking points (Platzer et al., 2009a) in snowboarders. Although balance measured in non-specific conditions is not a key factor of performance in the tennis serve, sport experience in expert tennis players leads to specific adaptations demonstrated by better ability to perform proper postural adjustments (Caballero et al., 2021).

Taking these findings into account, it is clear that little attention has been paid to the relationship between postural sway variables and athletic performance measures in the existing research. Though balance exercises performed within a given sport may contribute to the improvement of neuromuscular control of postural stability, the transfer to sport-specific performance has not been sufficiently demonstrated. Therefore, further studies including general body conditioning exercises and specific exercises within particular sports and their applications for enhancing athletic performance should be conducted. Specifically, the effectiveness of various exercise programs in generalizing transfer to sport-specific skills should be evaluated. Neurophysiological mechanisms underpinning adaptive changes in postural balance control following these exercises should be more precisely addressed. A better understanding of long-term changes in body balance under conditions specific to particular sports and their associations with athletic performance can provide useful information for designing exercise programs best suited to individual athlete needs.

4. Conclusion

Out of 126 articles, only 14 investigated the association between variables of postural balance control and measures of sport-specific performance. A relationship between static and/or dynamic balance and criterion measures of athletic performance was revealed in shooting, archery, golf, baseball, ice-hockey, tennis, and snowboarding. This may be ascribed to improved ability of athletes to perform postural adjustments in highly balanced task demands. However, the extent to which sport-specific exercises contribute to their better postural stability is unknown. Although there is a good deal of evidence supporting neurophysiological adaptations in postural balance control induced by body conditioning exercises, little effort has been made to explain balance adaptations induced by sport-specific exercises and their effects on athletic performance. While an enhancement in the athletic performance is often attributed to an improvement of neuromuscular functions induced by sport-specific balance exercises, it can be equally well ascribed to their improvement by general body conditioning exercises. Therefore, the relevant experiments have yet to be conducted to investigate the relative contributions of each of these exercises to improving athletic performance.

Author contributions

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

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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