frontiers | Research Topics

Human movement coordination in healthy and pathological conditions: from neuromuscular and kinetic principles to muscle-tendon function

Edited by
Adamantios Arampatzis, Lida Mademli, Kiros Karamanidis
and Maria-Elissavet Nikolaidou

Published in
Frontiers in Sports and Active Living

ISSN 1664-8714
ISBN 978-2-8325-6427-1
DOI 10.3389/978-2-8325-6427-1

About Frontiers

Frontiers is more than just an open access publisher of scholarly articles: it is a pioneering approach to the world of academia, radically improving the way scholarly research is managed. The grand vision of Frontiers is a world where all people have an equal opportunity to seek, share and generate knowledge. Frontiers provides immediate and permanent online open access to all its publications, but this alone is not enough to realize our grand goals.

Frontiers journal series

The Frontiers journal series is a multi-tier and interdisciplinary set of open-access, online journals, promising a paradigm shift from the current review, selection and dissemination processes in academic publishing. All Frontiers journals are driven by researchers for researchers; therefore, they constitute a service to the scholarly community. At the same time, the *Frontiers journal series* operates on a revolutionary invention, the tiered publishing system, initially addressing specific communities of scholars, and gradually climbing up to broader public understanding, thus serving the interests of the lay society, too.

Dedication to quality

Each Frontiers article is a landmark of the highest quality, thanks to genuinely collaborative interactions between authors and review editors, who include some of the world's best academicians. Research must be certified by peers before entering a stream of knowledge that may eventually reach the public - and shape society; therefore, Frontiers only applies the most rigorous and unbiased reviews. Frontiers revolutionizes research publishing by freely delivering the most outstanding research, evaluated with no bias from both the academic and social point of view. By applying the most advanced information technologies, Frontiers is catapulting scholarly publishing into a new generation.

What are Frontiers Research Topics?

Frontiers Research Topics are very popular trademarks of the *Frontiers journals series*: they are collections of at least ten articles, all centered on a particular subject. With their unique mix of varied contributions from Original Research to Review Articles, Frontiers Research Topics unify the most influential researchers, the latest key findings and historical advances in a hot research area.

Find out more on how to host your own Frontiers Research Topic or contribute to one as an author by contacting the Frontiers editorial office: frontiersin.org/about/contact

Human movement coordination in healthy and pathological conditions: from neuromuscular and kinetic principles to muscle-tendon function

Topic editors

Adamantios Arampatzis — Humboldt University of Berlin, Germany
Lida Mademli — Aristotle University of Thessaloniki, Greece
Kiros Karamanidis — London South Bank University, United Kingdom
Maria-Elissavet Nikolaidou — National and Kapodistrian University of Athens, Greece

Citation

Arampatzis, A., Mademli, L., Karamanidis, K., Nikolaidou, M.-E., eds. (2025). *Human movement coordination in healthy and pathological conditions: from neuromuscular and kinetic principles to muscle-tendon function*. Lausanne: Frontiers Media SA. doi: 10.3389/978-2-8325-6427-1

Table of contents

Frontiers in Sports and Active Living

TYPE Editorial
PUBLISHED 21 May 2025
DOI 10.3389/fspor.2025.1623815

Check for updates

OPEN ACCESS

EDITED AND REVIEWED BY
Gregoire P. Millet,
Université de Lausanne, Switzerland

*CORRESPONDENCE
Adamantios Arampatzis
a.arampatzis@hu-berlin.de

RECEIVED 06 May 2025
ACCEPTED 12 May 2025
PUBLISHED 21 May 2025

CITATION
Nikolaidou M-E, Mademli L, Karamanidis K and Arampatzis A (2025) Editorial: Human movement coordination in healthy and pathological conditions: from neuromuscular and kinetic principles to muscle-tendon function.
Front. Sports Act. Living 7:1623815.
doi: 10.3389/fspor.2025.1623815

Editorial: Human movement coordination in healthy and pathological conditions: from neuromuscular and kinetic principles to muscle-tendon function

Maria-Elissavet Nikolaidou[1], Lida Mademli[2], Kiros Karamanidis[3,4] and Adamantios Arampatzis[5,6]*

[1]Sports Biomechanics Laboratory, Faculty of Physical Education and Sport Science, School of Physical Education and Sport Science, National and Kapodistrian University of Athens, Athens, Greece, [2]Laboratory of Adapted Physical Education, Faculty of Physical Education and Sport Sciences, School of Physical Education and Sport Science (Serres), Aristotle University of Thessaloniki, Thessaloniki, Greece, [3]Department of Sport Science, Faculty of Mathematics and Natural Sciences, University of Koblenz, Koblenz, Germany, [4]School of Applied and Health Sciences, London South Bank University, London, United Kingdom, [5]Department of Training and Movement Sciences, Humboldt-Universität zu Berlin, Berlin, Germany, [6]Berlin School of Movement Science, Humboldt-Universität zu Berlin, Berlin, Germany

KEYWORDS
human movement, kinematics, neuromuscular, muscle-tendon, coordination

Editorial on the Research Topic
Human movement coordination in healthy and pathological conditions: From neuromuscular and kinetic principles to muscle-tendon function

Effective and safe movement in complex environments relies on task-specific limb coordination with high temporal and spatial variability; it is also crucial for independent living. Regular exercise and repeated exposure trigger neural and musculoskeletal adaptations, leading to new or refined coordination patterns. Understanding these principles requires integrating knowledge from neuromuscular circuits to joint mechanics and muscle-tendon function. This Research Topic in Frontiers in Sports and Active Living focuses on current research exploring how human movement coordination in both healthy individuals and in those with pathological conditions is organized from a neuromuscular perspective to kinetic principles and to the level of muscle-tendon function.

In a comprehensive study, Fischer et al. investigated lumbo-pelvic posture and coordination, trunk dynamic stability, and movement variability in asymptomatic and chronic low back pain participants. Clear sex-specific differences in lumbo-pelvic coordination and posture measures were found, while pain intensity had only a small effect, indicating the limited utility in the clinical diagnosis and management of low back pain of these variables. Their findings of greater local trunk instability in the participants with chronic low back pain, which predict control errors in the regulation of trunk movement, could be considered as a useful diagnostic tool. Hakamata et al. used immersive virtual reality to manipulate the placement of the leading limb, examining whether controlling its position could facilitate the need to prioritize movement planning of the trailing limb during a virtual obstacle-crossing task. Their findings suggested a reduction in the collision rate of the trailing limb as a result of a safer motor strategy of obstacle avoidance by the leading limb, thus highlighting the trade-off between speed and accuracy, which older adults could be trained to effectively execute. Köhler et al. investigated the kinetics of the javelin throw and found that the energy generated during the acceleration phase significantly influences joint moments and energy transfer in the subsequent deceleration phase. Their study highlights the crucial role of the acceleration phase in optimizing energy flow and reducing injury risk while ensuring the athlete's better technical preparation. The work by Steingrebe et al. offers a comprehensive review and meta-analysis of hip osteoarthritis on lower limb joint angles in the three planes of motion during locomotor tasks of varying difficulty involving gait, stair walking, and turning. Overall, this work highlights the complex interplay of kinematic changes affecting both the ipsilateral and contralateral sides as well as the task-specific kinematic alterations during different phases of locomotion.

Biomechanical and sensorimotor orthotic interventions have distinct roles depending on the action mechanisms of the joint kinematics of the lower extremities. The study by Simon et al. aimed to analyze the effects of biomechanical and sensorimotor foot orthoses on gait kinetics in patients with patellofemoral pain through a randomized controlled trial. Their results revealed that sensorimotor foot orthoses induced distinct kinematic adaptations in lower extremity motion, potentially reducing patellofemoral pain during walking. In their contribution, Pfile et al. investigated the effect of the technique of anterior cruciate ligament reconstruction with a quadriceps tendon autograft during gait in patients with unilateral injury almost one-year post-surgery. Their findings showed significant biomechanical alterations and gait patterns associated with quadricep avoidance and diminished proximal forces during specific phases of gait. The role of the "upper body strategy" as a possible compensatory mechanism to the impaired dynamic balance performance following exercise-induced lower limb muscle fatigue was examined in healthy youth by Borgmann et al., where findings showed that free arm movement did not moderate the impact of neuromuscular fatigue on dynamic balance performance.

Mademli et al. investigated the diversity and flexibility of activation patterns within the synergistic triceps surae and quadriceps muscles during a visually guided postural task in stable and unstable conditions. Although the similarity of activation patterns of the synergistic muscles decreased during the unstable condition, the lower values within the triceps surae muscles in both examined conditions indicate increased flexibility and diversity of neuromuscular control to meet specific joint stabilization challenges. Liang et al. focused on the chronic adaptation of the biomechanical properties of the Achilles tendon in long-term post-stroke patients. They reported region-specific degeneration in tendon thickness and reduced collagen fiber organization in the paretic limb, increasing the risk for the initiation of tendinopathy and, thus, supporting more targeted rehabilitation strategies to cope with the impaired lower extremity function typically characterizing this population. Further, Magris et al. assessed the intrinsic and morphological properties of vastus lateralis muscle in patients with Parkinson's disease and found altered force generation capacity of the muscle during contraction, whereas muscle mechanical power emerged as a potential parameter useful for clinical evaluation between the less and the more affected side. Kim et al. synthesized the current evidence on the effect of sensorimotor training on pain and functional outcomes in the Achilles tendon where, despite the low number of included studies, potential positive effects of sensorimotor training in conjunction to high loading strategies were shown with regards to pain, strength, and short-term performance outcomes. Changes in hormonal concentrations during the menstrual circle have been suggested as a possible causal factor for the altered mechanical properties in tendons amounting to a higher injury rate in females. In their contribution, Saito et al. found that muscle as well as tendon stiffness in the anterior and lower posterior thigh regions were unaffected by menstruation, although a positive correlation only in the anterior thigh region with the early luteal phases of menstruation was found. Finally, the systematic treatment of eccentric muscle function as a training modality in sports and rehabilitation due to well-documented acute and prolonged adaptations has been sufficiently investigated. Vila-Chã et al. focused on the less explored neuromuscular and temporal alterations of eccentric exercise and their motor-task specificity and showed that isotonic eccentric exercise induced different responses depending on the investigated neuromuscular functional outputs and motor tasks.

The article collection presents findings that contribute to our understanding of the principles of human movement control and coordination in both healthy individuals and those with pathological conditions. Collectively, the results point to significant advances in targeted exercise or rehabilitation strategies. Yet, due to the complex interaction between human motor control at the neuromuscular, joint-limb, and muscle-tendon levels and their resulting modulation and/or adaptation to environmental demands, many issues require further investigation. Some studies suggest that more sensitive and clinically useful measures are needed to better clarify adaptations in temporal and spatial motor responses as well as in the intrinsic properties of the neuromuscular system between patients and asymptomatic controls in pathologies like chronic low back pain, Parkinson's disease, or stroke. Likewise, it is necessary that these altered functional coordination patterns be investigated in experimental designs able to manipulate task complexity, populations with different functional levels, and populations at various stages of recovery. We would like to thank all contributors for their participation to this Research Topic and look forward to joining our efforts in human movement coordination research.

Author contributions

MN: Writing – review & editing, Writing – original draft. LM: Writing – original draft, Writing – review & editing. KK: Writing – review & editing, Writing – original draft. AA: Writing – review & editing, Writing – original draft.

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.

The author(s) declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision.

Generative AI statement

The author(s) declare that no Generative AI was used in the creation of this manuscript.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

TYPE Original Research
PUBLISHED 03 November 2023
DOI 10.3389/fspor.2023.1273152

OPEN ACCESS

EDITED BY
Yaodong Gu,
Ningbo University, China

REVIEWED BY
Xuanzhen Cen,
Óbuda University, Hungary
Yuqi He,
University of Pannonia, Hungary

*CORRESPONDENCE
Carolina Vila-Chã
✉ cvilacha@ipg.pt

RECEIVED 05 August 2023
ACCEPTED 18 October 2023
PUBLISHED 03 November 2023

CITATION
Vila-Chã C, Bovolini A, Francisco C, Costa-Brito AR, Vaz C, Rua-Alonso M, de Paz JA, Vieira T and Mendonca GV (2023) Acute effects of isotonic eccentric exercise on the neuromuscular function of knee extensors vary according to the motor task: impact on muscle strength profiles, proprioception and balance.
Front. Sports Act. Living 5:1273152.
doi: 10.3389/fspor.2023.1273152

Acute effects of isotonic eccentric exercise on the neuromuscular function of knee extensors vary according to the motor task: impact on muscle strength profiles, proprioception and balance

Carolina Vila-Chã[1,2]*, Antonio Bovolini[1,2], Cristiana Francisco[1], Ana R. Costa-Brito[1], Cláudia Vaz[1], María Rua-Alonso[1,2,3], José Antonio de Paz[4], Taian Vieira[5,6] and Goncalo V. Mendonca[7,8]

[1]Laboratory for the Assessment of Sports Performance, Physical Exercise and Health (Labmov), Polytechnic of Guarda, Guarda, Portugal, [2]Research Center in Sports Sciences, Health Sciences, and Human Development, Vila Real, Portugal, [3]Performance and Health Group, Department of Physical Education and Sport, Faculty of Sports Sciences and Physical Education, University of A Coruna, A Coruña, Spain, [4]Institute of Biomedicine, University of León, León, Spain, [5]Laboratorio di Ingegneria del Sistema Neuromuscolare (LISiN), Dipartimento di Elettronica e Telecomunicazioni, Politecnico di Torino, Turin, Italy, [6]PoliToBIOMed Lab, Politecnico di Torino, Turin, Italy, [7]Neuromuscular Research Laboratory, Faculdade de Motricidade Humana, Universidade de Lisboa, Lisbon, Portugal, [8]Interdisciplinary Center for the Study of Human Performance (CIPER), Faculdade de Motricidade Humana, Universidade de Lisboa, Lisbon, Portugal

Introduction: Eccentric exercise has often been reported to result in muscle damage, limiting the muscle potential to produce force. However, understanding whether these adverse consequences extend to a broader, functional level is of apparently less concern. In this study, we address this issue by investigating the acute and delayed effects of supramaximal isotonic eccentric exercise on neuromuscular function and motor performance of knee extensors during tasks involving a range of strength profiles, proprioception, and balance.

Methods: Fifteen healthy volunteers (23.2 ± 2.9 years old) performed a unilateral isotonic eccentric exercise of the knee extensors of their dominant lower limb (4 × 10 reps at 120% of one Repetition Maximum (1RM)). The maximum voluntary isometric contraction (MVC), rate of force development (RFD), force steadiness of the knee extensors, as well as knee joint position sense and mediolateral (MLI) and anteroposterior stability (API) of the dominant lower limb, were measured pre-, immediately, and 24 h after the eccentric exercise. The EMG amplitude of the vastus medialis (VM) and biceps femoris (BF) were concomitantly evaluated.

Results: MVC decreased by 17.9% immediately after exercise ($P < 0.001$) and remained reduced by 13.6% 24 h following exercise ($P < 0.001$). Maximum RFD decreased by 20.4% immediately after exercise ($P < 0.001$) and remained reduced by 15.5% at 24 h ($P < 0.001$). During the MVC, EMG amplitude of the VM increased immediately after exercise while decreasing during the RFD task. Both values returned to baseline 24 h after exercise. Compared to baseline, force steadiness during submaximal isometric tasks reduced immediately after exercise, and it was accompanied by an increase in the EMG amplitude of the VM. MLI and knee joint position sense were impaired immediately after isotonic eccentric exercise ($P < 0.05$). While MLI returned to baseline values 24 h later, the absolute error in the knee repositioning task did not.

Discussion: Impairments in force production tasks, particularly during fast contractions and in the knee joint position sense, persisted 24 h after maximal isotonic eccentric training, revealing that neuromuscular functional outputs were affected by muscle fatigue and muscle damage. Conversely, force fluctuation and stability during the balance tasks were only affected by muscle fatigue since fully recovered was observed 24 h following isotonic eccentric exercise.

KEYWORDS

eccentric exercise, isotonic load, rate of force development, force steadiness, joint position sense, postural control

1. Introduction

In the last decades, eccentric exercise has gained a growing interest in sports and rehabilitation fields, leading to the development of a variety of training protocols through the manipulation of different training variables, including mechanical loads (1, 2). Eccentric exercise can be performed at constant angular velocity movement (isokinetic) or against a constant external load (isotonic), providing distinct mechanical loads as a stimulus, which might lead to specific neuromuscular adaptations. Most studies have used isokinetic mechanical loads to investigate the chronic and acute effects of eccentric training on different parameters of health and sports performance (3–5). However, isotonic loads are more commonly applied in the training field (2, 6). Isotonic eccentric training with external loads exceeding the load that subjects are able to lift for one repetition (1 RM load) has become popular among coaches, athletes, and fitness practitioners (6, 7). Several studies have shown that this training mode is more effective than isokinetic eccentric training in improving muscle strength, muscle architecture, and activation (5, 6, 8). The greater effectiveness of this type of training has been justified by the mechanical overload and the greater limb acceleration at the beginning of the isotonic contractions (6, 8). On the other hand, such mechanical stress can favor muscle damage and related delayed onset muscle soreness (DOMS) (3, 9, 10). Several studies have revealed that eccentric contractions are strongly associated with skeletal muscle impairments, consisting of structural disruption of sarcomeres, disturbed excitation–contraction coupling, and calcium signaling, leading to an inflammatory response (10–13). Moreover, these alterations are frequently accompanied by altered neural drive to the muscles (4, 14).

The neurophysiological changes induced by eccentric exercise can negatively impact neuromuscular function (3, 15, 16). Evidence has shown that isokinetic eccentric contractions of the knee extensors induce an acute and delayed decrease of different muscle force profiles, including maximal muscle strength (10, 13), rate of force development (4, 17), and force steadiness (18, 19). However, the magnitude of changes might vary depending on the muscle strength profile measured. Besides alterations in contractile properties, eccentric contractions can also disturb proprioception (20, 21). Significant disturbances in the joint position sense were found after maximal and sub-maximal eccentric exercise (22–24). Such impairments may contribute to an increased risk of injury and impaired postural control. Nevertheless, despite extensive research on the effects of eccentric exercise on proprioception, no study has investigated its impact on postural balance, especially when isotonic eccentric exercise with supramaximal loads is applied.

The extent to which eccentric exercise and related DOMS affects neuromuscular integrity and motor performance may vary depending on the mechanical characteristics of the eccentric load (5, 15). Moreover, the performance of different motor tasks seems to rely on distinct neuromuscular pathways and neural drive delivered to specific muscle groups. As a result, the impact of eccentric exercise and DOMS on motor performance can vary depending on the task specificity. Certain motor tasks may be more sensitive to the effects of DOMS, leading to a longer recovery period before optimal performance is restored. On the other hand, some motor tasks may exhibit resilience to DOMS or may recover more rapidly. Therefore, it is essential to consider the specific demands of different motor tasks when designing eccentric exercise programs and determining appropriate recovery strategies. Thus, this study aimed to investigate the acute effects of supramaximal isotonic eccentric exercise on different force profiles of the knee extensors, specifically maximum strength, rate of force development and force steadiness. Additionally, we aimed to evaluate the impact of this eccentric exercise mode on knee proprioception and dynamic balance. Changes in agonist and antagonist activity during force and balance tasks were also examined. By comprehensively investigating these factors, our study seeks to enhance the understanding of the implications of isotonic eccentric exercise and DOMS on neuromuscular function and motor performance. It was postulated that the magnitude of changes in the motor performance induced by a supramaximal isotonic eccentric exercise would depend on the motor task specificity. Moreover, we also hypothesized that changes in motor performance would be accompanied by alterations in the activation of the agonist and antagonist muscles.

2. Methods

2.1. Participants

Fifteen volunteers participated in the study (nine men and six women, age 23.2 ± 2.8 years; height 169.3 cm ± 6.5 cm; weight 63.8 ± 8.4 kg). All participants were practitioners of exercise, familiarized with strength training, and without any pathology in the dominant lower limb. The study was approved by the Polytechnic of Guarda Committee on Research Ethics and performed according to the Declaration of Helsinki. All subjects gave written informed consent before undertaking testing and training.

2.2. Experimental design

The participants visited the laboratory 3 times: 48 h before the experimental sessions to get familiarized with the experimental setup and to estimate the 1RM of the knee extensor of the dominant lower limb (session 1); to perform the isotonic eccentric exercise session and assess the neuromuscular function before and immediately after exercise session (session 2) and 24 h after eccentric exercise for neuromuscular function reassessment and DOMS evaluation (session 3). All experimental measurements were assessed by following the same order: (i) dynamic balance; (ii) knee proprioception; (iii) maximum voluntary isometric contraction; (iv) explosive isometric contraction; and (v) force steadiness. The flow chart in **Figure 1** summarizes the experimental protocol.

2.3. Isotonic eccentric exercise protocol

In order to determine the exercise intensity of the isotonic eccentric exercise, the 1RM of the knee extensors of the dominant lower limb was estimated for all participants using a multiple repetition test protocol and a regular leg extension machine (1MTH081, Panatta SLR, Italy). After a 5-min warm-up in a cycle ergometer, familiarization trials were performed to ensure proper execution of the test. The assessment protocol followed the guidelines outlined by National Strength and Conditioning Association Guidelines (25). The 1RM was estimated based on the Lombardi linear regression equation (26). The estimated 1RM was then used to define the exercise intensity of the isotonic eccentric protocol. After at least 48 h, the participants visited the laboratory to perform a single exercise session consisting of 4 sets of 10 repetitions at 120% of 1RM, with a recovery interval of 1 min and 30 s between sets. To guarantee proper technical execution and physical integrity of the participants, the following criteria were established: (i) participants were well seated in the machine, with their back well supported on the machine seat (support cushion adjusted individually); (ii) the participants were instructed to exert maximal force only during the descending phase of the movement (eccentric action) for 2 s; (iii) joint range of motion was established between 180° (complete extension) and 90° of knee flexion; (iv) the non-dominant lower limb remained stationary, in a neutral and supported position; and (v) the concentric phase of the movement was executed by two assistants.

2.4. Dynamic balance test

Unilateral dynamic balance of the dominant lower limb was measured with the Biodex Balance System (BBS) (model 945-300, United States of America). Participants were instructed to stand in their dominant lower limb on the BBS locked foot platform, keeping their body straight and the upper limbs lateral to the body and with the unsupported leg placed in a comfortable knee-flexed position without touching any surface. Then, the foot platform was released, and participants were asked to maintain an upright standing position for 20 s without contact with any other surfaces. During the test, the platform stability decreased progressively from level 8 to level 2. Each participant performed five to six trials with an interval in-between of 40 s. Deviations in the anteroposterior (x-axis) and mediolateral (y-axis) relative to the platform center were sampled at frequency of 20 Hz. Then, anteroposterior stability index (API), mediolateral stability index

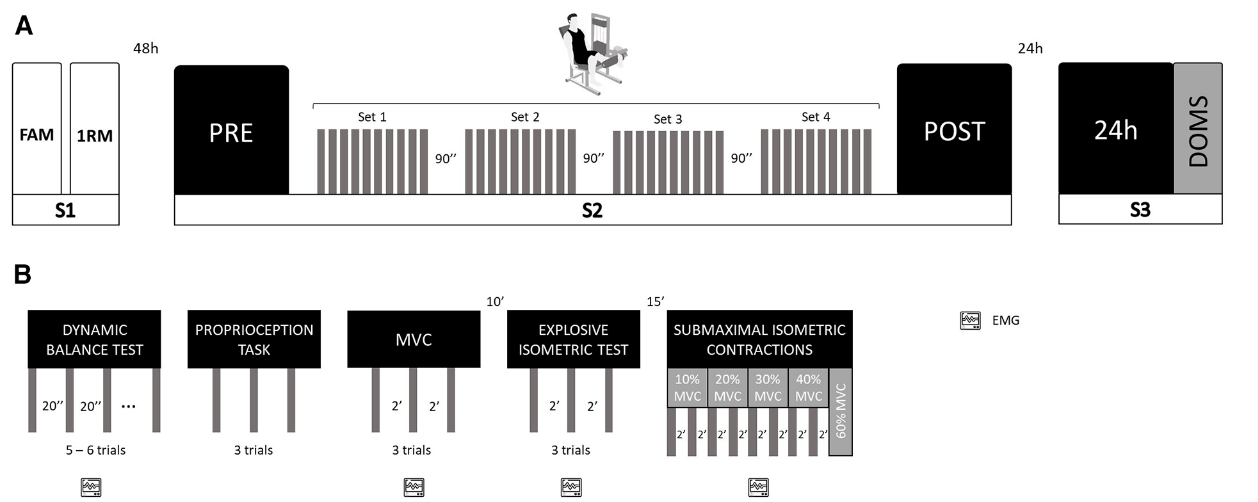

FIGURE 1
Schematical representation of the study design. (A) Experimental design; S, session; FAM, familiarization; 1RM, 1-repetition maximum test; PRE, assessments before isotonic eccentric protocol; POST, assessments immediately after isotonic eccentric protocol; 24 h, 24 h after isotonic eccentric protocol; DOMS, delayed onset muscle soreness; (B) Neuromuscular function assessment design; EMG, electromyographic recording; MVC, maximum voluntary isometric contraction.

(MLI) and global stability index (SI) were computed with the BBS software, as described by Biodex Medical Systems, Inc. (27). API and MLI represent the variance of foot platform displacement in degrees for motion in the sagital and frontal plans, respectively (27). SI represents the variance of foot platform displacement in degrees, in all motions during a test. These indices were calculated using the degree of oscillation of the platform, in which low values indicated that the individual had good stability (28). For each participant the indexes corresponded to the average of the values obtained in the five to six trials. Electromyography (EMG) data of the vastus medialis (VM), and biceps femoris (BF) of the dominant leg were recorded while the participants were performing the test.

2.5. Knee proprioception test

The participants were seated in an elevated chair and blindfolded, with the knees approximately at 90° and the legs freely moving, without contact with any surface. Markers were placed on the external condyles of the femur (approximately 20 cm above the knee) and tibia (approximately 20 cm below the knee) for further knee angle position identification. The protocol consisted of positioning the non-dominant leg at an angle of 130° (manually measured with a goniometer), and subsequently participant was requested to move their dominant leg to match the position non-dominant leg as accurately as possible. Participants were allowed to practice the tasks and then repeated it three times. The trials were recorded with a video camera (Nikon, D3200, 1,920 × 1,080 pixels, 30 fps), positioned perpendicular to the subject, at two meters from the chair. The joint angle was then later assessed using the Kinovea program (version 0.8.15). For each trial the joint position error was assessed by computing: (*i*) absolute (shows the magnitude of the error); (*ii*) constant (indicates the direction of the error); and (*iii*) variable (indicates the response variability).

2.6. Muscle strength profiles

Maximal, explosive and submaximal isometric contractions of the knee extensors of the dominant lower limb were performed to assess the impact of the isotonic eccentric exercise on the muscle strength and motor control profiles. MVC was measured with the participants seated in an elevated chair (customized chair made by INEGI-UP), with their arms crossed in front of their chest and with the trunk and hips firmly strapped to the chair. The dominant leg was positioned at 90° of flexion and adjustable padded strap was placed around the tibiotarsal region, attached to a load cell (model load cell 614, SENSOR, United Kingdom), and fixed to the chair. The unassessed lower limb was positioned atop a box, ensuring it remained flexed, while exercising utmost caution to avoid applying any force. The participants were then instructed to progressively increase force against the padded strap and encouraged to exert their maximum within the first 5 s. Each participant performed 3 repetitions with 2 min of rest in between. The MVC value corresponded to the maximal force exerted in the three trials. Ten minutes after the MVC test, and in the same previous position, the participants performed three explosive isometric contractions, and they were encouraged to exert their maximum force as fast as possible. A rest of 2 min was given between trials.

After 15 min of rest, the participants performed submaximal isometric contractions at 10%, 20%, 30%, 40%, and 60% MVC, keeping it as stable as possible for 10 s. The participants completed two trials for each submaximal isometric contraction at 10%, 20%, 30%, and 40% MVC, as well as one trial for the submaximal isometric contraction at 60% MVC. A two-minute recovery interval was given between each trial. Participants were provided with visual feedback of the force exerted, that was displaced in computer screen of 22 inches in from them. It was required to the participants to maintain the target force level within two error bars of 2% MVC, centered around the target force level. EMG signals from the VM and BF and force were concomitantly record.

2.7. Assessment of muscle pain

To confirm the presence of DOMS 24 h post-eccentric exercise, participants verbally rated their perceived pain on a scale from 0 ("'no soreness"') to 10 ("'worst soreness imaginable"'). The subjects were asked to rate the average pain intensity in the quadriceps during their regular activities of daily living (e.g., descending stairs) since their last visit to the laboratory (over the past 24 h).

2.8. EMG and force recordings

Surface EMG signals were acquired from the VM and BF muscles during the dynamic balance and isometric tasks used to measure the different muscle strength profiles. Signals from the selected muscles were recorded with three pairs Ag-AgCl electrodes (Ambu Neuroline, Denmark; conductive area 28 mm^2), placed as recommended by Hermens et al. (29). Before placement of the electrodes, the skin was shaved, lightly abraded, and cleansed with water. A ground electrode was placed around the ankle of the dominant lower limb. Surface EMG signals were amplified as bipolar derivations (EMG amplifier, OT Bioelettronica, Turin, Italy), band-pass filtered (−3 dB bandwidth, 10–500 Hz), sampled at 2048 samples/s, and converted to digital data by a 12-bit A/D converter board. The EMG electrodes remained in place during the exercise session and were used for all EMG measurements on the same day (before and immediately after training). At the end of the experiment, the electrodes were removed, and the electrode positions were marked on the skin to ensure that the electrodes were placed in the same location for the EMG recordings obtained 24 h after training. The force signals were measured with a load cell (load sensitivity = 0.0048 V/N; SENSOR, load cell 614, United Kingdom) and collected through an auxiliary channel of the EMG amplifier (OT Bioelettronica, Italy), sampled at 2048 samples/s, and converted to digital data by a 12-bit A/D converter board and recorded in the OT Biolab software (OT Bioelettronica, Italy), thus allowing the simultaneous collection of force and EMG signals.

2.9. EMG and force signal analysis

For the reference MVC, the average rectified value (ARV) was computed from a time interval of 250 ms centered at instant of the maximal force. During the explosive contractions, the ARV was computed from two-time intervals: (i) 70 ms prior to the onset of the contraction and (ii) 50 ms centered at the time instant of the maximal slope (19). The ARV values for each muscle and task were normalized by the respective muscle's ARV obtained during the MVC [(task ARV/ARV MVC) × 100]. During dynamic balance tests the percent of activation time of the VM and BF was determined by using the algorithm described by Vincent and Soile (30). The analog signals from the load cell were converted into force (N, Newton) and the maximum rate of force development (RFD) was computed as the maximum slope of the force-time curve. The onset of the contraction was defined as the time instant that force exceeded 3 times the standard deviation value observed at baseline. For each submaximal contraction, the coefficient of variation (CoV) of force was calculated over the entire duration of the contraction. The CoV of force was computed by dividing the standard deviation (SD) of the force signal by the mean force multiplied by 100 [(SD/mean force) × 100]. This measure expresses the absolute force variability as a fraction of the mean force exerted (31).

2.10. Statistical analysis

Statistical analyses were performed using the Statistical Package for Social Sciences (SPSS Version 24, IBM Corporation, Armonk, New York, USA) software. The normality of the dependent variables was confirmed using the Shapiro-Wilk test. The acute effects of the isotonic eccentric training on muscle force profile (MVC and maximum RFD), dynamic balance (GBI, API and MLI), proprioception (absolute, constant and variable errors) and muscle activity were assessed with one-way repeated-measures ANOVA, with factor time (Pre-, Post-, and 24 h after isotonic eccentric exercise). During submaximal isometric contractions, changes in the CoV of the force and ARV of the VM and BF were assessed with a two-way repeated measures ANOVA with factor time and target force levels (10%, 20%, 30%, 40% and 60% of MVC). Pairwise comparisons were performed with the Bonferroni *post hoc* test when ANOVA was significant. Partial eta-squared (ηp^2) was used to calculate the effect size of the statistical results, which were classified as weak ($\eta p^2 < 0.01$), medium (ηp^2 $0.01 < 0.06$) or high ($\eta p^2 > 0.14$) (32). The significance level was set to $P < 0.05$. Results are reported as means ± SD.

3. Results

3.1. Pain perception

Twenty-four hours after isotonic eccentric exercise, the participants rated their perceived pain intensity as 3.6 ± 2.6 on a scale from 0 to 10.

3.2. Maximum voluntary contraction

The MVC of the knee extensors decreased immediately after (−17.9%; $P < 0.001$) and 24 h after training session (−13.6%; $P < 0.001$) compared to pre-exercise values (**Figure 2A**; main effect: $P < 0.001$, $\eta p^2 = 0.671$). These alterations were accompanied by a decline in the ARV of the vastus medialis ($P < 0.040$, $\eta p^2 = 0.238$; **Table 1**), while no changes were observed in the ARV of the biceps femoris ($P < 0.207$, $\eta p^2 = 0.106$;

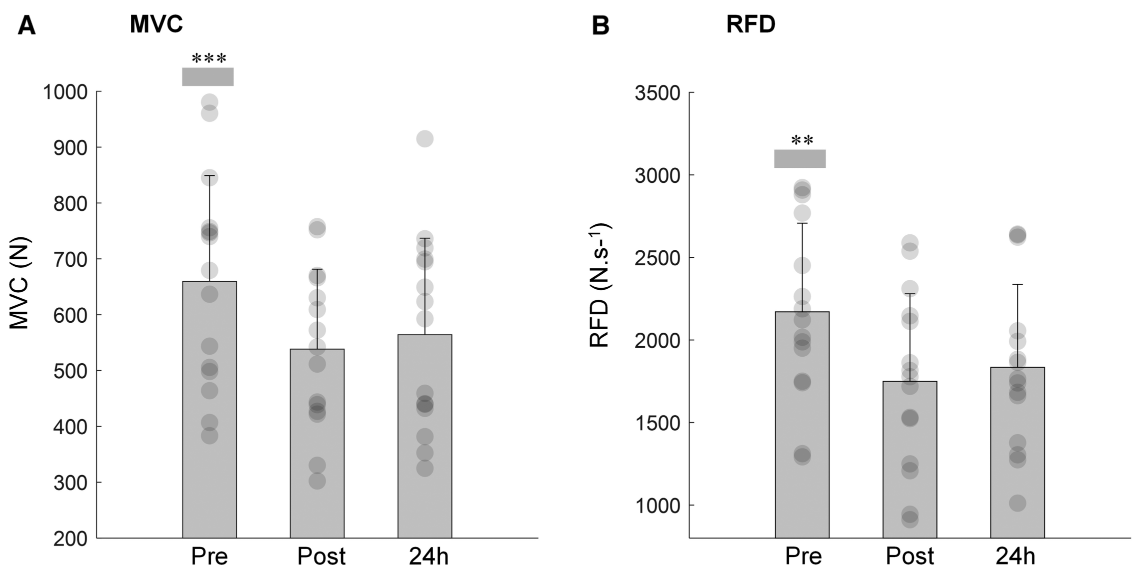

FIGURE 2
Mean and SD of the maximal voluntary contraction (A; MVC) and rate of force development (B; RFD) of the knee extensor pre, post and 24 h after isotonic eccentric exercise. ***MVC immediately post and 24 h after exercise was significantly reduced when compared with pre-condition ($P < 0.001$). **RFD immediately post and 24 h after exercise was significantly reduced when compared with pre-condition ($P < 0.001$).

TABLE 1 Average rectified value (ARV) for the vastus medialis (VM) and biceps femoris (BF) obtained during the maximal voluntary contraction (MVC), the isometric explosive contractions at onset of the contraction and at time instant that maximum rate of force development (RFD) was reached and during submaximal isometric contractions at target force levels of 10%, 20%, 30%, 40% and 60% MVC obtained at baseline (pre), immediately post, and 24 h after isotonic eccentric exercise.

	Vastus medialis			Biceps femoris		
Variable	Pre	Post	24 h	Pre	Post	24 h
MVC [ARV (μV)]	367.72 ± 122.65	312.66 ± 110.25[§]	347.92 ± 130.38	113.04 ± 63.50	120.61 ± 45.01	97.28 ± 25.21
RFD onset (% ARV MVC)	59. 63 ± 23.70	79.80 ± 20.27[§]	62. 99 ± 24.92	71.99 ± 25.74	86.23 ± 34.92	76.70 ± 36.34
RFD max (% ARV MVC)	69.89 ± 15.33	92.50 ± 33.49[‖]	72.58 ± 33.49	69.78 ± 13.05	82. 82 ± 24.88	80. 01 ± 26.23
10% MVC (% ARV MVC)	12.53 ± 3.64	16.22 ± 5.29[§,†]	11.93 ± 3.97	14.42 ± 4.52	17.13 ± 8.31	15.97 ± 5.25
20% MVC (% ARV MVC)	21.62 ± 6.24	29.59 ± 8.86[§,‡]	20.52 ± 6.37	22.51 ± 4.95	28.27 ± 10.31	25.04 ± 7.84
30% MVC (% ARV MVC)	33.06 ± 6.93	41.50 ± 9.72[‡,†]	32.89 ± 10.59	35.30 ± 7.03	39.01 ± 12.43	36.92 ± 13.48
40% MVC (% ARV MVC)	46.03 ± 11.20	56.39 ± 10.68[§,‖]	45.98 ± 12.05	50.39 ± 16.05	39.32 ± 8.11	41. 91 ± 10.39
60% MVC (% ARV MVC)	48.97 ± 8.84	53.83 ± 7.65	49.62 ± 11.97	39. 78 ± 8.11	41. 91 ± 10.39	39.78 ± 11.27
Dynamic balance (% time activation)	46.62 ± 15.53	57.17 ± 13.86	49.82 ± 23.10	38.83 ± 16.13	45.67 ± 15.88	39.73 ± 21.80

[†]Post significantly different from 24 h ($P < 0.01$).
[‡]Post significantly different from pre-condition ($P < 0.01$).
[§]Post significantly different from pre-condition ($P < 0.05$).
[‖]Post significantly different from pre and after 24 h ($P < 0.05$).

Table 1). Following exercise session, the vastus medialis ARV significantly decreased ($P = 0.017$) returning to baseline values 24 h after ($P = 1.000$).

3.3. Rate of force development

As for MVC, the maximum RFD was substantially reduced immediately after the isotonic eccentric exercise (−20.3%; $P < 0.001$), remaining diminished 24 h following exercise (−15.5%; $P < 0.001$) when compared to pre-exercise values (Figure 2B; time effect: $P < 0.001$, $\eta p^2 = 0.608$). The time to reach maximum RFD did not differ between pre (179 ± 3.6 ms), post (183 ± 4.0 ms) and 24 h after eccentric exercise (174 ± 2.9 ms) (time effect: $P = 0.615$.) The EMG amplitude of the VM, at onset of the explosive isometric contraction (time effect: $P = 0.025$, $\eta p^2 = 0.435$) and at the time instant that maximum RFD was reached (time effect: $P = 0.014$, $\eta p^2 = 0.483$) increased immediately post ($P < 0.05$, for both time intervals) and returned to baseline 24 h after exercise session (Table 1). No alterations were observed in the ARV of the BF for both time intervals of the explosive contractions (Table 1, time effect: $P > 0.180$, $\eta p^2 > 0.122$, in both time intervals).

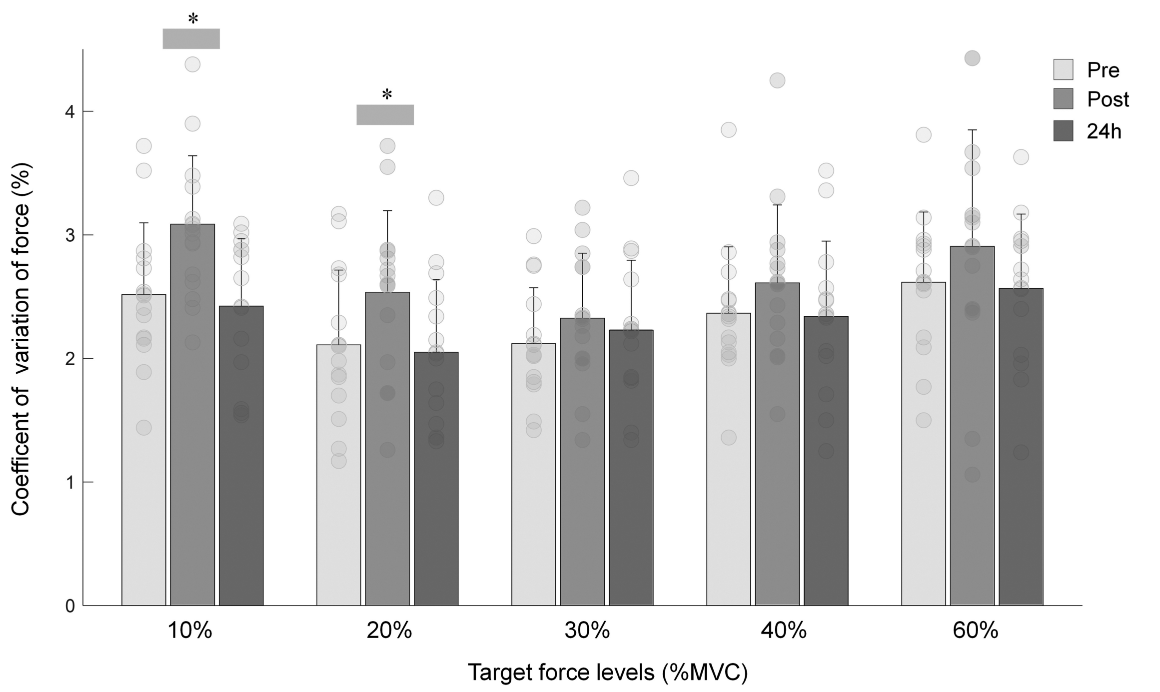

FIGURE 3
Mean and SD of the coefficient of variation (CoV) of force during submaximal isometric contractions at target force levels of 10%, 20%, 30%, 40% and 60% MVC obtained at baseline (pre), immediately post, and 24 h after eccentric exercise. *Indicates that CoV post eccentric exercise is significantly different from the CoV measured before (pre) and 24 h after exercise ($P < 0.05$).

3.4. Force steadiness

Figure 3 shows the CoV of force at the different targe force levels measured pre, immediately post and 24 h after isotonic eccentric exercise. Data analysis revealed a significant effect for both factors load ($P = 0.004$, $\eta p^2 = 0.289$) and time ($P < 0.001$, $\eta p^2 = 0.491$). CoV at 10% MVC was higher than the CoV observed at 20% and 30% of MVC ($P = 0.028$ and $P = 0.042$, respectively). At 60% MVC, there was a significantly higher CoV compared to the CoV at 20% and 30% ($P = 0.020$ and $P = 0.013$, respectively).

The pooled data showed that CoV significantly increased post exercise (+14.7%; $P = 0.025$), returning to baseline values in the 24 h after (Figure 3). No interaction time × load was observed ($P = 0.343$, $\eta p^2 = 0.075$). When stratified by force levels, the results revealed that only at 10% and 20% MVC, the CoV was significantly higher immediately after exercise ($P = 0.014$), returning to baseline values 24 h after exercise. Changes in the force steadiness were accompanied by significant alterations in the ARV of the VM (time × load effect: $P = 0.041$, $\eta p^2 = 0.166$; Table 1). The ARV of the VM increased immediately after exercise for at all force levels ($P < 0.030$), except for the 60%MVC ($P = 0.331$). Twenty-four hours later the ARV values returned to baseline (Table 1). No significant changes were observed in the EMG amplitude of the biceps femoris either immediately post or 24 h after the eccentric exercise (time effect: $P = 0.441$, $\eta p^2 = 0.057$; Table 1).

3.5. Proprioception

Figure 4 illustrates the absolute, constant, and variable errors during the knee repositioning task, before, immediately post, and 24 h after the isotonic eccentric session. The results showed that participants significantly increased the magnitude of the repositioning error following eccentric exercise (time effect: $P = 0.002$ $\eta p^2 = 0.358$; Figure 4A). The absolute error increased immediately after exercise (+2.4 ± 2.1°; $P = 0.021$) and remained higher 24 h after when compared to baseline (+2.7 ± 2.7°; $P = 0.020$) (Figure 4A). After training, participants moved into a more extended knee position relative to the reference leg, as indicated by the constant error (time effect: $P = 0.012$ $\eta p^2 = 0.270$; Figure 4B). Immediately after the training session, the constant error increased by +5.5 ± 3.1° when compared to baseline ($P = 0.015$). Twenty-four hours later the constant error still increased, but no statistical differences were observed between pre and 24 h after training conditions ($P = 0.133$; Figure 4B). The position consistency, as indicated by the variable error, although declined was not significantly affected by the present exercise protocol (time effect: $P = 0.435$ $\eta p^2 = 0.057$; Figure 4C).

3.6. Dynamic balance

Figure 5 shows the results on the global balance, anteroposterior and mediolateral stability indexes before, immediately post and 24 h after eccentric isotonic exercise. Global balance and anteroposterior indexes were not affect by the exercise protocol (time effect: $P = 0.483$ $\eta p^2 = 0.051$ and $P = 0.685$ $\eta p^2 = 0.025$, respectively; Figures 5A,B). Nonetheless, significant alterations were observed in the mediolateral direction (time effect: $P = 0.002$, $\eta p^2 = 0.399$; Figure 5C). The instability in the mediolateral direction significantly increased immediately post exercise ($P < 0.001$). Following 24 h, the mediolateral index remained higher, but no statistical differences were observed when compared to baseline values ($P = 0.092$) (Figure 5C).

Although there was an increase in the relative time of VM and BF activation during balance tests immediately after the eccentric training, the results were not statistically significant (time effect: $P > 0.434$; $\eta p^2 < 0.080$, for both muscles; Table 1).

4. Discussion

The present study showed that eccentric exercise with a supramaximal isotonic load caused DOMS and impaired motor output across different motor tasks. Force output assessed either

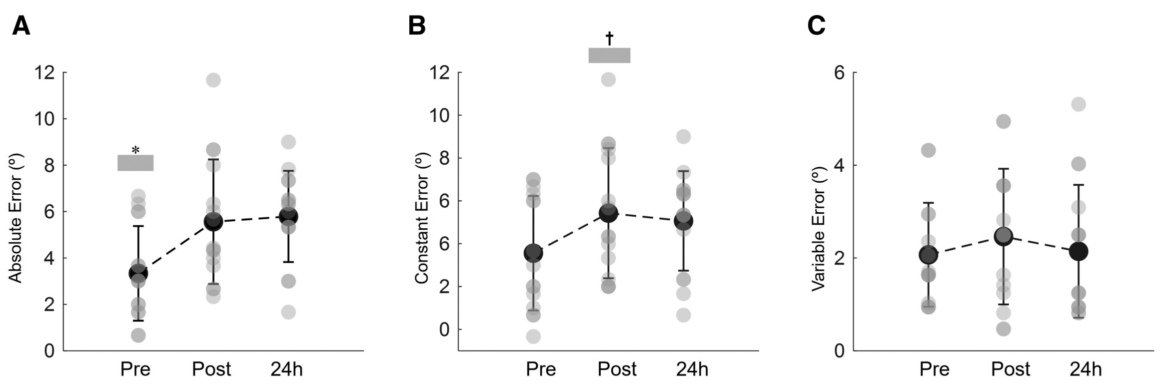

FIGURE 4
Mean and SD of the absolute joint position error (A), constant error (B) and variable error (C) during the knee repositioning task recorded at baseline (pre), immediately post and 24 h post eccentric exercise. *Indicates that absolute joint position error pre-eccentric exercise significantly lower than immediate post and 24 h after eccentric exercise ($P < 0.05$). †Denotes a significant difference between post and pre constant error values ($P < 0.05$).

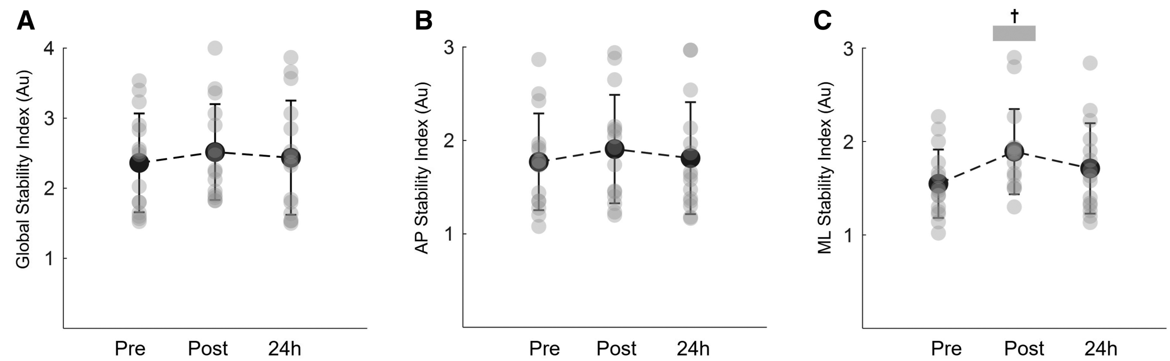

FIGURE 5
Mean and SD of the global stability index (A) anteroposterior stability (AP) index (B) and mediolateral stability (ML) index (C) during the unilateral dynamic balance task recorded at baseline (pre), immediately post and 24 h post eccentric exercise. †Denotes a significant difference between post and pre constant error values ($P < 0.001$).

by MVC, RFD or force steadiness declined immediately after eccentric training. Twenty-four hours later, MVC and RFD remained impaired, while isometric force steadiness, at different target levels, was recovered. These alterations were accompanied by adjustments in agonist muscle activity that followed a different time course and pattern depending on the motor task characteristics. Postural control and proprioception were also impaired by the isotonic eccentric exercise as well. The stability in the mediolateral direction decreased, and the error magnitude of knee positioning increased immediately after exercise and in the presence of DOMS. Collectively, these findings demonstrate that the magnitude and the time course of the impairments induced by eccentric exercise with a supramaximal isotonic vary depending on the motor task specificity.

4.1. Muscle soreness and muscle strength performance

The participants reported soreness in the quadriceps 24 h after training, confirming the presence of DOMS. The average soreness level was 3.6 out of 10, which is in agreement with studies on acute responses to isokinetic eccentric exercise (3, 19, 33). Both MVC and RFD of the knee extensors decreased immediately after isotonic eccentric exercise and remained impaired 24 h later. Nonetheless, the magnitude of decline in the RFD post and 24 h after exercise was greater than in the MVC (20.3% and 15.5% vs. 17.9% and 13.6%, respectively). Several studies have reported similar results, suggesting that RFD is more sensitive to muscle fatigue (34, 35) and muscle damage after eccentric exercise (15, 17, 19, 36). The observed decrease in force-generating capacity is a well-known consequence of performing unaccustomed and/or vigorous eccentric exercise. This type of exercise can lead to disruptions in cytoskeletal structures and microlesions in muscle fibers, thereby affecting the excitation-contraction coupling mechanisms (16, 37). Consequently, a long-lasting deficit in force production (> 24 h) is commonly observed (38, 39). The magnitude of the force production deficit and time course of the recovery seems to be partially determined by the content and size of the muscle fibers type II (4). Several studies have shown that type II fibers are particularly susceptible to damage from intense eccentric exercise, mainly due to their higher tension generating capacity and shorter optimum length for tension (10, 40). Although we did not directly measure muscle fiber damage, it is very likely that the greater changes observed in the RFD are a consequence of subcellular perturbances within type II fibers. Fiber type is often considered a major factor influencing RFD, in which type II fibers play an important role in producing faster rates of tension development (9). Muscle activation is also a critical determinant of RFD, particularly in the earlier phases of the rising force during rapid contractions (41). Given the potential for muscle damage caused by eccentric exercise, neural adaptations may occur as an effort to overcome muscle impairments and generate the required muscle tension. In the current study, the EMG amplitude of the vastus medialis exhibited a significant increase at the onset of rapid contraction and around the time of maximum RFD immediately after eccentric exercise, while no changes were observed in the antagonist activity. Still, EMG amplitude of the vastus medialis returned to baseline levels within 24 h, even though RFD impairment persisted.

The findings indicate that changes in EMG patterns are strongly associated with neuromuscular fatigue, rather than the neuromuscular mechanisms that cause maximum RDF impairments from isotonic eccentric exercise. While several factors could potentially account for the immediate post-eccentric exercise increment in the EMG amplitude, it is highly plausible that these changes are a consequence of an increment of the recruitment and/or firing rate of the motor units (18, 42). Because at the contraction onset and at the instant of maximum RFD, force is submaximal, the neural drive can be increased to compensate the force impairments induced by muscle fatigue. In fact, increased EMG amplitude of the agonist muscle was also observed during the sustained submaximal contractions immediately post, but not in the 24 h after eccentric exercise. The presence of muscle fatigue is also indirectly confirmed by the decline in EMG activity during the MVC. Contrary to submaximal efforts, during maximal contractions, the EMG amplitude typically decreases in the presence of muscle fatigue, indicating the limitations of the central nervous system to continually increase synaptic input to the motoneurons (43).

Peñailillo et al. (17) did not observed any change in the EMG amplitude of the vastus lateralis during explosive contractions following eccentric cycling protocols at 60% of maximal concentric contraction. Conversely, Vila-Chã et al. (19) reported a decline of the EMG amplitude of the vasti muscles at the contraction onset and at the time of maximum RFD, which persisted in the 24 h after a maximal isokinetic eccentric exercise protocol. The conflicting results might be explained by differences arising from variations in the modes of eccentric exercise employed across each study, that might have induced different magnitudes of fatigue and muscle damage. It is well known that both phenomena are dependent of modulation based on task-specific factors, which in turn govern the neurophysiological changes (44, 45). In the current study, the immediate decline in the maximum RFD and MVC after isotonic eccentric exercise is very likely explained by a combination of fatigue and initial muscle damage, leading to changes in muscle activation. However, the sustained alteration in RFD and MVC values observed 24 h after isotonic eccentric exercise appears to be predominantly stem from peripheral adaptations prompted by muscle damage, with negligible impact on the neural drive.

Another important parameter about muscle strength is the force steadiness measured during submaximal contractions. It has been shown that force steadiness during submaximal isometric contractions is moderately associated with performance on different motor tasks, such as standing balance and walking performance (31). Therefore, gaining insight into the influence of isotonic eccentric exercise on force fluctuation can yield valuable functional and physiological information regarding complementary indicators of function recovery after eccentric exercise. Such significance is accentuated by the fact that this parameter is predominantly influenced by neural factors that are not directly associated with MVC or RFD. The present study showed that force steadiness during different submaximal isometric levels significantly decreased immediately after eccentric exercise, but returned to baseline values in the 24 h after exercise. The EMG amplitude of the VM followed a similar pattern and no changes were observed in the activity of the antagonist muscle. The results are in line with previous studies that applied isokinetic eccentric protocols (19, 23, 46). In these studies, the force steadiness of the elbow flexors (23, 46) and knee extensors (19) decreased immediately after eccentric exercise for all submaximal levels (between 2.5% and 50% MVC). Twenty-four hours later the force steadiness remained impaired only during the submaximal contractions at very low intensities ($\leq$ 10% MVC). The observed alterations were accompanied by an increased EMG amplitude of the agonist muscles, indicating the presence of low-frequency fatigue following eccentric exercise (46, 47). In the current study, the lowest intensity used was 10% MVC. Even for this very low intensity, the force steadiness of the knee extensors was recovered 24 h after the eccentric exercise. The findings suggest that the CoV of force is a better indicator of muscle fatigue than potential muscle damage resulting from the isotonic eccentric protocol applied in this study.

4.2. Proprioception and balance

Immediately after isotonic eccentric exercise, an increase of the position errors was observed. The magnitude of the error (measured as absolute error) remained higher after 24 h after exercise, while direction of the error (constant error) and position consistency (variable error) returned to pre-exercise values. The results indicate that the participants matched the predetermined knee position by adopting a more extended position. Similar results have been obtained previously (20–22, 48–50). It is well known that a period of intense exercise increases errors in limb position sense (51). The disturbance in joint position sense seems to arise from the altered sense of effort due to the presence of fatigue, either after concentric or eccentric exercise (24, 52). Moreover, exercise effects on position sense are directly dependent on the level of muscle fatigue produced by the exercise (51). It has been shown that the accumulation of metabolites, stimulation of small muscle afferents in the group III and IV due to muscle fatigue might be involved in the sense of effort, perturbing the sense of position (51, 53). However, following eccentric exercise it was observed that position matching errors might persist in the subsequent days, where chemical products of fatigue were long gone (21, 24, 50). This suggests that factors beyond fatigue affected the position sense. In the past it has been hypothesized that eccentric exercise would cause also damage to muscle spindles, leading to perturbations in proprioception. Nonetheless, animal studies have shown that after intense eccentric exercise the responsiveness of muscle spindles is not disturbed despite extensive muscle damage (51, 54).

Tsay et al. (50) observed an association between de decrease in force and increase position errors, suggesting that exercise-induced decline in force (of 20%–30%) can disrupt limb position sense. It has been suggested that the effects of exercise on proprioception resides on the operation of an internal feed-forward model (53, 55). This hypothesis indicates that the expected sensory feedback for a particular limb position is based on past memories and compares it with the actual sensory feedback from the fatigued limb. Based on previous experience, the sensory feedback from the fatigued muscle might be greater than anticipated from the motor command, altering the neural drive to the fatigued limb (23, 50). Conversely Da Silva et al. (48) observed a better association between position matching errors and voluntary activation deficits than with MVC decline. However, in their study, and despite a long-lasting MVC decrease and the presence of DOMS up to 48 h postexercise, position errors returned to baseline values within in the 24 h after eccentric exercise. In line with Tsay et al. (50) results, in the present study, the magnitude of errors increased immediately after eccentric exercise and remained higher 24 h later, which accompanied the observed changes in MVC and RFD.

Notwithstanding the observed changes in position sense, during the balance task only alterations in the mediolateral stability index were observed. This parameter was increased immediately after isotonic eccentric exercise but returned to pre-exercise values 24 h after. The global and the anteroposterior stability index was not affected by the exercise protocol. Dabbs & Chander (56) investigated the effects of eccentric exercise on balance performance and despite the significant reductions in knee extensor torque up to at least 48 h, no changes were observed during the balance task. The lower sensitivity of the balance task to fatigue and muscle damage induced by the eccentric exercise might be explained by the fact that postural control is not only dependent of the proprioceptive and somatosensory system (57). The maintenance of upright balance is a complex task that involves afferent visual and vestibular systems that are not primary affected by muscle fatigue or muscle damage (58).

Although this study provides insights into the complex effects of supramaximal isotonic eccentric exercise on neuromuscular function, motor performance, and proprioception, it is essential to acknowledge its specific limitations. Firstly, muscle damage induced by the isokinetic eccentric exercise was not directly measured, enabling further investigation of the relationship between muscle damage characteristics and the degree of motor performance impairments across a variety of motor tasks. The study focused on the 24 hours after exercise recovery period. However, extending the follow-up duration could shed light on the longer-term effects and the persistence of neuromuscular alterations induced by eccentric exercise. The evaluation of postural control during a balance task revealed a degree of insensitivity to fatigue and muscle damage in this study. To provide a more comprehensive understanding on the impact of the eccentric exercise on balance, a thorough analysis of additional balance-related parameters would be necessary. Lastly, given the highly specific nature of the eccentric exercise protocol used in this study, it is important to acknowledge that the results are inherently constrained by the confines of the defined experimental protocol. Consequently, the applicability of these findings to broader eccentric exercise regimens or different populations may be limited.

5. Conclusion

The supramaximal isotonic eccentric exercise affected differently the performed motor tasks. When compared to baseline, both, MVC and RFD were impaired immediately after exercise and in the subsequent 24 h. However, the magnitude of changes was more pronounced in the RFD than in the MVC knee extensors profile. Proprioception of knee position was also affected by the isotonic eccentric exercise and remained altered 24 hours after the exercise protocol. The results revealed that both RFD, MVC and position sense are affected by muscle fatigue and muscle damage. Conversely, force steadiness during submaximal contractions and mediolateral stability index during the balance task were only impaired immediately after eccentric exercise, returning to baseline values within 24 h. The current study showed that isotonic eccentric exercise induces specific alterations on different neuromuscular functional outputs. The findings underscore the importance of understanding the specific alterations and recovery timelines associated with eccentric exercise for effective exercise and rehabilitation planning. However, future research needs to be performed to enhance understanding of these phenomena.

Data availability statement

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

Ethics statement

The studies involving humans were approved by Polytechnic of Guarda Committee on Research Ethics. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

CV-C: Conceptualization, Formal Analysis, Methodology, Supervision, Writing – original draft, Writing – review & editing. AB: Conceptualization, Formal Analysis, Methodology, Writing – original draft, Writing – review & editing. CF: Conceptualization, Formal Analysis, Methodology, Writing – original draft. AC-B: Conceptualization, Formal Analysis, Project administration, Writing – review & editing. CV: Formal Analysis, Writing – original draft, Writing – review & editing. MR-A: Conceptualization, Formal Analysis, Methodology, Writing – original draft, Writing – review & editing. JP: Methodology, Writing – original draft, Writing – review & editing. TV: Conceptualization, Formal Analysis, Methodology, Writing – review & editing. GM: Conceptualization, Formal Analysis, Writing – original draft, Writing – review & editing.

Funding

The author(s) declare financial support was received for the research, authorship, and/or publication of this article.

This study was funded by the Portuguese Foundation for Science and Technology, IP, under project UID04045/2020. The Polytechnic of Guarda partly supported the research reported in this publication.

Acknowledgments

ARB and AB are part of the Highly Qualified Human Resources Project, reference number CENTRO-04-3559-FSE-000162. MR-A acknowledges the financial support received from the Spanish Ministry of Universities through the Grants for the Requalification of the Spanish University System under the Postdoctoral Margarita Salas Programme (RSUC.UDC.MS09), funded by the European Union – Next Generation.

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.

The author(s) declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

References

1. Lates AD, Greer BK, Wagle JP, Taber CB. Accentuated eccentric loading and cluster set configurations in the bench press. *J Strength Cond Res.* (2022) 36 (6):1485–9. doi: 10.1519/JSC.0000000000003664

2. Wagle JP, Taber CB, Cunanan AJ, Bingham GE, Carroll KM, DeWeese BH, et al. Accentuated eccentric loading for training and performance: a review. *Sports Med.* (2017) 47(12):2473–95. doi: 10.1007/s40279-017-0755-6

3. Douglas J, Pearson S, Ross A, McGuigan M. Eccentric exercise: physiological characteristics and acute responses. *Sports Med.* (2017b) 47(4):663–75. doi: 10.1007/s40279-016-0624-8

4. Farup J, Rahbek SK, Bjerre J, de Paoli F, Vissing K. Associated decrements in rate of force development and neural drive after maximal eccentric exercise. *Scand J Med Sci Sports.* (2016) 26(5):498–506. doi: 10.1111/sms.12481

5. Guilhem G, Cornu C, Guével A. Neuromuscular and muscle-tendon system adaptations to isotonic and isokinetic eccentric exercise. *Ann Phys Rehabil Med.* (2010) 53(5):319–41. doi: 10.1016/j.rehab.2010.04.003

6. Guilhem G, Cornu C, Maffiuletti NA, Guével A. Neuromuscular adaptations to isoload versus isokinetic eccentric resistance training. *Med Sci Sports Exerc.* (2013) 45(2):326–35. doi: 10.1249/MSS.0b013e31826e7066

7. Douglas J, Pearson S, Ross A, McGuigan M. Chronic adaptations to eccentric training: a systematic review. *Sports Med.* (2017a) 47(5):917–41. doi: 10.1007/s40279-016-0628-4

8. Vogt M, Hoppeler HH. Eccentric exercise: mechanisms and effects when used as training regime or training adjunct. *J Appl Physiol.* (2014) 116(11):1446–54. doi: 10.1152/japplphysiol.00146.2013

9. Harridge SDR, Bottinelli R, Canepari M, Pellegrino MA, Reggiani C, Esbjörnsson M, et al. Whole-muscle and single-fibre contractile properties and myosin heavy chain isoforms in humans. *Pflug Arch Eur J Phy.* (1996) 432(5):913–20. doi: 10.1007/s004240050215

10. Jones DA, Newham DJ, Round JM, Tolfree SE. Experimental human muscle damage: morphological changes in relation to other indices of damage. *J Physiol.* (1986) 375(1):435–48. doi: 10.1113/jphysiol.1986.sp016126

11. Allen DG, Lamb GD, Westerblad H. Skeletal muscle fatigue: cellular mechanisms. *Physiol Rev.* (2008) 88(1):287–332. doi: 10.1152/physrev.00015.2007

12. Baumert P, Lake MJ, Stewart CE, Drust B, Erskine RM. Genetic variation and exercise-induced muscle damage: implications for athletic performance, injury and ageing. *Eur J Appl Physiol.* (2016) 116(9):1595–625. doi: 10.1007/S00421-016-3411-1

13. Friden J, Sjostrom M, Ekblom B. Myofibrillar damage following intense eccentric exercise in man. *Int J Sports Med.* (1983) 4(3):170–6. doi: 10.1055/s-2008-1026030

14. Goodall S, Thomas K, Barwood M, Keane K, Gonzalez JT, St Clair Gibson A, et al. Neuromuscular changes and the rapid adaptation following a bout of damaging eccentric exercise. *Acta Physiol.* (2017) 220(4):486–500. doi: 10.1111/apha.12844

15. Alemany JA, Delgado-Díaz DC, Mathews H, Davis JM, Kostek MC. Comparison of acute responses to isotonic or isokinetic eccentric muscle action: differential outcomes in skeletal muscle damage and implications for rehabilitation. *Int J Sports Med.* (2014) 35(1):1–7. doi: 10.1055/s-0032-1327652

16. Proske U, Allen TJ. Damage to skeletal muscle from eccentric exercise. *Exerc Sport Sci Rev.* (2005) 33(2):98–104. doi: 10.1097/00003677-200504000-00007

17. Peñailillo L, Blazevich A, Numazawa H, Nosaka K. Rate of force development as a measure of muscle damage. *Scand J Med Sci Sports.* (2015) 25(3):417–27. doi: 10.1111/SMS.12241

18. Dartnall TJ, Rogasch NC, Nordstrom MA, Semmler JG. Eccentric muscle damage has variable effects on motor unit recruitment thresholds and discharge patterns in elbow flexor muscles. *J Neurophysiol.* (2009) 102(1):413–23. doi: 10.1152/jn.91285.2008

19. Vila-Chã C, Hassanlouei H, Farina D, Falla D. Eccentric exercise and delayed onset muscle soreness of the quadriceps induce adjustments in agonist-antagonist activity, which are dependent on the motor task. *Exp Brain Res.* (2012) 216 (3):385–95. doi: 10.1007/s00221-011-2942-2

20. Torres R, Vasques J, Duarte JA, Cabri JMH. Knee proprioception after exercise-induced muscle damage. *Int J Sports Med.* (2010) 31(06):410–5. doi: 10.1055/s-0030-1248285

21. Vila-Chã C, Riis S, Lund D, Møller A, Farina D, Falla D. Effect of unaccustomed eccentric exercise on proprioception of the knee in weight and non-weight bearing tasks. *J Electromyogr Kinesiol.* (2011) 21(1):141–7. doi: 10.1016/j.jelekin.2010.10.001

22. Givoni NJ, Pham T, Allen TJ, Proske U. The effect of quadriceps muscle fatigue on position matching at the knee. *J Physiol.* (2007) 584(1):111–9. doi: 10.1113/jphysiol.2007.134411

23. Semmler JG, Tucker KJ, Allen TJ, Proske U. Eccentric exercise increases EMG amplitude and force fluctuations during submaximal contractions of elbow flexor muscles. *J Appl Physiol.* (2007) 103(3):979–89. doi: 10.1152/japplphysiol.01310.2006

24. Walsh LD, Allen TJ, Gandevia SC, Proske U. Effect of eccentric exercise on position sense at the human forearm in different postures. *J Appl Physiol.* (2006) 100(4):1109–16. doi: 10.1152/japplphysiol.01303.2005

25. Haff GG, Triplett NT. *Essentials of strength training and conditioning 4th edition.* Champaign, IL: Human kinetics (2015).

26. Lombardi VP. *Beginning weight training: The safe and effective way.* Dubuque: Wm. C. Brown Company Publishers (1989).

27. Biodex Medical Systems, I. Balance System SD Operation/Service Manual (Shirley, N, Vol. 6339). www.biodex.com (2011).

28. Cachupe WJC, Shifflett B, Kahanov L, Wughalter EH. Reliability of biodex balance system measures. *Meas Phys Educ Exerc Sci.* (2001) 5(2):97–108. doi: 10.1207/S15327841MPEE0502_3

29. Hermens H, Freriks B, Merletti R, Stegeman DF, Blok J, Rau G, et al. *European recommendations for surface electromyography. Results of the SENIAM project. Roessingh Research and Development.* Enschede: Roessingh Research and Development (1999).

30. Vincent L, Soille P. Watersheds in digital spaces: an efficient algorithm based on immersion simulations. *IEEE Trans Pattern Anal Mach Intell.* (1991) 13(06):583–98. doi: 10.1109/34.87344

31. Enoka RM, Farina D. Force steadiness: from motor units to voluntary actions. *Physiology.* (2021) 36(2):114–30. doi: 10.1152/physiol.00027.2020

32. Cohen J. Statistical power analysis Jbr the behavioral. *Sciences.* (1988):18–74. doi: 10.4324/9780203771587

33. Hedayatpour N, Hassanlouei H, Arendt-Nielsen L, Kersting UG, Falla D. Delayed-onset muscle soreness alters the response to postural perturbations. *Med Sci Sports Exercise*. (2011) 43(6):1010–6. doi: 10.1249/MSS.0b013e3182058628

34. Andersen LL, Andersen JL, Zebis MK, Aagaard P. Early and late rate of force development: differential adaptive responses to resistance training? *Scand J Med Sci Sports*. (2010) 20(1):e162–9. doi: 10.1111/j.1600-0838.2009.00933.x

35. D'Emanuele S, Maffiuletti NA, Tarperi C, Rainoldi A, Schena F, Boccia G. Rate of force development as an indicator of neuromuscular fatigue: a scoping review. *Front Hum Neurosci*. (2021) 15:701916. doi: 10.3389/fnhum.2021.701916

36. Crameri RM, Aagaard P, Qvortrup K, Langberg H, Olesen J, Kjaer M. Myofibre damage in human skeletal muscle: effects of electrical stimulation *versus* voluntary contraction. *J Physiol*. (2007) 583(1):365–80. doi: 10.1113/jphysiol.2007.128827

37. Prasartwuth O, Taylor JL, Gandevia SC. Maximal force, voluntary activation and muscle soreness after eccentric damage to human elbow flexor muscles. *J Physiol*. (2005) 567(1):337–48. doi: 10.1113/jphysiol.2005.087767

38. Clarkson PM, Nosaka K, Braun B. Muscle function after exercise-induced muscle damage and rapid adaptation. *Med Sci Sports Exerc*. (1992) 24(5):512–20.

39. Butterfield TA. Eccentric exercise in vivo: strain-induced muscle damage and adaptation in a stable system. *Exerc Sport Sci Rev*. (2010) 38(2):51–60. doi: 10.1097/JES.0b013e3181d496eb

40. Vijayan K, Thompson JL, Norenberg KM, Fitts RH, Riley DA. Fiber-type susceptibility to eccentric contraction-induced damage of hindlimb-unloaded rat AL muscles. *J Appl Physiol*. (2001) 90(3):770–6. doi: 10.1152/JAPPL.2001.90.3.770

41. Folland JP, Buckthorpe MW, Hannah R. Human capacity for explosive force production: neural and contractile determinants. *Scand J Med Sci Sports*. (2014) 24 (6):894–906. doi: 10.1111/sms.12131

42. Proske U, Morgan DL. Muscle damage from eccentric exercise: mechanism, mechanical signs, adaptations and clinical applications. *J Physiol*. (2001) 537 (2):333–45. doi: 10.1111/j.1469-7793.2001.00333.x

43. Gandevia SC. Spinal and supraspinal factors in human muscle fatigue. *Physiol Rev*. (2001) 81(4):1725–89. doi: 10.1152/physrev.2001.81.4.1725

44. Enoka RM, Stuart DG. Neurobiology of muscle fatigue. *J Appl Physiol*. (1992) 72 (5):1631–48. doi: 0161-7567/92

45. Prasartwuth O, Allen TJ, Butler JE, Gandevia SC, Taylor JL, Prasartwuth O. Length-dependent changes in voluntary activation, maximum voluntary torque and twitch responses after eccentric damage in humans. *J Physiol*. (2006) 571(Pt 1):243–52. doi: 10.1113/jphysiol.2005.101600

46. Dartnall TJ, Nordstrom MA, Semmler JG. Motor unit synchronization is increased in Biceps brachii after exercise-induced damage to elbow flexor muscles. *J Neurophysiol*. (2008) 99(2):1008–19. doi: 10.1152/jn.00686.2007

47. Dundon JM, Cirillo J, Semmler JG. Low-frequency fatigue and neuromuscular performance after exercise-induced damage to elbow flexor muscles. *J Appl Physiol*. (2008) 105(4):1146–55. doi: 10.1152/japplphysiol.01339.2007

48. Da Silva F, Monjo F, Zghal F, Chorin F, Guérin O, Colson SS. Altered position sense after submaximal eccentric exercise-inducing central fatigue. *Med Sci Sports Exercise*. (2021) 53(1):218–27. doi: 10.1249/MSS.0000000000002444

49. Paschalis V, Nikolaidis MG, Giakas G, Jamurtas AZ, Pappas A, Koutedakis Y. The effect of eccentric exercise on position sense and joint reaction angle of the lower limbs. *Muscle Nerve*. (2007) 35(4):496–503. doi: 10.1002/mus.20723

50. Tsay A, Allen TJ, Leung M, Proske U. The fall in force after exercise disturbs position sense at the human forearm. *Exp Brain Res*. (2012) 222(4):415–25. doi: 10.1007/s00221-012-3228-z

51. Proske U. Exercise, fatigue and proprioception: a retrospective. *Exp Brain Res*. (2019) 237(10):2447–59. doi: 10.1007/s00221-019-05634-8

52. Brockett C, Warren N, Gregory JE, Morgan DL, Proske U. A comparison of the effects of concentric versus eccentric exercise on force and position sense at the human elbow joint. *Brain Res*. (1997) 771(2):251–8. doi: 10.1016/S0006-8993(97)00808-1

53. Allen TJ, Ansems GE, Proske U. Effects of muscle conditioning on position sense at the human forearm during loading or fatigue of elbow flexors and the role of the sense of effort. *J Physiol*. (2007) 580(2):423–34. doi: 10.1113/jphysiol.2006.125161

54. Gregory JE, Morgan DL, Proske U. Response of muscle spindles following a series of contractions. *Exp Brain Res*. (2004) 157(2):234–40. doi: 10.1007/S00221-004-1838-9

55. Bays PM, Wolpert DM. Computational principles of sensorimotor control that minimize uncertainty and variability. *J Physiol*. (2007) 578(2):387–96. doi: 10.1113/jphysiol.2006.120121

56. Dabbs NC, Chander H. The effects of exercise induced muscle damage on knee joint torque and balance performance. *Sports (Basel)*. (2018) 6(3):101. doi: 10.3390/sports6030101

57. Deliagina TG, Zelenin PV, Orlovsky GN. Physiological and circuit mechanisms of postural control. *Curr Opin Neurobiol*. (2012) 22(4):646–52. doi: 10.1016/j.conb.2012.03.002

58. Caron O. Effects of local fatigue of the lower limbs on postural control and postural stability in standing posture. *Neurosci Lett*. (2003) 340(2):83–6. doi: 10.1016/S0304-3940(02)01455-6

TYPE Systematic Review
PUBLISHED 17 November 2023
DOI 10.3389/fspor.2023.1197883

OPEN ACCESS

EDITED BY
Jaap Van Dieen,
VU Amsterdam, Netherlands

REVIEWED BY
Kaushik Mukherjee,
Indian Institute of Technology Delhi, India
Michael Samaan,
University of Kentucky, United States

*CORRESPONDENCE
Hannah Steingrebe
✉ hannah.steingrebe@kit.edu

RECEIVED 31 March 2023
ACCEPTED 09 October 2023
PUBLISHED 17 November 2023

CITATION
Steingrebe H, Spancken S, Sell S and Stein T (2023) Effects of hip osteoarthritis on lower body joint kinematics during locomotion tasks: a systematic review and meta-analysis.
Front. Sports Act. Living 5:1197883.
doi: 10.3389/fspor.2023.1197883

Effects of hip osteoarthritis on lower body joint kinematics during locomotion tasks: a systematic review and meta-analysis

Hannah Steingrebe[1,2]*, Sina Spancken[1], Stefan Sell[2,3] and Thorsten Stein[1]

[1]BioMotion Center, Institute of Sports and Sports Science, Karlsruhe Institute of Technology (KIT), Karlsruhe, Germany, [2]Sports Orthopedics, Institute of Sports and Sports Science, Karlsruhe Institute of Technology (KIT), Karlsruhe, Germany, [3]Joint Center Black Forest, Hospital Neuenbürg, Neuenbürg, Germany

Introduction: Motion analysis can be used to gain information needed for disease diagnosis as well as for the design and evaluation of intervention strategies in patients with hip osteoarthritis (HOA). Thereby, joint kinematics might be of great interest due to their discriminative capacity and accessibility, especially with regard to the growing usage of wearable sensors for motion analysis. So far, no comprehensive literature review on lower limb joint kinematics of patients with HOA exists. Thus, the aim of this systematic review and meta-analysis was to synthesise existing literature on lower body joint kinematics of persons with HOA compared to those of healthy controls during locomotion tasks.
Methods: Three databases were searched for studies on pelvis, hip, knee and ankle kinematics in subjects with HOA compared to healthy controls during locomotion tasks. Standardised mean differences were calculated and pooled using a random-effects model. Where possible, subgroup analyses were conducted. Risk of bias was assessed with the Downs and Black checklist.
Results and Discussion: A total of 47 reports from 35 individual studies were included in this review. Most studies analysed walking and only a few studies analysed stair walking or turning while walking. Most group differences were found in ipsi- and contralateral three-dimensional hip and sagittal knee angles with reduced ranges of motion in HOA subjects. Differences between subjects with mild to moderate and severe HOA were found, with larger effects in severe HOA subjects. Additionally, stair walking and turning while walking might be promising extensions in clinical gait analysis due to their elevated requirements for joint mobility. Large between-study heterogeneity was observed, and future studies have to clarify the effects of OA severity, laterality, age, gender, study design and movement execution on lower limb joint kinematics.

Systematic Review Registration: PROSPERO (CRD42021238237).

KEYWORDS
hip osteoarthritis, motion analysis, kinematics, gait, stair walking, turning, locomotion, clinical gait analysis

1. Introduction

Hip osteoarthritis (HOA) is a common joint disease with high prevalence especially in the elderly (1). Due to pain and limited function HOA has a strong impact on the quality of life of the affected persons (2).

Data from instrumented movement analysis can be helpful for disease diagnosis as well as for the design and evaluation of patient-specific intervention strategies in HOA populations (3, 4). Depending on the data required, motion analysis can be obtained at different levels of complexity. While temporal-spatial gait characteristics are relatively easy to record (5) they might contain only limited informative value. Analysis of joint dynamics, first, requires simultaneous recording of ground reaction force data and, secondly, advanced methods of modelling (6). Due to the complexity and large time requirements, the applicability in the clinical setting might be limited (7). In 2018, Diamond et al. (8) reviewed the existing literature on external hip flexion and adduction moments in subjects with HOA. They found a reduction of both moments in patients with severe HOA but not in those with mild to moderate symptoms. Hence, adjustments of joint loading might take place at a later disease stage and thus might not be suitable for the early diagnosis of HOA. Similarly, Emmerzaal and colleagues (9) found lower or only slightly higher classification results between HOA and healthy subjects when joint dynamics combined with joint kinematics were used as input variables compared to kinematic variables only.

Therefore, analysis of joint kinematics might be the right balance between information acquisition and feasibility, especially in the clinical context. Thereby, wearable sensors' increasing dissemination and popularity for the analysis of joint kinematics has also to be considered. With the availability of easy-to-use sensors and robust software tools for the estimation of joint angles, widespread usage of movement analysis in the clinical setting becomes feasible (10). In the past, several studies have shown kinematic changes at the hip joint in persons with varying degrees of HOA (11–13). Additionally, previous studies have shown that changes in gait patterns not only concern the hip joint itself but also the knee (14), ankle (15), pelvis (16) and upper body kinematics (17). Therefore, kinematics in general are of great interest for the characterisation of gait changes associated with HOA.

Gait, as one of the most fundamental movements of daily life, has often been used in the analysis of disease effects on movement kinematics, and previous reviews (4, 8, 18) focused on gait deviations caused by HOA. However, other locomotion tasks, such as stair ascent and descent might impose higher demands regarding the required range of motion (19), and thus their analysis might add essential insights into the movement restrictions and adaptations of people with HOA. A recently published article has shown that the most accurate classification between HOA subjects and healthy controls was done using joint angle data from stair walking trials compared to walking or stationary tasks such as lunges etc. (9).

Biomechanical gait analysis studies are often very time consuming (3, 4), and therefore mostly include only limited sample sizes (8). Additionally, subjects with HOA show heterogeneous disease characteristics such as varying degrees of functional and radiographic disease severity, uni- or bilateral involvement as well as primary or secondary HOA cause (8). An aggregation of multiple studies and collective evaluation of the results as well as the evaluation of specific subgroups might therefore add essential insights on the impact of HOA on movement biomechanics.

However, to the best of our knowledge, no comprehensive review exists regarding the kinematic changes observed during locomotion movements in the presence of HOA.

Therefore, the objective of the present systematic review was to summarise the current state of research on lower-limb joint kinematics during locomotion movements, such as gait or stair walking, in subjects diagnosed with HOA compared to healthy controls. Where possible a conjoint analysis of previous results in terms of a meta-analysis was performed, allowing special attention to be paid to the influence of HOA severity and uni- or bilateral involvement.

2. Materials and methods

This review was conducted according to the Preferred Reporting Items for Systematic Review and Meta-Analysis Statement (PRISMA) and registered in the International Prospective Register of Systematic Reviews (PROSPERO, no. CRD42021238237).

2.1. Information sources, search strategy and screening process

Eligible reports were searched in three electronic databases [PubMed (MEDLINE, PubMed Central, and additional PubMed records), Web of Science and Scopus] on August 2nd 2022.

The title, abstract and keywords were screened for disease description (Coxarthr*, "degenerative joint" AND hip, hip AND osteoarthr*), the outcome parameter (kinematic*, angle*, "range of motion", mobility, pattern, goniometric*, biomechanic*) and the movement task (gait, walk*, locomotion, ambulat*, stair*, movement). Titles including *fracture*, *perthes*, *amputee*, *rheumat**, *arthroscopy or arthroplasty* were excluded. The detailed search term for each database can be found in the **Supplementary Material**.

All records from the databases were imported into a reference manager (Citavi 6, Swiss Academic Software GmbH, Switzerland) and duplicates were removed. The reference lists of included articles were screened manually for additional eligible studies.

The literature search, title and abstract screening as well as full text analysis for eligibility were performed individually by two researchers (HS & SSp). Discrepancies were resolved by discussion between the two authors, and if no consensus was reached a third researcher (TS) was consulted.

2.2. Eligibility criteria

Original research studies written in English, evaluating pelvis or lower extremity (hip, knee, ankle) joint kinematics during locomotion movements (e.g., level walking, running or stair walking) in a cohort of subjects with HOA were eligible for this review. No restrictions were made regarding the OA severity, unilateral or bilateral involvement or whether the HOA was primary or secondary. Data from the HOA subjects had to be compared to a healthy control group (CON). Accepted outcome parameters were a quantitative description of lower body joint angle parameters (mean/median with measure of dispersion) and/or the presence of *p*-values from the comparison of lower body joint angle parameters. Studies with an intervention were included using the pre-intervention data if applicable. A detailed description of the eligibility criteria can be found in **Table 1**.

2.3. Data collection process

One researcher (HS) extracted the data from the retrieved reports using a predefined spreadsheet including the following sections: study design, number of HOA and CON subjects, participant characteristics (i.e., age, gender, weight, height, body mass index, radiographic disease severity, functional disease severity, uni-/bilateral involvement, primary/secondary OA), analysed movements (e.g., walking, stair climbing) and testing conditions (treadmill or overground, number of stairs, prescribed or self-selected movement velocity), measurement system and assessed joints (e.g., hip sagittal plane). Extracted data were randomly cross-checked by a second researcher (SSp).

The extracted data were analysed for indications of multiple reports from the same study. In case of doubt, report authors were contacted for clarification. Extracted data of the kinematic variables were summarised in tables and grouped according to planes of motion and joint location.

Disease severity was categorised as mild, moderate or severe based on the information of the reports. End-stage HOA as well as subjects scheduled for total hip replacement were classified as severe.

2.4. Synthesis methods

Peak angles were converted if necessary to obtain a unified definition of the angle direction. If both limbs were reported for the CON subjects, data from the right limb were used. One study (20) reported joint angles of two different examiners to calculate inter-rater reliability. For the meta-analysis, values of examiner 1 were extracted.

Confidence intervals for group means were transformed to standard deviations following the Cochrane handbook (21).

If mean and standard deviation were available, standardised mean differences (SMD) and 95% confidence intervals (95% CI) were calculated for all variables by dividing the difference between groups by the pooled standard deviation (effect size Cohen's d). Where possible (≥2 studies) pooled effect sizes were calculated using a random-effects model with a restricted maximum likelihood estimator and Knapp-Hartung adjustment. The random-effects model was applied to account for variability in the composition of the subject groups (e.g., HOA severity, gender etc.) and the movement execution (e.g., movement speed). SMDs and pooled SMDs ≥0.2 were interpreted as a small effect, ≥0.5 as a moderate effect and ≥0.8 as a large effect (22). Statistical heterogeneity was evaluated from pooled data using the I^2 statistic, with a value of 25% considered low, 50% considered moderate and 75% considered a high level of heterogeneity (23). Additionally, prediction intervals were reported (24, 25). Subgroup analyses for HOA severity and laterality (uni-/bilateral) were conducted if data from ≥2 studies for ≥2 clearly distinguishable subgroups were available. If multiple HOA subgroups from the same study were included, the sample size of the CON group was split equally to all HOA subgroups. For all analyses, the significance level was set *a priori* to $a < 0.05$. The meta-analysis (including calculation of I^2 and prediction intervals) was conducted in R (version 4.2.2) using the meta package.

For several studies, multiple effect sizes were available due to the presence of multiple reports including equal or overlapping subject samples, analyses of subgroups (e.g., men and women) or analyses of different movement conditions (e.g., walking speeds). In those cases, where data of independent subgroups was presented (e.g., OA severity, gender), we recreated the summary data (weighted mean & combined standard deviation) prior to effect size calculation (26). If the presented data were of dependent subgroups (e.g., multiple reports for one study, multiple walking speeds analysed), one study was selected and included in the meta-analysis based on the following criteria: (I) largest overall sample size, (II) movement condition most similar to other studies (27), (III) self-selected gait speed.

Data from studies that could not be synthesised in a meta-analysis were synthesised qualitatively.

TABLE 1 Description of the inclusion and exclusion criteria following a modified PECO scheme.

	Inclusion criteria	Exclusion criteria
Population	Humans with hip osteoarthritis (uni- & bilateral)	Subjects with other diseases (e.g. hip dysplasia, knee OA) or endoprosthesis
Exposure	Analysis of a locomotion tasks (e.g. gait, stair walking)	Stationary movements (e.g. sit to stand, one-leg stand), use of walking aids or handrails
Comparator	Healthy control group	No control group or contralateral limb as control
Outcome	Report of kinematic data on lower body joint angles (hip, knee, ankle) or pelvis movement	Sole report of temporal-spatial-parameters or other kinematic data (e.g. toe-out angles), or analyses of kinematic coordination (whole body analysis, coupling angles, asymmetry measures)
Language	English language	Other languages
Format	Original full text paper	Reviews, conferences proceedings, case studies etc.

2.5. Risk of bias & quality of reporting

Risk of bias and quality of reporting of all included studies were assessed individually by two researchers (HS & SSp) using the checklist created by Downs and Black (28). Of the 27 items, 11 concerning interventions were removed (items 4, 8. 9, 13, 14, 15, 17, 19, 23, 24, 26) as previously done (8). This resulted in a maximum score of 17, with higher scores representing a lower risk of bias. For question 5, age, weight or BMI and gender were defined as principal confounders. Question 27 was answered yes if an *a priori* or *post-hoc* power analysis was reported. Disagreements in initial ratings were discussed by HS and SSp to reach consensus. If no consensus was reached a third reviewer (TS) was consulted.

3. Results

The process of study identification and screening is displayed in **Figure 1**. In total, 47 reports representing 35 independent studies met the inclusion criteria and are reviewed below. Details of all included reports are listed in **Tables 2, 3**. In total, the studies included 949 subjects with HOA and 886 CON subjects. For 3.8% of the HOA subjects and 2.1% of the CON subjects, gender was not reported. Of the remaining subjects, 56.7% of HOA subjects and 62.7% of CON subjects were female.

Only 1 study included subjects with mild HOA, 7 with mild to moderate HOA, 2 with moderate HOA, 2 studies subjects with moderate to severe HOA and 21 with severe HOA. In 3 studies subjects with varying degrees of HOA were included and 2 studies gave no details on HOA severity.

In 20 studies subjects with unilateral HOA were included and 2 studies included those with bilateral HOA. Mixed samples were included in 5 studies and 9 studies did not give information on limb involvement. In 3 studies subjects were diagnosed with primary HOA. For all other studies, no information regarding primary or secondary OA was available.

Gait movement was analysed in 33 studies, of which 28 analysed overground walking, while 5 studies analysed walking on a treadmill. Two studies analysed stair ascent and descent and 1 study analysed turning while walking.

Hip kinematics were analysed in 31 studies, knee kinematics in 18 studies, ankle kinematics in 13 studies and pelvis kinematics in 16 studies. Thereof, 10 studies (hip and knee), 5 studies (ankle) and 2 studies (pelvis) also analysed contralateral joint kinematics.

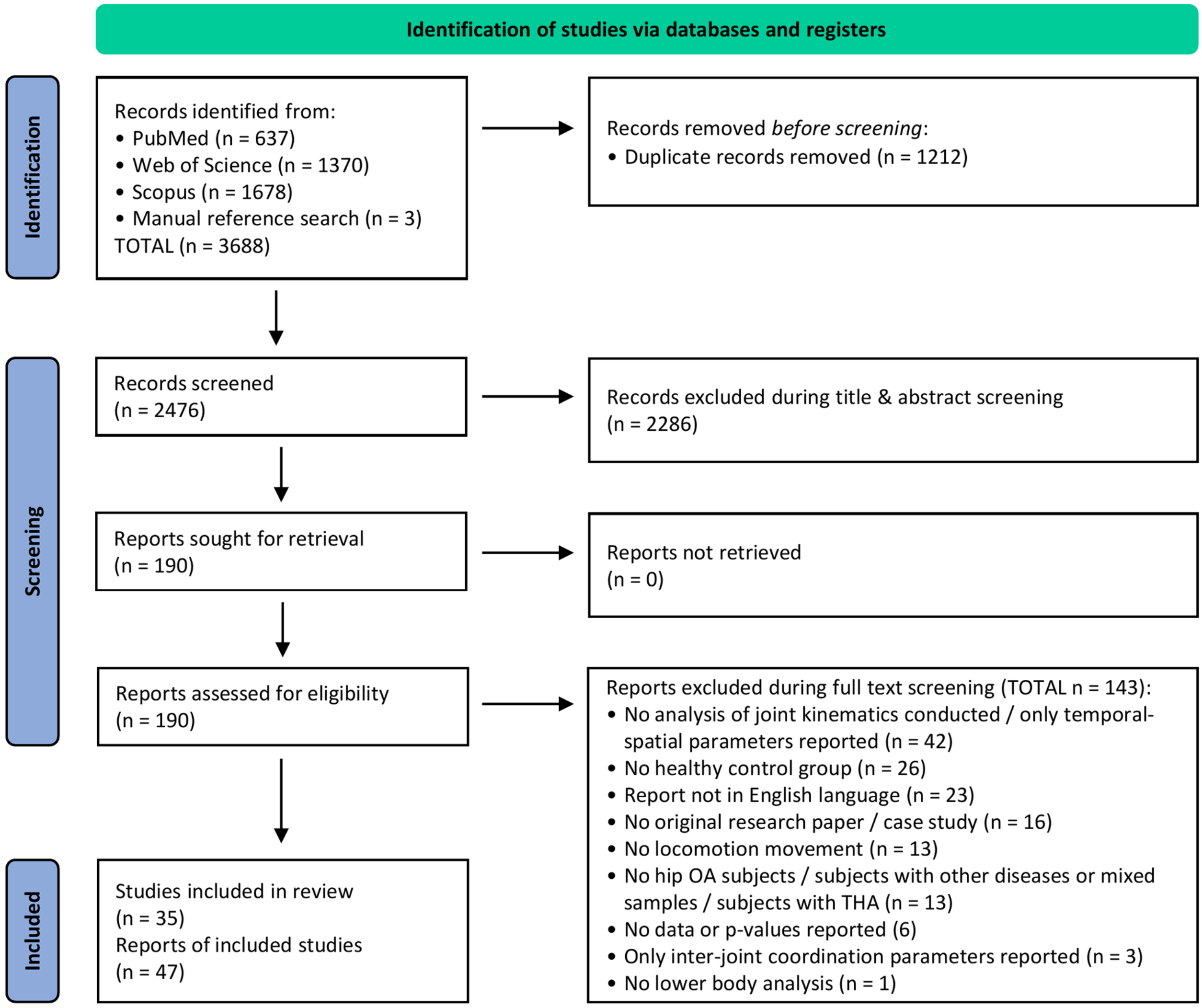

FIGURE 1
PRISMA flow diagram (29).

TABLE 2 Detailed population characteristics of the included studies. All data are presented as mean ± standard deviation.

Study	Report	Population characteristics						
		Group	n (♀)	age [years]	height [cm]	weight [kg]	BMI [kg/m^2]	OA characteristics (severity, laterality, cause)
A	Aminian et al. (30)	CON	9 (–)	63 ± 4	161 ± 10	63 ± 9		Severe; unilateral
		HOA	11 (–)	60 ± 9	167 ± 5	74 ± 8		
B	Ardestani & Wimmer (31)	CON	23 (13)	58.2 ± 9.7			27.98 ± 3.9	Mild, Moderate, Severe (Subgroups); unilateral
		HOA	45 (20)	59 ± 11				
C	Baker et al. (32)	CON	20 (–)	63 ± 6	172 ± 9	76.0 ± 18.4	25.6 ± 5.0	Moderate, Severe (Subgroups); unilateral
		HOA	36 (–)	61.2 ± 3.8	170.4 ± 2.7	84.2 ± 4.1	28.9 ± 2.0	
	Rutherford et al. (14)	CON	20 (10)	63 ± 6	172 ± 9	76 ± 18.4	25.6 ± 5.1	Moderate; unilateral
		HOA	20 (5)	59 ± 8	170 ± 8	83 ± 16.8	28.7 ± 4.3	
	Rutherford et al. (33)	CON	20 (–)	62 ± 6			25.6 ± 5	Moderate, Severe (Subgroups); unilateral
		HOA	37 (–)	60.8 ± 3.4			29.3 ± 2.2	
D	Bejek et al. (34)	CON	20 (12)	68.8 ± 9.1	169 ± 19	73.3 ± 11.4		Severe; unilateral
		HOA	20 (12)	69.7 ± 8.9	172 ± 11	70.1 ± 9.1		
E	Benedetti et al. (35)	CON	10 (4)	59				Severe; -
		HOA	8 (3)	48.7	173	78.5		
F	Bolink et al. (36)	CON	20 (11)	61.0 ± 6.1	173 ± 8.4	77.2 ± 12.7	25.8 ± 3.0	Severe; unilateral
		HOA	20 (10)	63.4 ± 8.5	172 ± 9.7	81.1 ± 17.8	27.2 ± 4.9	
G	Brand & Crowninshield (37)	CON	15 (7)	71				
	(CON data: Crowninshield et al.) (38)	HOA	8 (–)					
H	Constantinou et al. (39)	CON	26 (18)	59.3 ± 7.6	169 ± 8	70.5 ± 9.3	24.8 ± 3.0	Mild/Moderate; mixed (11 bi, 16 uni)
		HOA	27 (18)	63.2 ± 7.6	166 ± 9	77.6 ± 14.2	28.0 ± 4.1	
	Diamond et al. (13)	CON	23 (17)	60 ± 8	167 ± 8	69.9 ± 9.5	25.1 ± 3.1	Mild/Moderate; mixed (9 bi, 9 uni)
		HOA	18 (13)	65 ± 7	166 ± 10	76.2 ± 14.1	27.6 ± 4.8	
I	Eitzen et al. (11)	CON	22 (13)	58.5 ± 8.8	171.7 ± 10.8	70.8 ± 15.1	23.8 ± 3.5	Mild/Moderate; mixed (10 bi, 38 uni)
		HOA	48 (29)	59.1 ± 9.5	172.3 ± 8.4	73.2 ± 12.2	24.6 ± 3.3	
J	Foucher (40)	CON	159 (104)	55.7 ± 2.9			26.7 ± 2.3	Mixed; -
		HOA	150 (86)	62.3 ± 3.4			28.3 ± 2.3	
K	Foucher et al. (41)	CON	25 (11)	57.6 ± 7.7	173 ± 9	79.2 ± 17.8		Severe; unilateral
		HOA	28 (10)	63.6 ± 7.1	172 ± 11	86.7 ± 15.8		
L	Foucher & Wimmer (42)	CON	25 (–)	54 ± 6			28 ± 6	Severe; unilateral; primary
		HOA	26 (–)	59 ± 9			27 ± 3	
	Foucher et al. (43)	CON	25 (13)	54 ± 6	171 ± 6.9	81.6 ± 13.3	28 ± 6	Severe; unilateral; primary
		HOA	26 (11)	60 ± 4.2	175.5 ± 6.3	83.9 ± 5.9	27 ± 3	
M	Hall et al. (44)	CON	15 (11)	64.0 ± 8.7	163 ± 11	72.6 ± 11.6	23.8 ± 3.6	Mixed; mixed (9 bi, 6 uni)
		HOA	15 (8)	63.1 ± 6.7	168 ± 11	76.0 ± 11.8	28.1 ± 4.0	
N	Hara et al. (45)	CON	6 (0)	33	173	67		Severe; -
	(CON data: Hara et al. (46)	HOA	14 (12)	65 ± 7	154 ± 8	54 ± 9	23 ± 4	
O	Hurwitz et al. (47)	CON	19 (7)	61 ± 8	173 ± 10	72.4 ± 15.4		Severe; unilateral
		HOA	19 (7)	60 ± 8	170 ± 10	79.2 ± 11.2		
P	Ismailidis et al. (48)	CON	48 (30)	66.6 ± 7.2	169 ± 9	71.6 ± 12.5	25.1 ± 4.0	Severe; unilateral
		HOA	24 (10)	66.1 ± 10.3	171 ± 8	80.8 ± 13.3	27.5 ± 3.2	
	Ismailidis et al. (49)	CON	45 (29)	66.6 ± 7.4	169 ± 8	71.0 ± 11.9	25.0 ± 4.1	Severe; unilateral
		HOA	22 (10)	66.3 ± 10.2	171 ± 8	79.7 ± 13.2	27.3 ± 3.3	
	Nüesch et al. (50)	CON	54 (23)	66.4 ± 7.9	170 ± 9	73.1 ± 13.5	25.3 ± 4.0	Severe; unilateral
		HOA	30 (18)	64.9 ± 11.6	172 ± 8	80.6 ± 12.8	27.1 ± 3.1	
Q	Kataoka et al. (51)	CON	15 (15)	61.2 ± 6.3	155.8 ± 3.7	53.5 ± 7.3		Moderate/Severe; -
		HOA	15 (15)	60.4 ± 9.6	152.8 ± 2.9	57.1 ± 11.4		
R	Kubota et al. (52)	CON	12 (12)	64.3 ± 2.8	148.1 ± 5.1	52.5 ± 7.8	23.9 ± 2.8	Moderate/Severe; bilateral
		HOA	12 (12)	59.4 ± 11.1	150.6 ± 6.1	56.7 ± 7.1	24.9 ± 3.4	
S	Kumar et al. (12)	CON	30 (16)	48.2 ± 11.4			23.3 ± 3.3	Mild/Moderate; mixed
		HOA	36 (12)	54.5 ± 8.9			24.5 ± 3.0	
T	Leigh et al. (15)	CON	22 (13)	53.7 ± 8.3	168.8 ± 9.5	76.5 ± 9.4	26.8 ± 1.5	Mild/Moderate; mixed (6 bi, 16 uni)
		HOA	22 (12)	55.9 ± 7.5	170.2 ± 8.0	76.8 ± 15.4	26.5 ± 4.6	
U	Meyer et al. (53)	CON	17 (8)	52.7 ± 4.9	171 ± 10		24.1 ± 2.7	Severe; unilateral
		HOA	20 (5)	49.7 ± 9.5	173 ± 10		25.5 ± 3.2	
	Wesseling et al. (54)	CON	18 (9)	53.0 ± 5.0	171 ± 10	69.3 ± 12.5	23.7 ± 3.1	Severe; unilateral
		HOA	14 (5)	47.3 ± 11.8	173 ± 9	76.3 ± 16.6	25.2 ± 3.7	

(*Continued*)

TABLE 2 Continued

Study	Report	Population characteristics						
		Group	*n* (♀)	age [years]	height [cm]	weight [kg]	BMI [kg/m^2]	OA characteristics (severity, laterality, cause)
V	Ornetti et al. (55)	CON	9 (7)	60.3 ± 7				Mixed; unilateral; primary
		HOA	11 (8)	60.5 ± 7			25.7 ± 6	
W	Popovic et al. (56)	CON	30 (16)	44.7 ± 13.5			23.7 ± 2.4	Mild/Moderate; -
		HOA	42 (23)	49.6 ± 15.2			24.5 ± 3.3	
X	Porta et al. (57)	CON	11 (5)	67.8 ± 5.4	165.9 ± 7.6	75.9 ± 10.1		Severe; -; primary
		HOA	11 (5)	68.3 ± 5.8	162.0 ± 5.4	75.4 ± 16.9		
Y	Reininga et al. (17)	CON	30 (22)	66 ± 6	170 ± 9	69 ± 12		Severe; -
		HOA	60 (45)	59.7 ± 3.3	170.8 ± 2.7	77.7 ± 3.5		
Z	Schmidt et al. (58)	CON	18 (7)	60.4 ± 8.9	173.1 ± 8.9	71.5 ± 13.7	23.7 ± 3.0	Severe; unilateral
		HOA	18 (7)	64.4 ± 7.4	170.9 ± 7.7	80.7 ± 14.4	27.5 ± 3.4	
	Stief et al. (59)	CON	15 (6)	61.5 ± 8.0	174 ± 9	71.7 ± 14.7	23.5 ± 2.9	Severe; unilateral
		HOA	15 (6)	65.9 ± 8.6	172 ± 8	79.7 ± 9.9	27.0 ± 2.3	
AA	Schmitt et al. (60)	CON	15 (7)	49.2 ± 7.1	166 ± 18	67.4 ± 11.6		Severe; unilateral
		HOA	30 (15)	54.8 ± 6.7	172 ± 10	83.2 ± 20.6		
AB	Steingrebe et al. (61)	CON	21 (10)	63.1 ± 9.2	171.1 ± 8.8	74.4 ± 12.7	25.2 ± 2.7	Mild/Moderate; unilateral
		HOA	21 (10)	64.0 ± 9.6	171.2 ± 6.7	71.3 ± 11.9	24.2 ± 2.9	
AC	Tanaka (62)	CON	56 (56)					-; unilateral, bilateral (subgroups)
		HOA	24 (24)					
AD	Tateuchi et al. (63)	CON	13 (13)	62.6 ± 4.4	152.7 ± 4.9	50.6 ± 5.3		Severe; -
		HOA	14 (14)	59.3 ± 5.3	153.3 ± 5.5	53.3 ± 9.1		
AE	Thurston (64)	CON	10 (-)	63.4 ± 8.1				Severe; unilateral
		HOA	20 (0)	65.1 ± 7.8				
Z & AF	van Drongelen et al. (65)	CON	26 (16)	63.3 ± 7.9	168 ± 10	69.3 ± 12.8	24.6 ± 3.1	Severe; unilateral, bilateral (subgroups)
		HOA	52 (32)	64.3 ± 3.1	169 ± 3.1	77.9 ± 4.0	27.1 ± 2.1	
	van Drongelen et al. (66)	CON	46 (25)	64.2 ± 7.0	169 ± 10	69.0 ± 12.6	24.2 ± 2.8	Severe; unilateral
		HOA	51 (21)	60.6 ± 9.9	173 ± 7	80.3 ± 11.5	26.7 ± 2.9	
AF	van Drongelen et al. (67)	CON	18 (7)	60.4 ± 8.0	173 ± 9	72.0 ± 13.9	23.9 ± 3.2	Severe; unilateral
		HOA	17 (8)	60.5 ± 9.9	172 ± 9	83.3 ± 15.9	28.1 ± 4.9	
	van Drongelen et al. (68)	CON	15 (6)	61.5 ± 8.9	174 ± 9	71.7 ± 14.7	23.5 ± 2.9	Severe; unilateral
		HOA	22 (13)	62.3 ± 10.2	171 ± 10	82.4 ± 16.7	28.2 ± 4.9	
AG	Watanabe et al. (69)	CON	54 (54)	29.4				Mild/Moderate; unilateral
		HOA	30 (30)	31.2				
	Watanabe et al. (70)	CON	54 (54)	29.4 ± 4.6				Mild/Moderate; unilateral
		HOA	30 (30)	31.2 ± 5.6				
AH	Watelain et al. (16)	CON	17 (9)	63.6 ± 5.2	169 ± 6	71.1 ± 14.0		Severe; unilateral
		HOA	17 (9)	58.9 ± 7.1	167 ± 12	76.6 ± 14.7		
AI	Zügner et al. (20)	CON	20 (10)	45.4 ± 4.4			24.3 ± 1.4	Severe; unilateral
		HOA	20 (10)	58.5 ± 5.2			28.0 ± 2.5	

CON, control group; HOA, hip osteoarthritis group.

Data of 29 studies (27 studies on gait, 2 studies on stair walking) were included in the meta-analysis. For the sake of brevity, only forest plots of analyses with significant results are included in the text, and forest plots for all other analyses can be found in the **Supplementary Material**.

3.1. Risk of bias & quality of reporting

The results of the risk of bias assessment are presented in **Table 4**. The mean score was 10.5 (±2.3) with the minimum being 2 and the maximum being 15 (maximum achievable = 17). We did not exclude any of the studies on the basis of their total score.

Most studies have high scores regarding reporting (Q1–3, 5–7, 10) and internal validity (bias, Q16, 18, 20). However, external validity (Q11, 12) is hardly ever to determine as detailed information about the recruitment process is missing. Likewise, questions regarding internal validity (confounding, Q21, 22) are often undetermined as information on CON subject recruitment and time period of subject recruitment is missing. Power analysis was only reported in 6 of the revised 47 reports (Q27).

3.2. Gait

The results from 33 studies analysing gait movement are reviewed below. The results are presented separated by joint,

TABLE 3 Details on osteoarthritis (OA) assessment method and study design of included studies.

		OA assessment		Study design		
Study	Report	Radiographic	Functional	Movement	Measurement system	Analyzed joints
A	Aminian et al. (30)		HHS	Gait OG	IMU	Knee (I & C)
B	Ardestani & Wimmer (31)	KL		Gait OG	Optoelectronic	Hip (I), Knee (I), Ankle (I)
C	Baker et al. (32)	KL	HOOS	Gait TM	Optoelectronic	Hip (I)
	Rutherford et al. (14)		HOOS, WOMAC	Gait TM	Optoelectronic	Knee (I & C)
	Rutherford et al. (33)			Gait TM	Optoelectronic	Hip (I)
D	Bejek et al. (34)	KL	HHS	Gait TM	Ultrasound	Hip (I & C), Knee (I & C), Pelvis
E	Benedetti et al. (35)			Gait OG	Optoelectronic	Hip (I), Pelvis
F	Bolink et al. (36)	KL		Gait OG	IMU	Pelvis
G	Brand & Crowninshield (37) (CON data: Crowninshield et al.) (38)			Gait OG	Biplanar photography	Hip (I)
H	Constantinou et al. (39)	KL	HHS	Gait OG	Optoelectronic	Hip (I), Pelvis
	Diamond et al. (13)	KL	HHS	Gait OG	Optoelectronic	Hip (I)
I	Eitzen et al. (11)			Gait OG	Optoelectronic	Hip (I), Knee (I), Ankle (I)
J	Foucher (40)	KL		Gait OG	Optoelectronic	Hip (I)
K	Foucher et al. (41)		HHS	Gait OG	Optoelectronic	Hip (I)
L	Foucher & Wimmer (42)			Gait OG	Optoelectronic	Hip (C), Knee (C)
	Foucher et al. (43)			Gait OG	Optoelectronic	Hip (I)
M	Hall et al. (44)	KL	HOOS	Stairs (A & D)	Optoelectronic	Hip (I), Pelvis
N	Hara et al. (45) (CON data: Hara et al. (46)	KL		Gait TM	Continuous radiographic imaging	Hip (I), Pelvis
O	Hurwitz et al. (47)		HHS	Gait OG	Optoelectronic	Hip (I & C)
P	Ismailidis et al. (48)	KL	HOOS	Gait OG	IMU	Hip (I), Knee (I), Ankle (I)
	Ismailidis et al. (49)	KL	HOOS	Gait OG	IMU	Hip (I & C), Knee (I & C), Ankle (I & C)
	Nüesch et al. (50)	KL	HOOS	Gait OG	IMU	Hip (I & C), Knee (I & C), Ankle (I & C)
Q	Kataoka et al. (51)	KL	HHS	Gait TM	IMU	Hip (I), Knee (I), Ankle (I)
R	Kubota et al. (52)	KL	JOA HS	Gait OG	Optoelectronic	Hip (I), Ankle (I), Pelvis
S	Kumar et al. (12)	KL	HOOS	Gait OG	Optoelectronic	Hip (I)
T	Leigh et al. (15)	KL	WOMAC	Gait TM	Optoelectronic	Hip (I), Knee (I), Ankle (I), Pelvis
U	Meyer et al. (53)	Tönnis		Gait OG	Optoelectronic	Hip (I)
	Wesseling et al. (54)			Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C)
V	Ornetti et al. (55)	KL		Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C)
W	Popovic et al. (56)	KL	HOOS	Stairs (A & D)	Optoelectronic	Hip (I), Knee (I), Ankle (I)
X	Porta et al. (57)	KL		Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C)
Y	Reininga et al. (17)			Gait OG	IMU	Pelvis
Z	Schmidt et al. (58)	KL	HHS	Gait OG	Optoelectronic	Knee (I & C)
	Stief et al. (59)	KL	HHS	Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Pelvis
AA	Schmitt et al. (60)			Gait OG	Optoelectronic	Hip (I), Knee (I), Ankle (I)
AB	Steingrebe et al. (61)	KL	HHS, HOOS	Gait OG	Optoelectronic	Hip (I), Pelvis
AC	Tanaka (62)			Gait OG	Optoelectronic	Hip (I & C)
AD	Tateuchi et al. (63)		HHS	Gait OG, Turning 45°	Optoelectronic	Hip (I), Knee (I), Ankle (I)
AE	Thurston (64)			Gait OG	Video camera	Pelvis
Z & AF	van Drongelen et al. (65)	KL	HOOS, HHS	Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C), Pelvis
	van Drongelen et al. (66)		HOOS, HHS	Gait OG	Optoelectronic	Hip (I), Knee (I), Pelvis
AF	van Drongelen et al. (67)	KL		Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Pelvis
	van Drongelen et al. (68)			Gait OG	Optoelectronic	Hip (I & C), Knee (I & C)
AG	Watanabe et al. (69)			Gait OG	Optoelectronic	Pelvis
	Watanabe et al. (70)			Gait OG	Optoelectronic	Pelvis
AH	Watelain et al. (16)	KL	Lequesne Index	Gait OG	Optoelectronic	Hip (I), Pelvis
AI	Zügner et al. (20)	Ahlbäck		Gait OG	Optoelectronic	Hip (I)

KL, Kellgren-Lawrence-Score; HHS, Harris Hip Score; HOOS, Hip Osteoarthritis Outcome Score; WOMAC, Western Ontario and McMaster Universities Osteoarthritis Index; JOA HS, Japanese Orthopaedic Association hip score; OG, overground; TM, treadmill; IMU, inertial measurement unit; I, ipsilateral; C, contralateral.

TABLE 4 Results of the risk of bias assessment using the downs & black checklist (28).

Study	Report	Q1	Q2	Q3	Q5	Q6	Q7	Q10	Q11	Q12	Q16	Q18	Q20	Q21	Q22	Q25	Q27	Total	Total [%]
A	Aminian et al. (30)	1	1	0	1	1	1	0	0[a]	0[a]	1	0[a]	1	0[a]	0[a]	0	0	7	41
B	Ardestani & Wimmer (31)	1	1	1	2	1	1	1	0	0[a]	1	0[a]	1	0[a]	1	1	0	12	71
C	Baker et al. (32)	1	1	1	1	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	1	0	11	65
	Rutherford et al. (14)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	1	0	12	71
	Rutherford et al. (33)	1	1	1	1	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	1	0	11	65
D	Bejek et al. (34)	1	1	1	2	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	1	0	11	65
E	Benedetti et al. (35)	1	1	0	1	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	0	0	9	53
F	Bolink et al. (36)	1	1	1	2	1	1	0	1	0[a]	1	1	0	0[a]	0[a]	1	0	11	65
G	Brand & Crowninshield (37) [CON data: Crowninshield et al. (38)]	1	1	0	1	0	0	0	0[a]	0[a]	1	0[a]	1	0[a]	0[a]	0	0	5	29
H	Constantinou et al. (39)	1	1	1	2	1	1	1	0[a]	1	1	1	1	1	1	0	1	15	88
	Diamond et al. (13)	1	1	1	2	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	10	59
I	Eitzen et al. (11)	1	1	1	2	1	1	1	1	1	1	1	1	0	0[a]	1	0	14	82
J	Foucher (40)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	1	0[a]	1	0	13	76
K	Foucher et al. (41)	1	1	0	2	0	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	1	0	10	59
L	Foucher & Wimmer (42)	1	1	0	1	0	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	8	47
	Foucher et al. (43)	1	1	0	2	0	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	1	1	11	65
M	Hall et al. (44)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	1	0[a]	0	0	12	71
N	Hara et al. (45) [CON data: Hara et al. (46)]	1	1	1	1	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	9	53
O	Hurwitz et al. (47)	1	0	0	2	1	1	1	0[a]	0[a]	0	1	1	0[a]	0[a]	1	0	9	53
P	Ismailidis et al. (48)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	0	1	12	71
	Ismailidis et al. (49)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	0	1	12	71
	Nüesch et al. (50)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	0	1	12	71
Q	Kataoka et al. (51)	1	1	1	2	1	0	1	0[a]	0[a]	1	1	1	0[a]	0[a]	1	0	11	65
R	Kubota et al. (52)	1	1	1	2	1	1	0	0[a]	0[a]	1	0	1	0[a]	0[a]	1	0	10	59
S	Kumar et al. (12)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	1	0[a]	1	0	13	76
T	Leigh et al. (15)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	1	1	13	76
U	Meyer et al. (53)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	1	0	12	71
	Wesseling et al. (54)	1	1	0	2	1	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	1	0	11	65
V	Ornetti et al. (55)	1	1	1	1	1	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	10	59
W	Popovic et al. (56)	1	1	1	2	1	1	1	0[a]	0	1	1	1	1	0[a]	1	0	13	76
X	Porta et al. (57)	1	1	1	2	1	1	1	0	0[a]	1	?	1	0	0[a]	1	0	12	71
Y	Reininga et al. (17)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	1	0	12	71
Z	Schmidt et al. (58)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	0	0	11	65
	Stief et al. (59)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	0	0	11	65
AA	Schmitt et al. (60)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	0	0	11	65
AB	Steingrebe et al. (61)	1	1	1	2	1	1	1	0	0[a]	1	1	1	0[a]	0[a]	1	0	12	71
AC	Tanaka (62)	0	1	0	1	0	0	0	0[a]	0[a]	0	0[a]	0	0[a]	0[a]	0	0	2	12
AD	Tateuchi et al. (63)	1	1	0	2	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	9	53
AE	Thurston (64)	1	1	0	1	1	1	0	0[a]	0[a]	1	0	1	0[a]	0[a]	0	0	7	41

(*Continued*)

TABLE 4 Continued

Study	Report	Q1	Q2	Q3	Q5	Q6	Q7	Q10	Q11	Q12	Q16	Q18	Q20	Q21	Q22	Q25	Q27	Total	Total [%]
Z & AF	van Drongelen et al. (65)	1	1	0	2	1	1	0	0[a]	0[a]	1	1	1	0	0[a]	0	0	9	53
	van Drongelen et al. (66)	1	1	0	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	0	0	10	59
AF	van Drongelen et al. (67)	1	1	0	2	1	1	1	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	10	59
	van Drongelen et al. (68)	1	1	0	2	1	1	1	0[a]	0[a]	1	1	1	1	0[a]	0	0	11	65
AG	Watanabe et al. (69)	1	1	0	1	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	8	47
	Watanabe et al. (70)	1	1	0	1	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	8	47
AH	Watelain et al. (16)	1	1	1	2	1	1	0	0[a]	0[a]	1	1	1	0[a]	0[a]	0	0	10	59
AI	Zügner et al. (20)	1	1	1	2	1	1	1	0[a]	0[a]	1	1	1	0	0[a]	1	0	12	71

[a]Unable to determine.

movement plane and laterality. Within each subsection, the results are presented in the following order: results of the meta-analysis, a qualitative summary of data and studies not included in the meta-analysis, results of continuous angle-time curve analyses and, if available, results of subgroup analyses.

3.2.1. Hip joint

3.2.1.1. Sagittal plane kinematics

Ipsilateral. Subjects with HOA have reduced peak extension of the affected hip joint during the gait cycle (GC) (10 studies, SMD = −1.28; 95% CI = −1.61, −0.95; I^2 = 28; **Figure 2A**). Similar results were found for the peak hip extension during the stance phase (SP) (7 studies, SMD = −1.22; 95% CI = −1.72, −0.71; I^2 = 71; **Figure 2B**). Hip sagittal angle at toe-off was significantly reduced (5 studies, SMD = −0.86; 95% CI = −1.41, −0.32; I^2 = 45; **Figure 2C**). Peak flexion was not different between HOA and CON subjects (GC: 8 studies, SMD = −0.53; 95% CI = −1.25, 0.19; I^2 = 80; SP: 5 studies, SMD = −0.36; 95% CI = −1.34, 0.62; I^2 = 86; swing phase: 2 studies, SMD = −1.26; 95% CI = −6.81, 4.28; I^2 = 54).

Hip flexion at initial contact did not differ between subject groups (3 studies, SMD = −0.67; 95% CI = −2.09, 0.76; I^2 = 76).

There was a significant reduction of the sagittal hip range of motion (ROM) across the gait cycle (9 studies, SMD = −2.80; 95% CI = −3.84, −1.75; I^2 = 91, **Figure 2D**), but not across the stance phase (5 studies, SMD = −1.42; 95% CI = −3.15, 0.31; I^2 = 95).

Studies not included in the meta-analysis also showed reduced peak hip extension (60, 62). Results on peak hip flexion varied, with Schmitt et al. (60) reporting increased peak hip flexion in unilateral subjects and Tanaka (62) in bilateral subjects. In contrast, Tanaka (62) showed reduced peak hip flexion in unilateral HOA subjects. Additionally, a reduction of hip flexion at initial contact was found (60). Reduced sagittal hip ROM was found in 3 out of 4 studies [significant reduction in ROM during SP (41, 42) and swing phase (49); no significant difference in ROM during GC (37)].

All 6 of the studies [(31), study *P* (48, 50), study *U* (53, 54) , 57, study Z & AF (65, 66)] analysing continuous sagittal hip angle-time curves show differences between HOA and CON subjects.

Subgroup analysis did not find significant differences in peak hip flexion during the stance phase or gait cycle for subjects with mild/moderate HOA (ST: 2 studies, SMD = 0.31; 95% CI = −4.33, 4.95; I^2 = 86; GC: 2 studies, SMD = −0.36; 95% CI = −1.53, 0.82 I^2 = 0) or subjects with severe HOA (ST: 3 studies, SMD = −0.80; 95% CI = −2.28, 0.67; I^2 = 80; GC: 4 studies, SMD = −0.61; 95% CI = −2.61, 1.39; I^2 = 85).

Hip peak extension during the stance phase or gait cycle was significantly different for subjects with severe HOA (ST: 4 studies, SMD = −1.57; 95% CI = −2.14, −1.00; I^2 = 22; GC: 6 studies, SMD = −1.31; 95% CI = −1.78, −0.85; I^2 = 19, **Figure 3**) but not for subjects with mild/moderate HOA (ST: 3 studies, SMD = −0.80; 95% CI = −1.89, 0.29; I^2 = 55; GC: 2 studies, SMD = −0.90; 95% CI = −5.09, 3.30; I^2 = 39).

Hip sagittal angle at toe-off was significantly decreased in severe HOA subjects (2 studies, SMD = −1.31; 95% CI = −1.63,

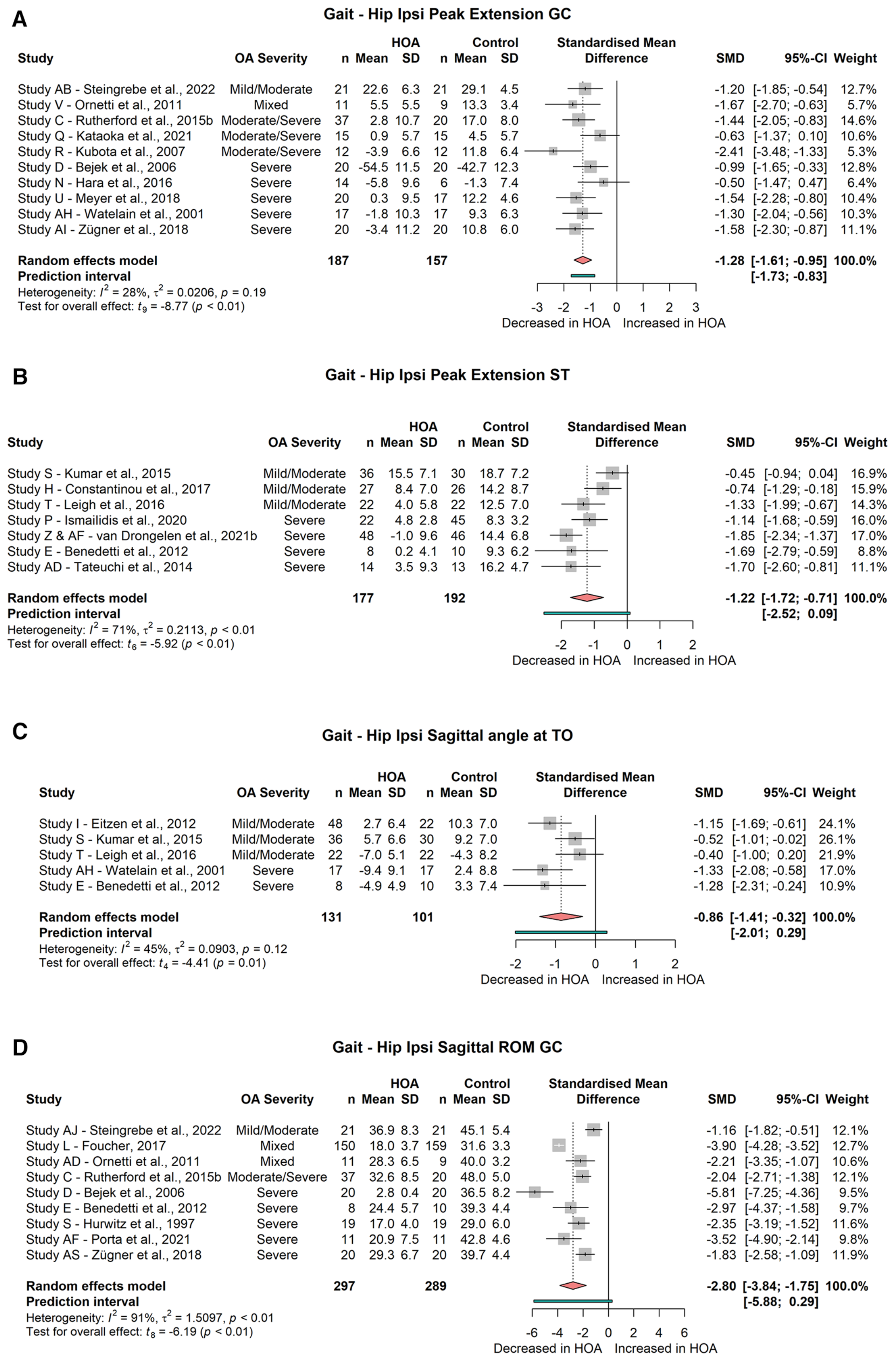

FIGURE 2
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) ipsilateral peak hip extension during gait cycle (GC), (B) ipsilateral peak hip extension during stance phase (ST), (C) ipsilateral hip sagittal angle at toe-off (TO), (D) ipsilateral hip sagittal range of motion (ROM) across gait cycle; during gait.

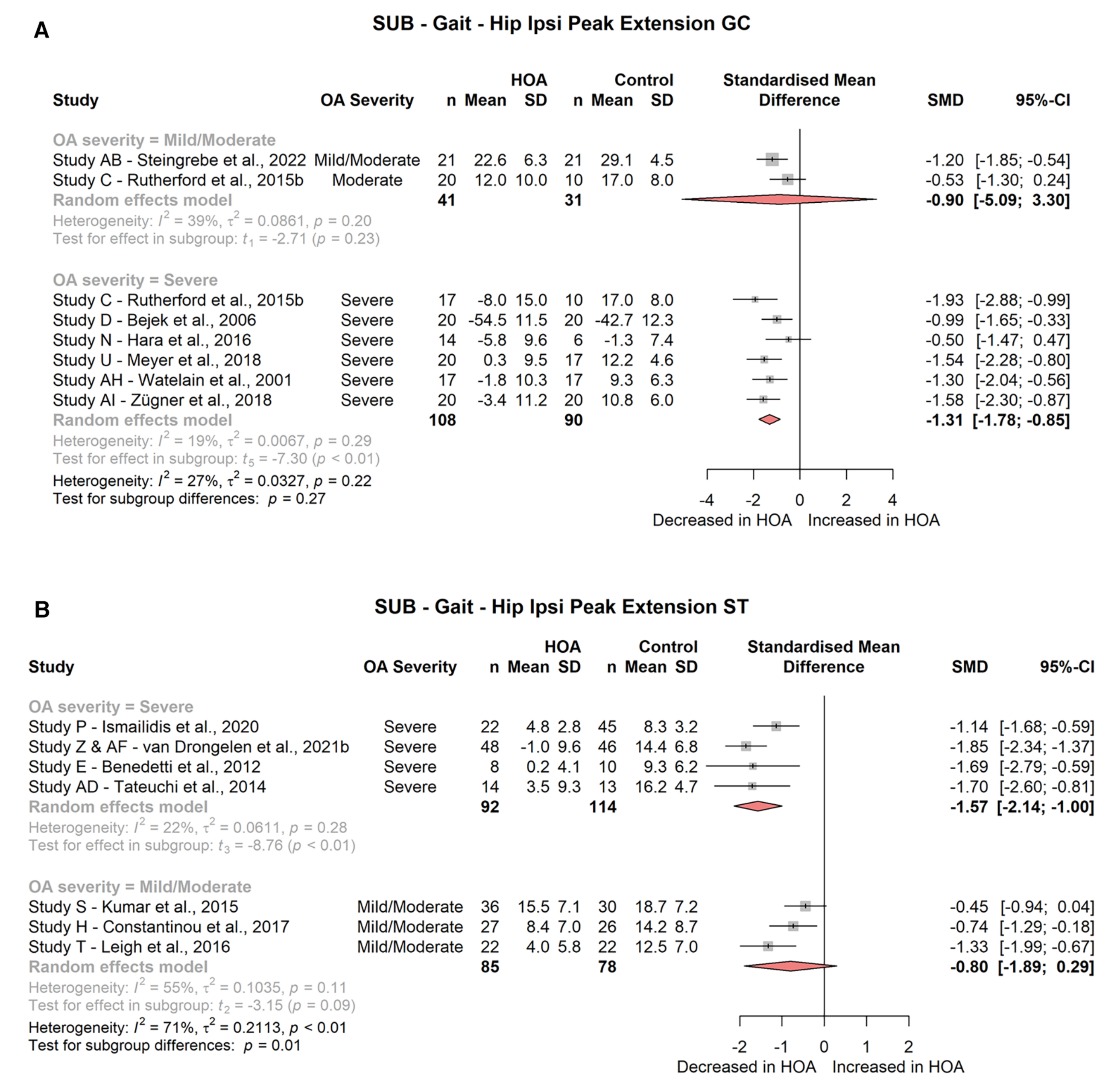

FIGURE 3
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on: (A) ipsilateral peak hip extension during gait cycle (GC), (B) ipsilateral peak hip extension during stance phase (ST); during gait.

−0.99; $I^2 = 0$, Figure 4B), but not in mild/moderate HOA subjects (3 studies, SMD = −0.69; 95% CI = −1.68, 0.30; $I^2 = 52$).

Hip sagittal ROM across the gait cycle was significantly decreased for mild/moderate (2 studies, SMD = −1.19; 95% CI = −1.71, −0.68; $I^2 = 0$) and severe HOA (6 studies, SMD = −3.23; 95% CI = −4.65, −1.80; $I^2 = 81$) subjects (Figure 4A). Hip sagittal ROM across the stance phase was not different for any of the HOA severity subgroups (mild/moderate: 3 studies, SMD = −0.70; 95% CI = −2.85, 1.45; $I^2 = 91$; severe: 2 studies, SMD = −2.51; 95% CI = −16.52, 11.51; $I^2 = 96$).

Contralateral. Subjects with unilateral HOA also displayed a reduced peak extension of the contralateral hip joint during the stance phase of gait (3 studies, SMD = −0.59; 95% CI = −0.97, −0.22; $I^2 = 0$, Figure 5) but not during the gait cycle (2 studies, SMD = −0.15; 95% CI = −7.76, 7.45; $I^2 = 78$). No differences between HOA and CON subjects were found for contralateral peak hip flexion during the stance phase (3 studies, SMD = 0.23; 95% CI = −0.41, 0.87; $I^2 = 0$) or gait cycle (2 studies, SMD = −0.7; 95% CI = −2.38, 0.98; $I^2 = 0$), nor for contralateral sagittal hip ROM across the stance phase (3 studies, SMD = −0.3; 95% CI = −0.74, 0.15; $I^2 = 0$) or gait cycle (4 studies, SMD = −0.79; 95% CI = −2.65, 1.06; $I^2 = 88$) during the meta-analysis.

A study not included in the meta-analysis also reported reduced contralateral peak hip extension and increased peak hip flexion angles (62).

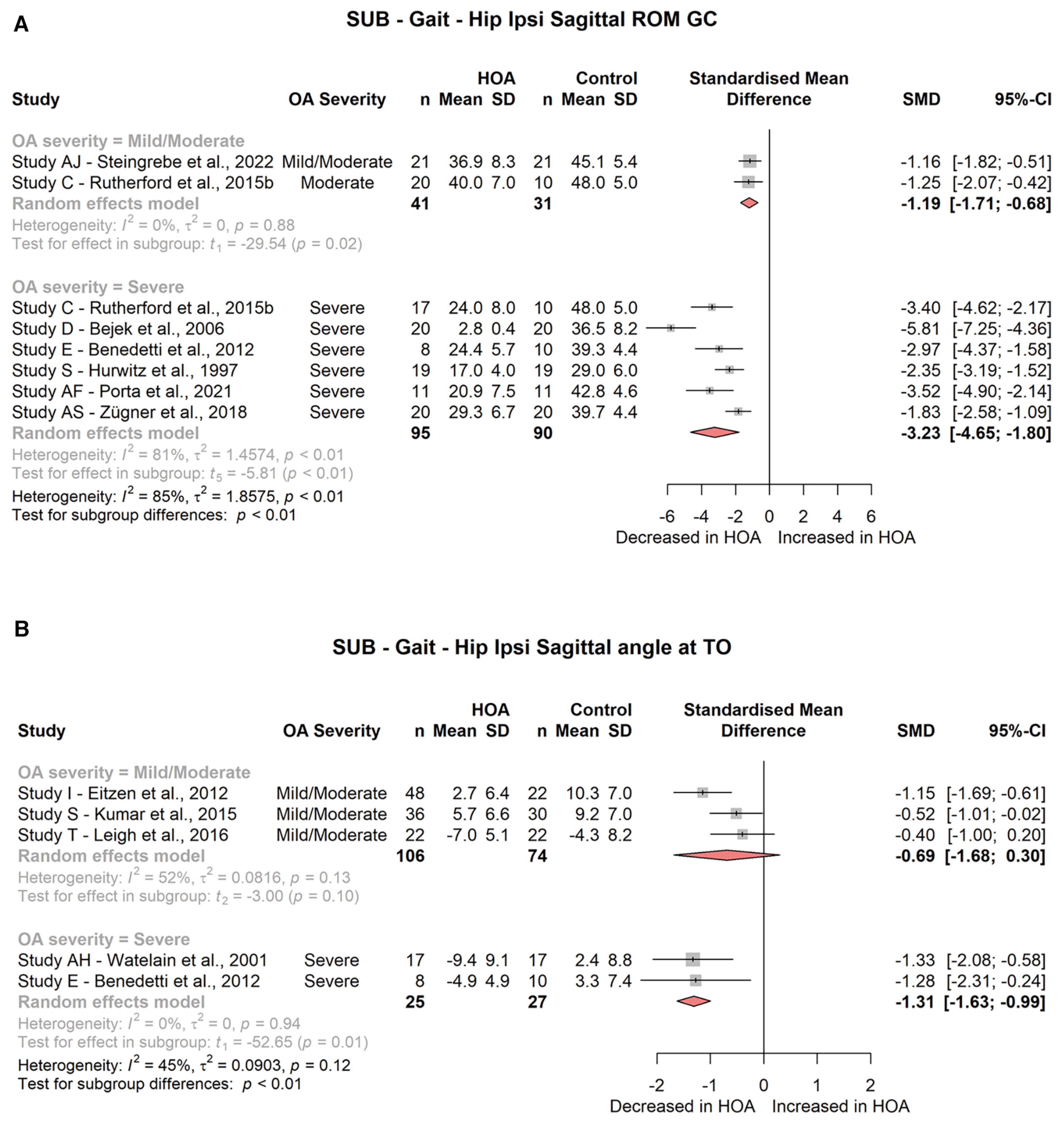

FIGURE 4
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on: (A) ipsilateral hip sagittal range of motion (ROM) across gait cycle (GC), (B) ipsilateral hip sagittal angle at toe-off (TO); during gait.

Of the 4 studies (50, 54, 57, 65) analysing contralateral sagittal hip angle-time curves, 3 found differences in contralateral hip sagittal angles.

3.2.1.2. Frontal plane kinematics

Ipsilateral. The meta-analysis showed no significant differences for peak hip abduction during the gait cycle (4 studies, SMD = −0.7; 95% CI = −1.83, 0.44; $I^2 = 75$) or at toe-off (2 studies, SMD = −0.16; 95% CI = −5.24, 4.92; $I^2 = 66$). Peak hip adduction did not differ significantly between groups either during the gait cycle or during stance (GC: 5 studies, SMD = −0.35; 95% CI = −0.87, 0.18; $I^2 = 42$; ST: 4 studies, SMD = −0.43; 95% CI = −1.24, 0.38; $I^2 = 46$), but a significant reduction of the frontal plane hip ROM across the gait cycle (4 studies, SMD = −0.86; 95% CI = −1.93, −0.33; $I^2 = 0$, **Figure 6A**) was found. Frontal plane hip ROM across the stance phase was not different between groups (2 studies, SMD = −1.38; 95% CI = −11.65, 8.88; $I^2 = 88$).

Studies not included in the meta-analysis showed an increased hip abduction at midstance (15), as well as

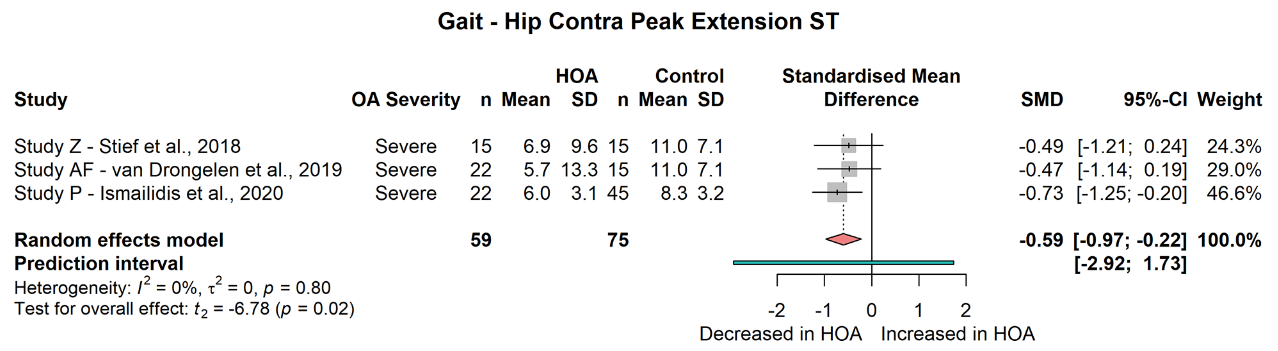

FIGURE 5
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for contralateral peak hip extension during stance phase (ST); during gait.

decreased peak hip adduction during early and late stance (13). Peak abduction during stance (63) and swing (35), and hip frontal angle at peak hip extension (15), did not differ between subject groups.

Both studies using SPM analysis found differences in hip frontal plane angle [study *U* (53, 54), 65]. Van Drongelen et al. (65) found differences for subjects with unilateral HOA but not for those with bilateral HOA.

Results from the subgroup analysis did not find group differences in peak hip abduction or adduction during the gait cycle for mild/moderate HOA subjects (abduction: 2 studies, SMD = −0.01; 95% CI = −2.05, 2.03; $I^2 = 0$; adduction: 2 studies, SMD = −0.44; 95% CI = −2.80, 1.91; $I^2 = 0$) or for severe HOA subjects (abduction: 2 studies, SMD = −0.94; 95% CI = −7.33, 5.46; $I^2 = 70$; adduction: 2 studies, SMD = −0.02; 95% CI = −7.02, 6.98; $I^2 = 78$).

Hip frontal ROM across the gait cycle was not different in mild/moderate (2 studies, SMD = −0.39; 95% CI = −0.93, 0.16; $I^2 = 0$) or in severe HOA subjects (2 studies, SMD = −1.35; 95% CI = −2.74, 0.04; $I^2 = 37$).

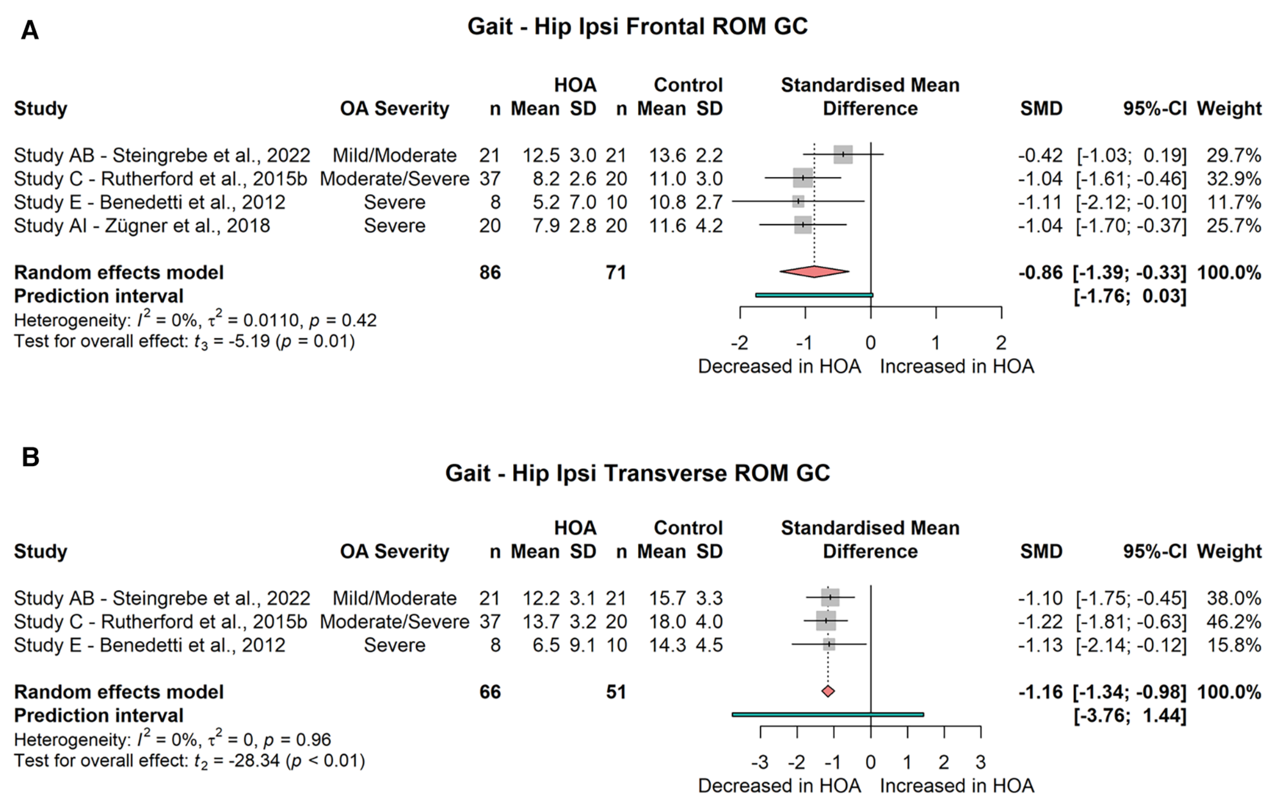

FIGURE 6
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral hip frontal range of motion (ROM) across gait cycle (GC), (B) ipsilateral hip transverse ROM across GC; during gait.

Contralateral. Van Drongelen et al. (67, 68) did not find a significant difference between CON and HOA subjects for the peak adduction angle during stance.

Results from two SPM analyses (54, 65) both show differences in contralateral hip frontal angle-time curves.

3.2.1.3. Transverse plane kinematics

Ipsilateral. The meta-analysis yielded no difference for the peak external rotation angle (2 studies, SMD = 0.17; 95% CI = −0.59, 0.94; $I^2 = 0$) or the peak internal rotation angle during the gait cycle (3 studies, SMD = −0.5; 95% CI = −1.75, 0.76; $I^2 = 59$). Also, the transverse angle at toe-off was not different (2 studies, SMD = −0.11; 95% CI = −14.61, 14.39; $I^2 = 95$). However, a significantly decreased transverse ROM across the gait cycle (3 studies, SMD = −1.16; 95% CI = −1.34, −0.98; $I^2 = 0$, **Figure 6B**) was found.

Data not included in the meta-analysis showed a significant reduction of the peak internal rotation angle during the stance phase and a significant increase in peak external rotation during the swing phase (35). Similarly, Leigh and colleagues (15) found the hip joint of HOA subjects was significantly more externally rotated at terminal hip extension but not at midstance.

One study (54) found differences in the hip transverse angle-time curve in an SPM analysis.

The subgroup analysis did not show significant differences in hip transverse ROM across the gait cycle for mild/moderate (2 studies, SMD = −0.84; 95% CI = −4.60, 2.93; $I^2 = 26$) or severe HOA subjects (2 studies, SMD = −1.47; 95% CI = −5.39, 2.45; $I^2 = 0$).

Contralateral. Only 1 study (54) reported data on transverse plane contralateral hip angles and did not find any significant differences during an SPM analysis.

3.2.2. Knee joint

3.2.2.1. Sagittal plane kinematics

Ipsilateral. The meta-analysis did not show a significant difference for peak knee extension during the gait cycle (3 studies, SMD = 0.72; 95% CI = −2.21, 3.65; $I^2 = 87$) or stance phase (2 studies, SMD = 0.81; 95% CI = −13.61, 15.23; $I^2 = 96$). Peak flexion was significantly reduced during the gait cycle (3 studies, SMD = −0.87; 95% CI = −1.19, −0.56; $I^2 = 0$, **Figure 7A**), but not during stance (3 studies, SMD = −0.68; 95% CI = −1.55, 0.2; $I^2 = 44$). Sagittal knee angle at initial contact (2 studies, SMD = −0.09; 95% CI = −0.54, 0.37; $I^2 = 0$), midstance (2 studies, SMD = 0.22; 95% CI = −4.26, 4.69; $I^2 = 68$), toe-off (2 studies, SMD = 0.22; 95% CI = −1.60, 2.03; $I^2 = 0$) and peak hip extension (2 studies, SMD = 1.30; 95% CI = −0.52, 3.11; $I^2 = 0$) was not different between groups. Sagittal knee ROM was significantly reduced across the gait cycle (4 studies, SMD = −1.64; 95% CI = −2.24, −0.86; $I^2 = 5$, **Figure 7B**) but not across the stance phase (3 studies, SMD = −1.06; 95% CI = −5.25, 3.13; $I^2 = 97$).

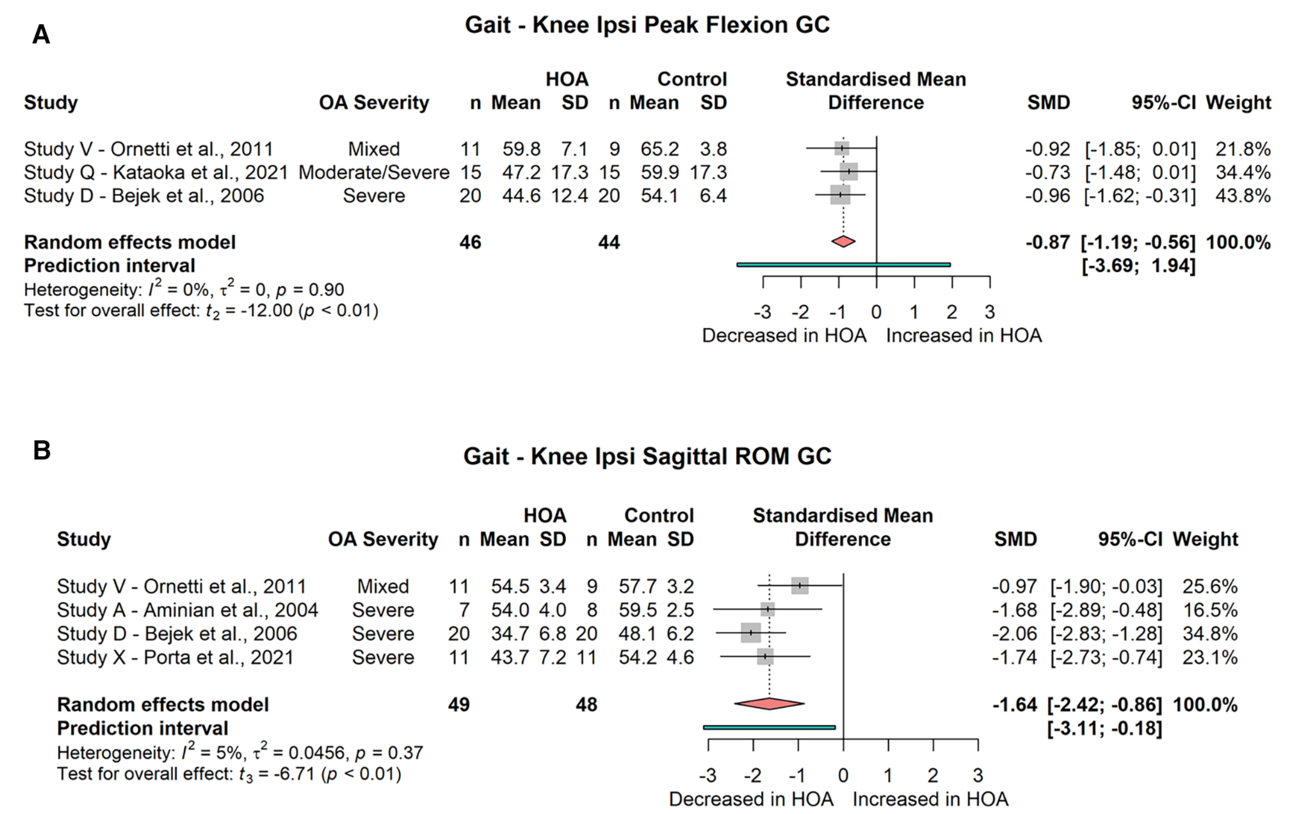

FIGURE 7
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) ipsilateral knee peak flexion during gait cycle (GC), (B) ipsilateral knee sagittal range of motion (ROM) across GC; during gait.

A study not included in the meta-analysis by Ismailidis et al. (49), described a significantly decreased knee ROM across the swing phase, while peak flexion during swing did not differ between groups.

Analyses of knee sagittal angle-time curves showed differences in all 5 studies [31, study *P* (48, 50) 54, 57, study Z & AF (65, 66)].

Contralateral. For the contralateral knee joint, the meta-analysis did not show a significant difference for peak knee extension during the gait cycle (2 studies, SMD = −0.03; 95% CI = −1.77, 1.71; $I^2 = 0$) but a significant difference during the stance phase (2 studies, SMD = 0.52; 95% CI = 0.18, 0.87; $I^2 = 0$, **Figure 8A**). Peak flexion was not significantly reduced during the gait cycle (2 studies, SMD = −0.87; 95% CI = −3.21, 1.47; $I^2 = 0$) or stance phase (3 studies, SMD = −0.05; 95% CI = −0.66, 0.55; $I^2 = 0$). Sagittal knee ROM was significantly reduced across both the gait cycle (5 studies, SMD = −0.73; 95% CI = −1.08, −0.39; $I^2 = 0$, **Figure 8B**) and the stance phase (3 studies, SMD = −0.65; 95% CI = −0.71, −0.59; $I^2 = 0$, **Figure 8C**).

Out of 4 studies analysing sagittal contralateral knee angles time-curves, 2 found differences between subject groups (54, 57) and 2 did not (50, 65).

3.2.2.2. Frontal plane kinematics

Ipsilateral. The meta-analysis did not show a difference between groups for the ROM across the stance phase (2 studies, SMD = −0.09; 95% CI = −5.55, 5.37; $I^2 = 68$).

One study not included in the meta-analysis reported no significant group differences for the frontal knee angle at midstance, toe-off or peak hip extension (15).

Contralateral. The meta-analysis did not show a difference between groups for the ROM across the stance phase (2 studies, SMD = −0.08; 95% CI = −5.67, 5.51; $I^2 = 69$).

3.2.2.3. Transverse plane kinematics

Ipsilateral. Rutherford et al. (14) did not find a significant difference in transverse knee ROM across the stance phase. However, Leigh et al. (15) found significantly increased external knee rotation angles at midstance and at peak hip extension, but not at toe-off.

Contralateral. No studies were retrieved that presented data on contralateral transverse plane knee kinematics.

3.2.3. Ankle joint

3.2.3.1. Sagittal plane kinematics

Ipsilateral. The meta-analysis did not show differences for the peak dorsiflexion angle during the gait cycle (2 studies, SMD = −0.01; 95% CI = −9.38, 9.37; $I^2 = 82$) or stance phase (2 studies, SMD = −0.62; 95% CI = −3.77, 5.00; $I^2 = 54$). Peak plantar flexion during the gait cycle (3 studies, SMD = 0.46; 95% CI = −0.98, 1.90; $I^2 = 51$) and during the stance phase (2 studies, SMD = −0.09; 95% CI = −3.67, 3.49; $I^2 = 35$) was not different between groups. Ankle sagittal ROM across the gait cycle (2 studies, SMD = −0.53; 95% CI = −6.47, 5.41; $I^2 = 53$) and across the stance phase (2 studies, SMD = 0.34; 95% CI = −1.61, 2.29; $I^2 = 0$) did not differ between groups. The ankle angles at initial contact (2 studies, SMD = 0.07; 95% CI = −2.45, 2.58; $I^2 = 25$), midstance (2 studies, SMD = 0.18; 95% CI = −1.24, 1.60; $I^2 = 0$), toe-off (2 studies, SMD = 0.02; 95% CI = −1.42, 1.47; $I^2 = 0$) and peak hip extension (2 studies, SMD = 0.25; 95% CI = −1.72, 2.21; $I^2 = 0$) were not different between groups.

In 5 studies ankle sagittal angle-time curves were analysed. Although 2 studies found differences between groups (54, 57), 2 did not (31, 65) and 1 study yielded contradicting results in 2 reports [study *P* (48, 50)].

Contralateral. Ankle angle ROM across the gait cycle was not different between groups (2 studies, SMD = −0.47; 95% CI = −3.88, 2.94; $I^2 = 0$).

Data not included in the meta-analysis showed differences in peak dorsiflexion angle during the gait cycle ($p = 0.05$) but not in peak plantar flexion angle ($p = 0.087$) (55).

In 4 studies contralateral sagittal ankle angle-time curves were analysed. Three of these studies found differences between groups (50, 54, 57) but 1 study did not (65).

3.2.3.2. Frontal plane kinematics

Ipsilateral. Only 1 study analysed frontal plane ankle angles, and found a reduced ankle inversion at toe-off. Frontal ankle angle at midstance and at peak hip extension did not differ between groups (15).

Contralateral. No studies were retrieved that presented data on contralateral frontal plane ankle kinematics.

3.2.3.3. Transverse plane kinematics

Ipsilateral. Only 1 study analysed transverse plane ankle kinematics and did not find differences in ankle angles at midstance, toe-off or peak hip extension (15).

Contralateral. No studies were retrieved that presented data on contralateral transverse plane ankle kinematics.

3.2.4. Pelvis

3.2.4.1. Sagittal plane kinematics

The meta-analysis found no differences in peak anterior tilt during either the gait cycle (4 studies, SMD = 0.70; 95% CI = −0.11, 1.51; $I^2 = 27$) or the stance phase (2 studies, SMD = 0.68; 95% CI = −5.07, 6.43; $I^2 = 85$). Likewise, pelvis peak posterior tilt did not differ between groups during the gait cycle (3 studies, SMD = 0.72; 95% CI = −1.78, 3.23; $I^2 = 79$). Pelvis angle at toe-off also did not differ between CON and HOA subjects (2 studies, SMD = −0.38; 95% CI = −12.24, 11.49; $I^2 = 93$).

Pelvis ROM across the gait cycle was not different between groups (3 studies, SMD = 1.85; 95% CI = −1.99, 5.69; $I^2 = 91$).

Studies not included in the meta-analysis found a significantly increased anterior pelvis tilt at peak hip extension (15), but not at midstance (15, 69) initial contact or toe-off (69) in HOA subjects.

One study analysed pelvis sagittal angle using SPM analysis and found differences between bilateral HOA and CON subjects but not between unilateral HOA and CON subjects (65).

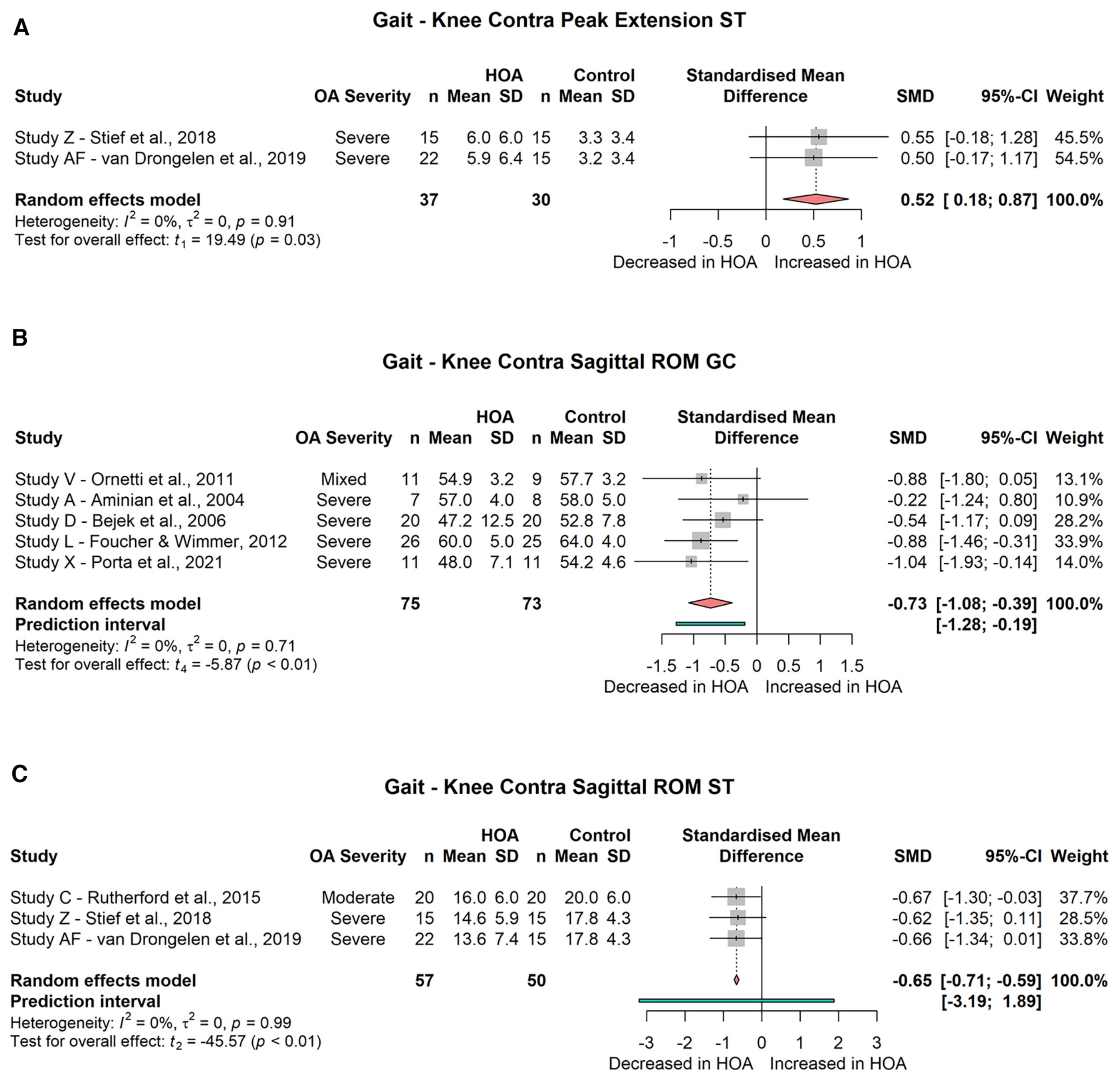

FIGURE 8
Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) contralateral knee peak extension during stance phase (ST), (B) contralateral knee sagittal range of motion (ROM) across gait cycle (GC), (C) contralateral knee sagittal ROM across ST; during gait.

3.2.4.2. Frontal plane kinematics

The meta-analysis found no differences in peak inferior obliquity during the gait cycle (2 studies, SMD = 2.25; 95% CI = −12.44, 16.93; I^2 = 92). Peak superior obliquity was not different in the gait cycle (2 studies, SMD = −0.28; 95% CI = −7.83, 7.27; I^2 = 80) or in the stance phase (3 studies, SMD = −0.43; 95% CI = −2.53, 1.61; I^2 = 78). Likewise, pelvic frontal angle at toe-off did not differ between groups (2 studies, SMD = −0.53; 95% CI = −8.60, 7.54; I^2 = 86).

Pelvis frontal plane ROM was not different between groups either across the gait cycle (5 studies, SMD = −0.19; 95% CI = −3.33, 2.96; I^2 = 96) or in the stance phase (2 studies, SMD = −0.31; 95% CI = −6.80, 6.18; I^2 = 81).

Studies not included in the meta-analysis found no differences in peak inferior obliquity during the swing phase (35); or for the pelvis frontal angle at peak hip extension (15), initial contact, midstance or toe-off [study AG (69, 70)].

Differences between groups were found for pelvis frontal angle at midstance (15) and peak inferior obliquity during single-limb stance (67).

One study analysed pelvis frontal angle using SPM analysis and found differences between bilateral HOA and CON subjects but not between unilateral HOA and CON subjects (65).

3.2.4.3. Transverse plane kinematics

The meta-analysis did not show differences in the pelvis transverse ROM across the gait cycle (3 studies, SMD = −0.06; 95% CI = −0.44, 0.32; $I^2 = 0$) or the pelvis transverse angle at toe-off (2 studies, SMD = 0.35; 95% CI = −3.13, 3.82; $I^2 = 29$).

Studies not included in the meta-analysis found no differences between groups for the transverse pelvic ROM across the stance phase (16), pelvis angle at peak hip extension (15) or peak posterior rotation (35). However, one study reported significant group differences for the pelvis transverse angle at midstance (15).

3.3. Stair walking

Two studies analysing stair walking are reviewed below, separated into stair ascent and stair descent.

3.3.1. Stair ascent

Peak ipsilateral hip flexion, adduction and internal rotation during stance phase did not differ between groups (flexion: 2 studies, SMD = −0.49; 95% CI = −9.71, 8.73; $I^2 = 90$; adduction: 2 studies, SMD = −0.17; 95% CI = −8.42, 8.08; $I^2 = 88$; internal rotation: 2 studies, SMD = 0.41; 95% CI = −3.03, 3.85; $I^2 = 38$).

Results not included in the meta-analysis showed no differences in peak hip extension during the gait cycle (44) or during the stance phase (56). Peak flexion was significantly reduced in the swing phase (44). Hip sagittal ROM across the gait cycle was significantly reduced (44).

Hip peak abduction was significantly reduced during swing (44) and stance phase (56). Hip frontal ROM was significantly reduced (44).

Peak external rotation was significantly reduced during stance phase (56) but not during swing phase (44). Transverse hip ROM was significantly reduced (44).

Peak ipsilateral knee angles in the sagittal and frontal planes during the stance phase did not differ between groups (56). Likewise, peak knee internal rotation was not different. However, peak knee external rotation was significantly increased in HOA subjects.

Peak ipsilateral sagittal and frontal ankle angles did not differ between groups (56). However, peak ankle internal rotation was significantly reduced while peak external rotation was significantly increased in HOA subjects (56).

Peak contralateral pelvis inferior and superior obliquity did not differ between groups (44).

3.3.2. Stair descent

Peak ipsilateral hip flexion during stance did not differ between groups (2 studies, SMD = −0.06; 95% CI = −1.36, 1.25; $I^2 = 0$). Likewise, peak hip adduction and abduction did not differ between CON and HOA subjects (adduction: 2 studies, SMD = 0.05; 95% CI = −5.10, 5.21; $I^2 = 71$; abduction: 2 studies, SMD = −0.03; 95% CI = −5.47, 5.42; $I^2 = 74$). Peak internal and external hip rotation were not different between groups (internal rotation: 2 studies, SMD = 0.15; 95% CI = −4.21, 4.52; $I^2 = 60$; external rotation: 2 studies, SMD = −0.37; 95% CI = −1.42, 0.69; $I^2 = 0$).

Results not included in the meta-analysis showed no differences in peak hip extension during either the stance phase (56) or gait cycle (44). Likewise, peak flexion during swing was not different between groups (44). However, sagittal hip ROM was significantly reduced in HOA subjects (44).

There was no difference between groups in peak hip abduction during swing, hip frontal plane ROM across the gait cycle (44) or transverse plane hip ROM (44).

While peak knee flexion did not differ between groups, peak knee extension was significantly increased in HOA subjects (56).

Frontal plane peak knee angles did not differ between groups (56).

Transverse plane knee angles showed significantly reduced peak internal and significantly increased peak external rotation (56).

At the ankle joint, HOA subjects showed significantly increased peak plantar flexion. Peak dorsi flexion did not differ between groups (56), and neither did peak ankle eversion and inversion (56).

Ankle peak internal rotation was significantly reduced while peak external rotation was significantly increased in HOA subjects (56).

Peak contralateral pelvis inferior and superior obliquity did not differ between groups (44).

3.4. Turning while walking

In a study by Tateuchi and colleagues (63), 45° turns conducted either in a step turn or in a spin turn manner were analysed in subjects with severe HOA. Peak angles during the stance phase were described.

During the step turn, significantly decreased peak hip flexion and extension were found. Peak hip abduction was also significantly reduced. No differences were found for peak hip adduction.

Sagittal plane peak angles of the knee and ankle joint did not differ between groups.

During the spin turn, significantly decreased peak hip flexion and extension were found. Peak hip abduction did not differ between groups. However, significantly reduced peak hip adduction was found in HOA subjects.

Peak knee extension did not differ between groups but peak knee flexion was significantly reduced in HOA subjects.

Ankle sagittal peak angles did not differ between groups.

4. Discussion

The aim of this review and meta-analysis was to summarise existing literature on lower-limb joint kinematics during locomotion in subjects with HOA compared to healthy controls. Where possible, a meta-analysis was performed with the focus on HOA severity and uni- or bilateral involvement.

Overall, 47 reports from 35 individual studies were reviewed. The total score regarding risk of bias and quality of reporting of the included studies varied strongly, with older reports tending to show lower scores.

The first outcome of this systematic literature review is that studies on locomotion tasks other than gait are rare, with only 2 studies on stair walking (44, 56) and 1 study on curve walking (63).

Secondly, a large portion of the analysed subjects are classified as having severe HOA as well as unilateral HOA. This observation might originate in the fact that HOA subjects are often recruited prior to total hip arthroplasty and studies aim at evaluating rehabilitation after surgery. Due to the small number of studies with subjects with mild or moderate HOA, subgroup analyses for HOA severity were only possible for some parameters of ipsilateral hip kinematics. Likewise, it was not possible to perform subgroup analysis regarding HOA laterality, as for none of the revised parameters, data of at least 2 studies for each subgroup, namely unilateral and bilateral HOA subjects, were available. This was mainly caused by a lack of bilateral HOA subject groups or the unavailability of laterality information. As no conjoint analysis was possible, insights on the impact of laterality still have to be based on individual study results, such as those of Tanaka (62) or van Drongelen et al. (65).

Generally, it has to be noted that although 33 individual studies were included on gait, a conjoint analysis is hindered by the multitude of calculated parameters. For example, calculating peak angles or ROM across the stance phase is not comparable to the same parameters calculated across the entire gait cycle. Therefore, of the 68 combined analyses calculated for parameters on gait kinematics, only 10 include 5 or more individual studies. Thereof, 7 refer to ipsilateral hip sagittal angles, 1 refers to ipsilateral hip frontal angle, contralateral knee sagittal angle and pelvis frontal angle, respectively.

In 5 studies continuous angle-time curves were analysed by means of an SPM analysis [31, study *P* (48, 50), study *U* (53, 54), 65] or by point-by-point analysis (57). While these approaches can be advantageous, especially during explorative data analysis, the aggregation of results across multiple studies is difficult.

Five studies assessed kinematics using IMU sensors [17, Study *P* (48–50), 30, 36, 51]. Most of these studies focused on sagittal hip, knee and ankle angles or frontal pelvis angles and demonstrated significant group differences. Only 1 study used IMUs on frontal and transverse hip angles (51), but did not find significant differences.

4.1. Effects of hip osteoarthritis on gait kinematics

For the gait movement, the main results for the hip joint were an overall reduced peak extension in HOA subjects. However, during the subgroup analysis, only subjects with severe HOA demonstrated significant group differences. Similar results were found for the hip sagittal angle at toe-off, which makes sense as peak hip extension and toe-off occur very close to each other.

Reduction of the sagittal hip ROM across the gait cycle occurred with a very large effect size, and was present in subjects with both mild/moderate as well as severe HOA. In contrast, sagittal hip ROM across the stance phase was not different between groups. Thus, it might be crucial to capture stance and swing phase kinematics to discover deviations in gait caused by HOA.

Interestingly, peak hip extension was also reduced in the contralateral limb in subjects with severe unilateral HOA. As hip extension is closely connected to step length (71), this contradicts the consideration of increased step length in the contralateral limb to compensate for decreased ipsilateral step length (18). However, this may be because in most studies healthy CON subjects are not evaluated radiographically so structural changes in the contralateral limb cannot be excluded.

Frontal as well as transverse plane hip ROMs were reduced across the gait cycle. As neither of the peak angles demonstrated significant group differences, the ROM might be more sensitive to group differences as it captures changes occurring at both extrema of the dynamic movement. Little is known about frontal and transverse hip kinematics of the contralateral limb; however, in two studies analyses of contralateral frontal hip angle-time curves found differences between HOA and CON subjects, which requires further investigation.

Ipsilateral as well as contralateral sagittal plane knee kinematics are influenced by HOA. The ipsilateral limb shows decreased peak flexion and a reduced sagittal ROM across the gait cycle. The contralateral limb demonstrates increased peak extension, but with negligible effect size. However, sagittal knee ROM across the stance phase and gait cycle was significantly decreased with small to moderate effect size.

Very few studies analysed the effect of HOA on knee frontal and transverse kinematics. Although no differences were found for frontal plane knee angles, individual study results on knee transverse kinematics varied. Some studies found increased foot-progression angles during walking for subjects with HOA (58, 63), which might be linked to changes in knee rotation.

The meta-analysis did not yield any significant differences in ipsi- or contralateral sagittal ankle kinematics. However, individual study results as well as analysis of angle-time curves partly yielded significant group differences. Thus, ankle kinematics might only differ in specific settings or groups: this should be considered in future studies.

For the frontal and transverse planes, no data exist regarding contralateral ankle kinematics. Ipsilateral ankle kinematics in those planes were only analysed in 1 study which only found a reduced ankle inversion at toe-off.

The meta-analysis did not yield any significant differences in pelvis sagittal, frontal or transverse movement. However, it has to be noted that large study heterogeneity was observed for all analysed parameters, especially for the sagittal and frontal plane. For example, in the analysis of the frontal pelvis ROM across the gait cycle, the results from Bejek et al. (34) differed dramatically from those of the other studies and, if excluded, the random-effects model approach statistical significance ($p = 0.06$). One possible explanation for this deviation might come from different measurement techniques, as Bejek and colleagues were the only ones to use an ultrasound-based motion capture system. Peak anterior pelvis tilt approached significance ($p = 0.07$) in the meta-analysis with a moderate effect size. Anterior tilting of the pelvis might allow the patients to increase step length despite limitations in hip extension (15, 64). Our subgroup analysis did not show a significant difference in peak hip extension angle for subjects with mild or moderate HOA. Thus, compensatory pelvic motion might not or only to a limited extent be necessary in this subject group. Yet, this has to be evaluated in future studies.

The results from our analyses show that modifications of kinematic patterns are not limited to the ipsilateral side nor the affected joint but rather are a complex interplay of changes occurring at the pelvis and both lower limbs. These results are in line with those from whole-body analyses that show the highest discriminatory capacity in hip, knee and ankle sagittal angles and partly frontal plane ankle angles (31, 72, 73).

4.2. Effects of hip osteoarthritis on stair walking kinematics

For stair ascent, no significant differences between HOA and CON subjects were found during the meta-analysis. However, individual study results found decreased hip peak flexion during swing, causing a decrease in sagittal hip ROM. Stair ascent requires a greater sagittal hip ROM than level walking, and a high peak flexion is crucial for ensuring step clearance and avoidance of tripping (74). Thus, decreased peak flexion and sagittal ROM might make HOA subjects prone to falling during stair ascent. Likewise, peak abduction and external rotation of the hip were reduced, causing reduced hip ROM in the frontal and transverse planes. In contrast, the knee and ankle joints seem to be more externally rotated. Meyer et al. (75) reported reduced peak adduction during stair ascent as a strategy for a wider base of support, which was not present in our meta-analysis. Thus, adopting a toe-out gait by externally rotating the foot and tibia might be another strategy to broaden the area of support for the stance limb (76).

For stair descent, the meta-analysis did not yield any significant differences in hip peak angles. However, an individual study still reported reduced sagittal hip ROM (44).

Additionally, an increase in peak knee extension as well as peak plantar flexion was observed in 1 study (56). This might stem from an attempt to prolong contact with the stair at toe-off or to contact the stair sooner to reduce single-support time. Yet, this is speculative and warrants further investigation.

As in stair ascent, a more pronounced external rotation of the knee and ankle joints was observed.

Overall, it has to be considered that only two studies analysed stair walking in HOA subjects, thus insights are still very limited.

Additionally, the staircases used during the studies contained two or four steps, with the force plate included in the first step. Hence, the analysed step always contained movement initiation and/or transition to level walking. Results from Alcock et al. (77) show large differences in hip, knee and ankle sagittal angles during steady-state stair ascent and the transition from gait to stair ascent. Hence, both of the retrieved studies give information on stair transition but not on steady-state stair walking.

4.3. Effects of hip osteoarthritis on turning kinematics

In their study on 45° turns during walking, Tateuchi and colleagues (63) found reduced peak hip extension angles, similar to level walking. Additionally, peak hip flexion was also reduced. The step turn has been found to require more hip abduction than straight walking, while the spin turn requires greater hip adduction (78). HOA subjects showed decreased peak abduction or adduction during these tasks, respectively (63), but our meta-analysis did not yield these results for straight walking. Thus, analysis of turning while walking might be a beneficial extension of common gait analysis, especially as turning movements are encountered frequently during activities of daily living (79). However, further studies on turning are needed to confirm these results and also to expand the knowledge on transverse plane kinematics, which are also more demanding during turning than straight walking (78).

4.4. Limitations

Alongside the strengths of our study, there are also some limitations.

As stated before, a lot of the calculations were based on a limited number of studies and thus have to be interpreted with caution. Yet, the results might still be helpful to identify research gaps and support hypothesis formulation for future studies.

To have sufficient data for the meta-analysis, studies using different gait conditions (overground vs. treadmill) as well as different walking speeds and subject characteristics were included in the same analysis. Thus, the calculated SMD represents the average effect across a variety of study designs, measurement methods and subject populations. We therefore included prediction intervals in our analyses to provide a range for a potential HOA effect that might occur in individual study settings. Most of the significant results of our meta-analysis prediction intervals cross zero, meaning that in some settings no difference between HOA and CON subjects might be present. However, it has to be borne in mind that prediction intervals may be overly wide when they are calculated from a limited number of studies or from studies at a higher risk of bias (80).

Likewise, we included the I^2 statistic to estimate between-study heterogeneity. However, the I^2 value can be imprecise and biased (81), especially in small meta-analyses such as the present one. Therefore, I^2 estimates have to be interpreted with caution.

Lastly, we refrained from calculating a meta-regression to explore potential sources for between-study heterogeneity due to the small number of included studies (26).

4.5. Conclusion

In summary, this was the first review to synthesise data on lower body joint kinematics during locomotion movements in HOA subjects. A total of 47 reports from 35 individual studies were retrieved.

Most studies focused on gait, where kinematic differences were found in the ipsi- and contralateral hip and knee joints. While changes at the hip occurred in all 3 motion planes, changes in knee kinematics occurred mainly in the sagittal plane. Differences were found between subjects with mild or moderate HOA and those with severe HOA. Thus, motion analysis for HOA patients should not exclusively focus on the kinematics of the affected hip joint but also include analysis of adjacent and contralateral joints. Despite no significant results of the meta-analysis in ankle or pelvis kinematics, several indications exist for further analyses in this area. Additionally, 3 studies on stair walking and turning while walking were reviewed, and both of these movement tasks might be promising extensions to clinical movement analysis due to their elevated requirements on joint mobility.

Overall, large heterogeneity was observed across studies, so future studies have to further clarify the role of factors such as OA severity, laterality, age, gender, study design or movement execution in the analysis of lower limb joint kinematics.

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.

Author contributions

HS, SSp, SSe, and TS contributed to conception of this review. HS and SSp searched the literature, conducted title and abstract screening, full text analysis for eligibility as well as risk of bias assessment. TS resolved disagreements during study eligibility analysis as well as during risk of bias assessment. HS conducted the statistical analyses and wrote the first draft of the manuscript. All authors contributed to the article and approved the submitted version.

Funding

We acknowledge support by the KIT-Publication Fund of the Karlsruhe Institute of Technology.

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.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2023.1197883/full#supplementary-material

References

1. Blanco FJ, Silva-Díaz M, Quevedo Vila V, Seoane-Mato D, Pérez Ruiz F, Juan-Mas A, et al. Prevalence of symptomatic osteoarthritis in Spain: EPISER2016 study. *Reumatol Clin (Engl Ed)*. (2021) 17:461–70. doi: 10.1016/j.reumae.2020.01.005

2. Salaffi F, Carotti M, Stancati A, Grassi W. Health-related quality of life in older adults with symptomatic hip and knee osteoarthritis: a comparison with matched healthy controls. *Aging Clin Exp Res*. (2005) 17:255–63. doi: 10.1007/BF03324607

3. Broström EW, Esbjörnsson A-C, Heideken Jv, Iversen MD. Gait deviations in individuals with inflammatory joint diseases and osteoarthritis and the usage of three-dimensional gait analysis. *Best Pract Res Clin Rheumatol*. (2012) 26:409–22. doi: 10.1016/j.berh.2012.05.007

4. Ornetti P, Maillefert J-F, Laroche D, Morisset C, Dougados M, Gossec L. Gait analysis as a quantifiable outcome measure in hip or knee osteoarthritis: a systematic review. *Joint Bone Spine*. (2010) 77:421–5. doi: 10.1016/j.jbspin.2009.12.009

5. Davis RB. Reflections on clinical gait analysis. *J Electromyogr Kinesiol*. (1997) 7:251–7. doi: 10.1016/S1050-6411(97)00008-4

6. Camomilla V, Cereatti A, Cutti AG, Fantozzi S, Stagni R, Vannozzi G. Methodological factors affecting joint moments estimation in clinical gait analysis: a systematic review. *Biomed Eng Online*. (2017) 16:106. doi: 10.1186/s12938-017-0396-x

7. Rahman J, Tang Q, Monda M, Miles J, McCarthy I. Gait assessment as a functional outcome measure in total knee arthroplasty: a cross-sectional study. *BMC Musculoskelet Disord*. (2015) 16:66. doi: 10.1186/s12891-015-0525-2

8. Diamond LE, Allison K, Dobson F, Hall M. Hip joint moments during walking in people with hip osteoarthritis: a systematic review and meta-analysis. *Osteoarthritis Cartilage*. (2018) 26:1415–24. doi: 10.1016/j.joca.2018.03.011

9. Emmerzaal J, van Rossom S, van der Straaten R, Brabandere A, Corten K, Baets L, et al. Joint kinematics alone can distinguish hip or knee osteoarthritis patients from asymptomatic controls with high accuracy. *J Orthop Res*. (2022) 40:2229–39. doi: 10.1002/jor.25269

10. Petraglia F, Scarcella L, Pedrazzi G, Brancato L, Puers R, Costantino C. Inertial sensors versus standard systems in gait analysis: a systematic review and meta-analysis. *Eur J Phys Rehabil Med*. (2019) 55:265–80. doi: 10.23736/S1973-9087.18.05306-6

11. Eitzen I, Fernandes L, Nordsletten L, Risberg MA. Sagittal plane gait characteristics in hip osteoarthritis patients with mild to moderate symptoms compared to healthy controls: a cross-sectional study. *BMC Musculoskelet Disord*. (2012) 13:258. doi: 10.1186/1471-2474-13-258

12. Kumar D, Wyatt C, Chiba K, Lee S, Nardo L, Link TM, et al. Anatomic correlates of reduced hip extension during walking in individuals with mild-moderate radiographic hip osteoarthritis. *J Orthop Res*. (2015) 33:527–34. doi: 10.1002/jor.22781

13. Diamond LE, Hoang HX, Barrett RS, Loureiro A, Constantinou M, Lloyd DG, et al. Individuals with mild-to-moderate hip osteoarthritis walk with lower hip joint contact forces despite higher levels of muscle co-contraction compared to healthy individuals. *Osteoarthritis Cartilage*. (2020) 28:924–31. doi: 10.1016/j.joca.2020.04.008

14. Rutherford D, Moreside J, Wong I. Knee joint motion and muscle activation patterns are altered during gait in individuals with moderate hip osteoarthritis compared to asymptomatic cohort. *Clin Biomech (Bristol, Avon)*. (2015) 30:578–84. doi: 10.1016/j.clinbiomech.2015.04.002

15. Leigh RJ, Osis ST, Ferber R. Kinematic gait patterns and their relationship to pain in mild-to-moderate hip osteoarthritis. *Clin Biomech*. (2016) 34:12–7. doi: 10.1016/j.clinbiomech.2015.12.010

16. Watelain E, Dujardin F, Babier F, Dubois D, Allard P. Pelvic and lower limb compensatory actions of subjects in an early stage of hip osteoarthritis. *Arch Phys Med Rehabil.* (2001) 82:1705–11. doi: 10.1053/apmr.2001.26812

17. Reininga IHF, Stevens M, Wagenmakers R, Bulstra SK, Groothoff JW, Zijlstra W. Subjects with hip osteoarthritis show distinctive patterns of trunk movements during gait-a body-fixed-sensor based analysis. *J Neuroeng Rehabil.* (2012) 9:1–8. doi: 10.1186/1743-0003-9-3

18. Constantinou M, Barrett R, Brown M, Mills P. Spatial-temporal gait characteristics in individuals with hip osteoarthritis: a systematic literature review and meta-analysis. *J Orthop Sports Phys Ther.* (2014) 44:291–B7. doi: 10.2519/jospt.2014.4634

19. Grimmer M, Elshamanhory AA, Beckerle P. Human lower limb joint biomechanics in daily life activities: a literature based requirement analysis for anthropomorphic robot design. *Front Robot AI.* (2020) 7:13. doi: 10.3389/frobt.2020.00013

20. Zügner R, Tranberg R, Lisovskaja V, Kärrholm J. Different reliability of instrumented gait analysis between patients with unilateral hip osteoarthritis, unilateral hip prosthesis and healthy controls. *BMC Musculoskelet Disord.* (2018) 19:224. doi: 10.1186/s12891-018-2145-0

21. Higgins JP, Thomas J, Chandler J, Cumpston M, Li T, Page MJ, et al. *Cochrane handbook for systematic reviews of interventions version 6.3.* (2022). Available at: www.training.cochrane.org/handbook (Updated February 2022).

22. Cohen J. *Statistical power analysis for the behavioral sciences.* Hove, London: Lawrence Erlbaum Associates (1988). 567.

23. Higgins JPT, Thompson SG, Deeks JJ, Altman DG. Measuring inconsistency in meta-analyses. *Br Med J.* (2003) 327:557–60. doi: 10.1136/bmj.327.7414.557

24. Higgins JPT, Thompson SG, Spiegelhalter DJ. A re-evaluation of random-effects meta-analysis. *J R Stat Soc Ser A Stat Soc.* (2009) 172:137–59. doi: 10.1111/j.1467-985X.2008.00552.x

25. IntHout J, Ioannidis JPA, Rovers MM, Goeman JJ. Plea for routinely presenting prediction intervals in meta-analysis. *BMJ Open.* (2016) 6:e010247. doi: 10.1136/bmjopen-2015-010247

26. Borenstein M. *Introduction to meta-analysis.* Chichester: Wiley (2009). 221f p.

27. López-López JA, Page MJ, Lipsey MW, Higgins JPT. Dealing with effect size multiplicity in systematic reviews and meta-analyses. *Res Synth Methods.* (2018) 9 (3):336–51. doi: 10.1002/jrsm.1310

28. Downs SH, Black N. The feasibility of creating a checklist for the assessment of the methodological quality both of randomised and non-randomised studies of health care interventions. *J Epidemiol Community Health.* (1998) 52:377–84. doi: 10.1136/jech.52.6.377

29. Page MJ, McKenzie JE, Bossuyt PM, Boutron I, Hoffmann TC, Mulrow CD, et al. The PRISMA 2020 statement: an updated guideline for reporting systematic reviews. *Br Med J.* (2021) 372:n71. doi: 10.1136/bmj.n71

30. Aminian K, Trevisan C, Najafi B, Dejnabadi H, Frigo C, Pavan E, et al. Evaluation of an ambulatory system for gait analysis in hip osteoarthritis and after total hip replacement. *Gait Posture.* (2004) 20:102–7. doi: 10.1016/S0966-6362(03)00093-6

31. Ardestani MM, Wimmer MA. Can a linear combination of gait principal component vectors identify hip OA stages? *J Biomech.* (2016) 49:2023–30. doi: 10.1016/j.jbiomech.2016.04.040

32. Baker M, Moreside J, Wong I, Rutherford DJ. Passive hip movement measurements related to dynamic motion during gait in hip osteoarthritis. *J Orthop Res.* (2016) 34:1790–7. doi: 10.1002/jor.23198

33. Rutherford DJ, Moreside J, Wong I. Hip joint motion and gluteal muscle activation differences between healthy controls and those with varying degrees of hip osteoarthritis during walking. *J Electromyogr Kinesiol.* (2015) 25:944–50. doi: 10.1016/j.jelekin.2015.10.010

34. Bejek Z, Paróczai R, Illyés A, Kiss RM. The influence of walking speed on gait parameters in healthy people and in patients with osteoarthritis. *Knee Surg Sports Traumatol Arthrosc.* (2006) 14:612–22. doi: 10.1007/s00167-005-0005-6

35. Benedetti MG, Berti L, Frizziero A, Ferrarese D, Giannini S. Functional recovery after hip resurfacing and rehabilitation. *J Sport Rehabil.* (2012) 21:167–74. doi: 10.1123/jsr.21.2.167

36. Bolink SAAN, Brunton LR, van Laarhoven S, Lipperts M, Heyligers IC, Blom AW, et al. Frontal plane pelvic motion during gait captures hip osteoarthritis related disability. *Hip Int.* (2015) 25:413–9. doi: 10.5301/hipint.5000282

37. Brand RA, Crowninshield RD. The effect of cane use on hip contact force. *Clin Orthop Relat Res.* (1980) 147:181–4. doi: 10.1097/00003086-198003000-00030

38. Crowinshield RD, Brand RA, Johnston RC. The effects of walking velocity and age on hip kinematics and kinetics. *Clin Orthop Relat Res.* (1978):140–4.

39. Constantinou M, Loureiro A, Carty C, Mills P, Barrett R. Hip joint mechanics during walking in individuals with mild-to-moderate hip osteoarthritis. *Gait Posture.* (2017) 53:162–7. doi: 10.1016/j.gaitpost.2017.01.017

40. Foucher KC. Sex-specific hip osteoarthritis-associated gait abnormalities: alterations in dynamic hip abductor function differ in men and women. *Clin Biomech.* (2017) 48:24–9. doi: 10.1016/j.clinbiomech.2017.07.002

41. Foucher KC, Hurwitz DE, Wimmer MA. Preoperative gait adaptations persist one year after surgery in clinically well-functioning total hip replacement patients. *J Biomech.* (2007) 40:3432–7. doi: 10.1016/j.jbiomech.2007.05.020

42. Foucher KC, Wimmer MA. Contralateral hip and knee gait biomechanics are unchanged by total hip replacement for unilateral hip osteoarthritis. *Gait Posture.* (2012) 35:61–5. doi: 10.1016/j.gaitpost.2011.08.006

43. Foucher KC, Wimmer MA, Moisio KC, Hildebrand M, Berli MC, Walker MR, et al. Time course and extent of functional recovery during the first postoperative year after minimally invasive total hip arthroplasty with two different surgical approaches–a randomized controlled trial. *J Biomech.* (2011) 44:372–8. doi: 10.1016/j.jbiomech.2010.10.026

44. Hall M, Wrigley TV, Kean CO, Metcalf BR, Bennell KL. Hip biomechanics during stair ascent and descent in people with and without hip osteoarthritis. *J Orthop Res.* (2017) 35:1505–14. doi: 10.1002/jor.23407

45. Hara D, Nakashima Y, Hamai S, Higaki H, Ikebe S, Shimoto T, et al. Dynamic hip kinematics in patients with hip osteoarthritis during weight-bearing activities. *Clin Biomech.* (2016) 32:150–6. doi: 10.1016/j.clinbiomech.2015.11.019

46. Hara D, Nakashima Y, Hamai S, Higaki H, Ikebe S, Shimoto T, et al. Kinematic analysis of healthy hips during weight-bearing activities by 3D-to-2D model-to-image registration technique. *Biomed Res Int.* (2014) 2014:457573. doi: 10.1155/2014/457573

47. Hurwitz DE, Hulet CH, Andriacchi TP, Rosenberg AG, Galante JO. Gait compensations in patients with osteoarthritis of the hip and their relationship to pain and passive hip motion. *J Orthop Res.* (1997) 15:629–35. doi: 10.1002/jor.1100150421

48. Ismailidis P, Kaufmann M, Clauss M, Pagenstert G, Eckardt A, Ilchmann T, et al. Kinematic changes in severe hip osteoarthritis measured at matched gait speeds. *J Orthop Res.* (2021) 39:1253–61. doi: 10.1002/jor.24858

49. Ismailidis P, Nuesch C, Kaufmann M, Clauss M, Pagenstert G, Eckardt A, et al. Measuring gait kinematics in patients with severe hip osteoarthritis using wearable sensors. *Gait Posture.* (2020) 81:49–55. doi: 10.1016/j.gaitpost.2020.07.004

50. Nüesch C, Ismailidis P, Koch D, Pagenstert G, Ilchmann T, Eckardt A, et al. Assessing site specificity of osteoarthritic gait kinematics with wearable sensors and their association with patient reported outcome measures (PROMs): knee versus hip osteoarthritis. *Sensors.* (2021) 21:5363. doi: 10.3390/s21165363

51. Kataoka Y, Shimizu T, Takeda R, Tadano S, Saito Y, Osuka S, et al. Effects of unweighting on gait kinematics during walking on a lower-body positive-pressure treadmill in patients with hip osteoarthritis. *BMC Musculoskelet Disord.* (2021) 22:1–9. doi: 10.1186/s12891-020-03909-8

52. Kubota M, Shimada S, Kobayashi S, Sasaki S, Kitade I, Matsumura M, et al. Quantitative gait analysis of patients with bilateral hip osteoarthritis excluding the influence of walking speed. *J Orthop Sci.* (2007) 12:451–7. doi: 10.1007/s00776-007-1160-z

53. Meyer CAG, Wesseling M, Corten K, Nieuwenhuys A, Monari D, Simon J-P, et al. Hip movement pathomechanics of patients with hip osteoarthritis aim at reducing hip joint loading on the osteoarthritic side. *Gait Posture.* (2018) 59:11–7. doi: 10.1016/j.gaitpost.2017.09.020

54. Wesseling M, Meyer C, Corten K, Desloovere K, Jonkers I. Longitudinal joint loading in patients before and up to one year after unilateral total hip arthroplasty. *Gait Posture.* (2018) 61:117–24. doi: 10.1016/j.gaitpost.2018.01.002

55. Ornetti P, Laroche D, Morisset C, Beis JN, Tavernier C, Maillefert J-F. Three-dimensional kinematics of the lower limbs in hip osteoarthritis during walking. *J Back Musculoskelet Rehabil.* (2011) 24:201–8. doi: 10.3233/BMR-2011-0295

56. Popovic T, Samaan MA, Link TM, Majumdar S, Souza RB. Patients with symptomatic hip osteoarthritis have altered kinematics during stair ambulation. *PM R.* (2021) 13:128–36. doi: 10.1002/pmrj.12398

57. Porta M, Pau M, Leban B, Deidda M, Sorrentino M, Arippa F, et al. Lower limb kinematics in individuals with hip osteoarthritis during gait: a focus on adaptative strategies and interlimb symmetry. *Bioengineering.* (2021) 8:47. doi: 10.3390/bioengineering8040047

58. Schmidt A, Meurer A, Lenarz K, Vogt L, Froemel D, Lutz F, et al. Unilateral hip osteoarthritis: the effect of compensation strategies and anatomic measurements on frontal plane joint loading. *J Orthop Res.* (2017) 35:1764–73. doi: 10.1002/jor.23444

59. Stief F, Schmidt A, van Drongelen S, Lenarz K, Froemel D, Tarhan T, et al. Abnormal loading of the hip and knee joints in unilateral hip osteoarthritis persists two years after total hip replacement. *J Orthop Res.* (2018) 36:2167–77. doi: 10.1002/jor.23886

60. Schmitt D, Vap A, Queen RM. Effect of end-stage hip, knee, and ankle osteoarthritis on walking mechanics. *Gait Posture.* (2015) 42:373–9. doi: 10.1016/j.gaitpost.2015.07.005

61. Steingrebe H, Stetter BJ, Sell S, Stein T. Effects of hip bracing on gait biomechanics, pain and function in subjects with mild to moderate hip osteoarthritis. *Front Bioeng Biotechnol.* (2022) 10:888775. doi: 10.3389/fbioe.2022.888775

62. Tanaka Y. Gait analysis of patients with osteoarthritis of the hip and the those with total hip arthroplasty. *Biomed Mater Eng.* (1998) 8:187–96.

63. Tateuchi H, Tsukagoshi R, Fukumoto Y, Akiyama H, So K, Kuroda Y, et al. Compensatory turning strategies while walking in patients with hip osteoarthritis. *Gait Posture*. (2014) 39:1133–7. doi: 10.1016/j.gaitpost.2014.01.021

64. Thurston AJ. Spinal and pelvic kinematics in osteoarthrosis of the hip joint. *Spine*. (1985) 10:467–71. doi: 10.1097/00007632-198506000-00012

65. van Drongelen S, Braun S, Stief F, Meurer A. Comparison of gait symmetry and joint moments in unilateral and bilateral hip osteoarthritis patients and healthy controls. *Front Bioeng Biotechnol*. (2021) 9:756460. doi: 10.3389/fbioe.2021.756460

66. van Drongelen S, Stetter BJ, Böhm H, Stief F, Stein T, Meurer A. Identification of patients with similar gait compensating strategies due to unilateral hip osteoarthritis and the effect of total hip replacement: a secondary analysis. *J Clin Med*. (2021) 10:2167. doi: 10.3390/jcm10102167

67. van Drongelen S, Kaldowski H, Fey B, Tarhan T, Assi A, Stief F, et al. Determination of leg alignment in hip osteoarthritis patients with the EOS(R) system and the effect on external joint moments during gait. *Appl Sci-Basel*. (2020) 10:7777. doi: 10.3390/app10217777

68. van Drongelen S, Kaldowski H, Tarhan T, Assi A, Meurer A, Stief F. Are changes in radiological leg alignment and femoral parameters after total hip replacement responsible for joint loading during gait? *BMC Musculoskelet Disord*. (2019) 20:1–13. doi: 10.1186/s12891-019-2832-5

69. Watanabe H, Shimada Y, Kagaya H, Sato K. Gait analysis following varus osteotomy of the femur for hip osteoarthritis. *J Orthop Sci*. (1999) 4:89–98. doi: 10.1007/s007760050080

70. Watanabe H, Shimada Y, Sato K, Tsutsumi Y, Sato M. Gait analysis before or after varus osteotomy of the femur for hip osteoarthritis. *Biomed Mater Eng*. (1998) 8:177–86.

71. Wonsetler EC, Miller EL, Huey KL, Frye SE, Bowden MG. Association between altered hip extension and kinetic gait variables. *Am J Phys Med Rehabil*. (2018) 97:131–3. doi: 10.1097/PHM.0000000000000840

72. Laroche D, Tolambiya A, Morisset C, Maillefert JF, French RM, Ornetti P, et al. A classification study of kinematic gait trajectories in hip osteoarthritis. *Comput Biol Med*. (2014) 55:42–8. doi: 10.1016/j.compbiomed.2014.09.012

73. Ridgway HM, Bisson EJ, Brouwer B. A review of the physical demands of stair negotiation in healthy aging and following stroke. *Phys Med Rehabil Int*. (2015) 2:1057.

74. Meyer CAG, Corten K, Fieuws S, Deschamps K, Monari D, Wesseling M, et al. Biomechanical gait features associated with hip osteoarthritis: towards a better definition of clinical hallmarks. *J Orthop Res*. (2015) 33:1498–507. doi: 10.1002/jor.22924

75. Meyer CAG, Corten K, Fieuws S, Deschamps K, Monari D, Wesseling M, et al. Evaluation of stair motion contributes to new insights into hip osteoarthritis-related motion pathomechanics. *J Orthop Res*. (2016) 34:187–96. doi: 10.1002/jor.22990

76. Cui W, Wang C, Chen W, Guo Y, Jia Y, Du W, et al. Effects of toe-out and toe-in gaits on lower-extremity kinematics. *Dyn Electromyogr Appl Sci*. (2019) 9:5245. doi: 10.3390/app9235245

77. Alcock L, O'Brien TD, Vanicek N. Biomechanical demands differentiate transitioning vs. continuous stair ascent gait in older women. *Clin Biomech (Bristol Avon)*. (2014) 29:111–8. doi: 10.1016/j.clinbiomech.2013.10.007

78. Taylor MJD, Dabnichki P, Strike SC. A three-dimensional biomechanical comparison between turning strategies during the stance phase of walking. *Hum Mov Sci*. (2005) 24:558–73. doi: 10.1016/j.humov.2005.07.005

79. Glaister BC, Bernatz GC, Klute GK, Orendurff MS. Video task analysis of turning during activities of daily living. *Gait Posture*. (2007) 25:289–94. doi: 10.1016/j.gaitpost.2006.04.003

80. Riley RD, Higgins JPT, Deeks JJ. Interpretation of random effects meta-analyses. *Br Med J*. (2011) 342:d549. doi: 10.1136/bmj.d549

81. Hippel PT. The heterogeneity statistic I(2) can be biased in small meta-analyses. *BMC Med Res Methodol*. (2015) 15:35. doi: 10.1186/s12874-015-0024-z

frontiers | Frontiers in Sports and Active Living

TYPE Original Research
PUBLISHED 26 March 2024
DOI 10.3389/fspor.2024.1323598

Check for updates

OPEN ACCESS

EDITED BY
Kiros Karamanidis,
London South Bank University,
United Kingdom

REVIEWED BY
Gaspar Epro,
London South Bank University,
United Kingdom
Josh Walker,
Leeds Beckett University, United Kingdom
Marco Aurelio Vaz,
Federal University of Rio Grande do Sul, Brazil

*CORRESPONDENCE
Mutsuaki Edama
✉ edama@nuhw.ac.jp

[†]These authors share first authorship

RECEIVED 28 November 2023
ACCEPTED 12 March 2024
PUBLISHED 26 March 2024

CITATION
Saito R, Shagawa M, Sugimoto Y, Hirai T, Kato K, Sekine C, Yokota H, Hirabayashi R, Ishigaki T, Akuzawa H, Togashi R, Yamada Y, Osanami H and Edama M (2024) Changes in the mechanical properties of the thigh and lower leg muscle-tendon units during the early follicular and early luteal phases.
Front. Sports Act. Living 6:1323598.
doi: 10.3389/fspor.2024.1323598

Changes in the mechanical properties of the thigh and lower leg muscle-tendon units during the early follicular and early luteal phases

Rina Saito[†], Mayuu Shagawa, Yuzuka Sugimoto, Tomoki Hirai, Koyo Kato, Chie Sekine, Hirotake Yokota, Ryo Hirabayashi, Tomonobu Ishigaki, Hiroshi Akuzawa, Ryoya Togashi, Yuki Yamada, Haruki Osanami and Mutsuaki Edama[*†]

Institute for Human Movement and Medical Sciences, Niigata University of Health and Welfare, Niigata, Japan

Background: This study aimed to determine changes in the muscle and tendon stiffness of the thigh and lower leg muscle-tendon units during the early follicular and early luteal phases, and check for possible relations between muscle and tendon stiffness in each phase.

Methods: The sample consisted of 15 female university students with regular menstrual cycles. The basal body temperature method, ovulation kit, and salivary estradiol concentration measurement were used to estimate the early follicular and early luteal phases. A portable digital palpation device measured muscle-tendon stiffness in the early follicular and early luteal phases. The measurement sites were the rectus femoris (RF), vastus medialis (VM), patellar tendon (PT), medial head of gastrocnemius muscle, soleus muscle, and Achilles tendon.

Results: No statistically significant differences in the thigh and lower leg muscle-tendon unit stiffness were seen between the early follicular and early luteal phases. Significant positive correlations were found between the stiffness of the RF and PT ($r = 0.608$, $p = 0.016$) and between the VM and PT ($r = 0.737$, $p = 0.002$) during the early luteal phase.

Conclusion: The present results suggest that the stiffness of leg muscle-tendon units of the anterior thigh and posterior lower leg do not change between the early follicular and early luteal phases and that tendons may be stiffer in those women who have stiffer anterior thigh muscles during the early luteal phase.

KEYWORDS

menstrual cycle, mechanical properties, stiffness, estradiol, muscle-tendon complex

Introduction

The menstrual cycle is regulated by various regular fluctuations in female steroid hormones (1–3). And based on these fluctuations, specifically the gonadotropin-releasing hormone from the hypothalamus, follicle-stimulating hormone, and luteinizing hormone from the anterior pituitary, and estrogen (E2) and progesterone (P4) from the ovary gland (3), it is classified into four phases. These changes in hormonal concentrations may have an effect on the mechanical properties of muscles and tendons.

A previous study investigating the rate of muscle-tendon injuries has reported that the injuries are 88% higher during the late follicular phase when E2 levels are maximal than during the follicular phase. In addition, compared with other phases, muscle tears, strains, spasms, tendon disorders, and tendon ruptures have been shown to occur more than twice as often during the late follicular phase (4). These findings suggest that female hormones may affect the structure of muscles, tendons and ligaments.

A relationship between female hormones and the structure of the muscle-tendon unit has been reported regarding the E2 receptor expression in myofibers (5) and tendons (6). E2 has also been found to act on the type I collagen synthesis and fibroblast proliferation, as well as type I collagen degradation by matrix metalloproteases (MMPs) (7). A previous study that treated fibroblasts detached from human thigh fascia with E2, followed by culturing, found decreases in type I collagen and increases in type III collagen and fibrillin, which enhanced the elasticity of the fascia, when E2 concentrations were increased to a level equivalent to that present in humans before ovulation (8). Therefore, multiple pathways exist for the effect of E2 on muscles and tendons, and different tissues appear to be affected in different ways, which suggests that changes in mechanical properties during the menstrual cycle vary by individual.

However, an *in vivo* study using ultrasonography that examined the mechanical properties of human muscle-tendon units during the early follicular, ovulatory, and luteal phases found no significant changes in maximal isometric voluntary contraction or muscle activation levels and tendon properties (maximal elongation and stiffness) of the knee extensors and ankle plantar flexors during the menstrual cycle (9). On the other hand, studies using shear-wave elastography have demonstrated lower muscle stiffness during contraction in the ovulation compared to menstruation phase (10) however no menstrual cycle-induced effects on mechanical properties of the muscle-tendon units (11). As described above, studies have been conducted using ultrasonography and shear-wave elastography, which are considered the golden standards for evaluating muscle and tendon units. On the other hand, recently, several studies have used a noninvasive digital palpation device (MyotonPRO; Myoton AS, Tallinn, Estonia) to evaluate the mechanical properties of human muscle (12–14). The MyotonPro is a non-invasive hand-held, affordable, and easy-to-use myotonometer device aimed at recording the biomechanical and viscoelastic stiffness of myofascial tissues (15, 16). In addition, the short measurement time also has the advantage of allowing measurement of a large number of objects. One of these studies that examined changes in the mechanical properties of the thigh musculature during the early follicular, ovulatory and luteal phases reported increased stiffness of the vastus medialis and semitendinosus during the ovulatory compared with the luteal phase, but no changes in the stiffness of the vastus lateralis or biceps femoris throughout the menstrual cycle (12). A previous study that investigated the mechanical properties of lower leg muscle groups during the early follicular and ovulatory phases found no changes in the stiffness of the peroneus longus, tibialis anterior, or medial head of the gastrocnemius (13, 14). Thus, a sufficient consensus has yet to be reached. Therefore, to determine whether the mechanical properties of the muscle-tendon units change during the same menstrual cycle, it is necessary to examine muscle and tendon separately, site-specific differences, and the relationship between each.

Given this background and assuming that different ovarian hormonal concentrations may affect the E2 receptor of the parallel elastic elements, thereby changing its collagen content and corresponding mechanical resistance, the present study aimed to investigate changes in stiffness of the muscles and tendons of the anterior aspect of the thigh and the posterior aspect of the lower leg during the early follicular and early luteal phases and to determine the existence of correlations between muscular and tendinous stiffness in female university students. We hypothesized that, compared with the early follicular phase, the stiffness of the thigh and lower leg muscles and tendons would not decrease during the ovulatory phase and that correlations would not be found between the stiffness of muscle-tendon units.

Materials and methods

Subjects

We conducted a questionnaire survey on 88 female university students. The inclusion criteria were as follows: (1) menstruation approximately 10 times/year, with a menstrual cycle ranging from 25 to 38 days (17); (2) not currently exercising more than twice a week or holding a membership in a designated strength or athletic club (17); (3) no use of oral contraceptives or other hormonal agents within the previous 6 months (18); and (4) no history of doctor-diagnosed disorders or surgery on any part of the lower limb. Of the 88 students, 15 [mean age ± standard deviation (SD), 20 ± 0.5 years; height, 159.2 cm ± 7.1 cm; body mass, 55.9 ± 7.5 kg] met the inclusion criteria and provided written informed consent to participate in the study after receiving a full explanation of all study contents (Figure 1). All

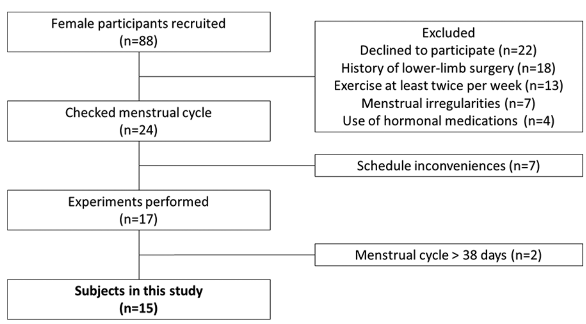

FIGURE 1
Flowchart for the selection of the female participants.

study protocols were carried out according to the Declaration of Helsinki after receiving approval from our institutional ethics committee (approval No. 17946).

Recording of the menstrual cycle

For 1–2 months before the start of the experiment, all participants were asked to measure their basal body temperature every morning after waking up using an electronic basal thermometer (CTEB503l; Citizen Systems, Tokyo, Japan). The date of ovulation was then estimated using an ovulation kit (Doctor's Choice One Step Ovulation Test Clear; Beauty and Health Research, Torrance, CA, USA). When urinary LH was 20 mIU/ml or higher, a red line was displayed on the kit, indicating the start of an LH surge, and about 24 h later was estimated to be the day of ovulation. Using a conditioning management system (ONE TAP SPORTS; Euphoria Corporation, Tokyo), each subject recorded the results of daily basal body temperature, menstruation, and ovulation kits on their own (17–20).

Timing of measurements

The mechanical properties of the muscle-tendon unit and the concentrations of salivary E2 and P4 were each measured once on the same day during the early follicular and early luteal phases. The measurements in the early follicular phase were recorded on the second to fourth day after menstruation onset, and those in the early luteal phase on the second to fourth day after a positive result from the ovulation kit (17–20). All measurements were conducted from 08:00 to 12:00 to allow for diurnal variations under a room temperature of 20–25°C.

Measurement methods

Concentrations of estradiol and progesterone

The concentrations of E2 and P4 were measured from saliva. Based on previous studies (17–20), the participants were asked to observe the following six points strictly before collecting saliva to help avoid the possibility of influencing E2 and P4 concentrations: (1) refrain from consuming alcohol for at least 12 h prior to measurement; (2) refrain from consuming food for at least 60 min prior to measurement; (3) refrain from toothbrushing at least 45 min prior to measurement; (4) refrain from consuming dairy products for at least 20 min prior to measurement; (5) refrain from consuming beverages with high sugar content, high acid content, or caffeine; and (6) in the case of dental treatment, avoid collecting saliva within 48 h after treatment. All participants were also asked to rinse their mouth before the beginning of the experiment to ensure that no food particles remained in the mouth. To prevent a decrease in E2 and P4 concentrations, saliva was collected at least 10 min after rinsing the mouth using a special straw (Saliva Collection Aid; Salimetrics, State College, PA, USA) and then placed in a saliva collection vessel (Cryovial; Salimetrics) after collection in the mouth for 1 min. Next, all saliva samples were immediately frozen in a freezer at −80°C or colder. E2 concentrations were analyzed by Funakoshi Corporation (Tokyo, Japan) after all samples had been collected. The samples were then thawed at room temperature, mixed immediately by vortexing, centrifuged at 1,500 × g for 15 min, and subjected to analysis by enzyme-linked immunosorbent assay (17β-Estradiol and Salivary Progesterone Enzyme Immunoassay Kits; Salimetrics). The dilution factor was uniformly onefold (undiluted solution) (17–20).

Mechanical properties of the muscles and tendons

The stiffness of the muscles and tendons was measured at the following six sites: rectus femoris (RF), vastus medialis (VM), patellar tendon (PT), Medial head of gastrocnemius (MG), soleus (SOL), and Achilles tendon (AT). An ultrasound imaging system (Aplio500; Canon Medical Systems, Tochigi, Japan) and a linear probe (PL0081; Canon Medical Systems) were used to identify all measurement positions, with reference to previous studies (11, 21–23). The RF muscle was identified at half the length from the anterior superior iliac spine (ASIS) to the bottom of the patella (22), the VM at the distal 20% of the length from the ASIS to the bottom of the patella (22), the PT at the midpoint between the patellar apex and tibial rough surface (23), the MG and SOL at the proximal 30% of the length from the orbital skin line to the external capsule (11), and the AT at 3 cm proximal to the tendon attachment site (i.e., the calcaneal tuberosity) (21) (Figure 2). The noninvasive MyotonPRO digital palpation device, which can evaluate the stiffness of muscle on the surface of the skin by making a mark with a pen just above the muscle belly or central part of the tendon, was used to measure the stiffness of the muscles and tendons (Figure 3). The measurement technique used by this device applies five brief mechanical impulses (time, 15 ms; force, 0.4 N) under a steady pre-compression force (0.18 N) of the subcutaneous tissue layer above the evaluated muscle. A device probe (diameter, 3 mm) set perpendicular to the surface of the skin delivers mechanical deformation, and an acceleration sensor connected to the frictionless measurement mechanism detects the damped oscillation produced by the muscle after the brief mechanical impulse. If the Device was rolled more than 10°, a warning notice "Rotate" was displayed. And the device offered the coefficient of variation (CV) for stiffness. The CV described the relative dispersion of the measurement. The CV above 3% was showed in red and re-measured if the CV was above 3%. The highest and lowest results from five measurements were discarded, after which, the average of the three remaining measurements was calculated for each site. All measurements were taken with the axial foot (opposite to the dominant foot). The thigh was measured in the supine position with the knee joint extended, whereas the lower leg was measured in the prone position with the knee joint extended and the foot hanging down over the edge of the bed (ankle-joint resting

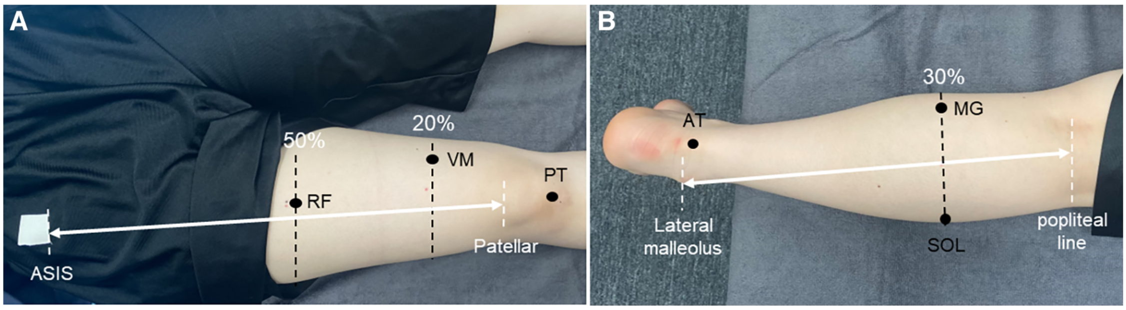

FIGURE 2
Measurement site for mechanical properties. (A) Measurement sites on the thigh. (B) Measurement sites on the lower leg. ASIS, anterior superior iliac spine; RF, rectus femoris; VM, vastus medialis; PT, patellar tendon; AT, achilles tendon; MG, medial head of the gastrocnemius; SOL, soleus.

position). The order of the measurements was randomized for the lower leg, thigh, and each part of the body. Considering that changes in position during measurement may affect the values of the stiffness, all participants were instructed to rest for 10 min after identifying the muscle-tendon units and after changes in the lower leg and thigh position. All measurements were carried out by the same experienced physical therapist (R.S.).

Reliability of measurements

The reliability of the musculotendon mechanical property measurements was examined in five healthy male university students (10 feet; mean age ± SD, 21 ± 0 years; height, 173.6 cm ± 2.2 cm; weight, 63.3 ± 3.9 kg) without any history of orthopedic disease using the same digital device. Repeat measurements were carried out between 2 days and 1 week. The intraclass correlation coefficient (ICC) (1, 3) was used to calculate reliability. The ICCs for the RF, VM, PT, MG, SOL, and AT were 0.802, 0.653, 0.968, 0.746, 0.774, and 0.835, respectively. Based on the criteria of Landis and Koch (24), reliability was considered substantial for an ICC of 0.61–0.80, which was taken to represent practical reliability.

FIGURE 3
Mechanical property measurement using MyotonPRO.

Statistical analysis

IBM SPSS Statistics version 28.0 (IBM Corp., Armonk, NY, USA) was used for the data analysis. Paired *t*-tests were performed to compare E2 and P4 concentrations and the stiffness of the muscles and tendons in the early follicular and early luteal phases, and Pearson's product-moment correlation coefficient to investigate correlations between the thigh and lower leg muscle and tendons. Cohen's *d* was used to calculate effect sizes as follows: trivial (0–0.19), small (0.20–0.49), medium (0.50–0.79), and large (>0.80) (25). Differences at the 5% level were considered to indicate statistical significance.

Results

The results showed that the E2 concentration was significantly higher in the early luteal than in the early follicular phase ($p = 0.017$) (Table 1). However, no significant changes in the P4 concentration were observed between the early follicular and early luteal phases ($p = 0.153$) (Table 1). In

TABLE 1 Changes in estradiol and progesterone concentrations and stiffness during the menstrual cycle.

	Early follicular phase	Early luteal phase	*p*-values	Effect size
Estradiol [pg/ml]	1.44 ± 0.40	1.70 ± 0.40	0.017	0.66
Progesterone [pg/ml]	65.85 ± 38.94	82.73 ± 39.03	0.153	0.43
Stiffness[N/m]				
Rectus femoris	220.38 ± 26.20	222.84 ± 33.06	0.640	0.08
Vastus medialis	201.07 ± 29.29	203.27 ± 29.33	0.561	0.07
Patellar tendon	261.13 ± 48.55	266.89 ± 44.58	0.590	0.12
Medial head of gastrocnemius	263.84 ± 32.82	264.73 ± 30.04	0.834	0.03
Soleus muscle	290.53 ± 42.01	285.29 ± 34.09	0.500	0.13
Achilles tendon	720.07 ± 53.57	713.18 ± 47.82	0.696	0.13

Values are presented as mean ± standard deviation (*n* = 15).
RF, rectus femoris muscle; VM, vastus medialis muscle; PT, patellar tendon; SOL, soleus muscle; MG, medial head of gastrocnemius muscle; AT, achilles tendon.

addition, no significant difference in the stiffness of all muscles and tendons was found between the early follicular and early luteal phases ($p > 0.05$) (Table 1). No correlation was found between various muscles and tendon stiffness in the lower leg (Figure 4A). By contrast, during the early luteal phase, significant positive correlations were found between the RF and PT ($r = 0.608$, $p = 0.016$) and between the VM and PT ($r = 0.737$, $p = 0.002$) in the thigh (Figure 4B).

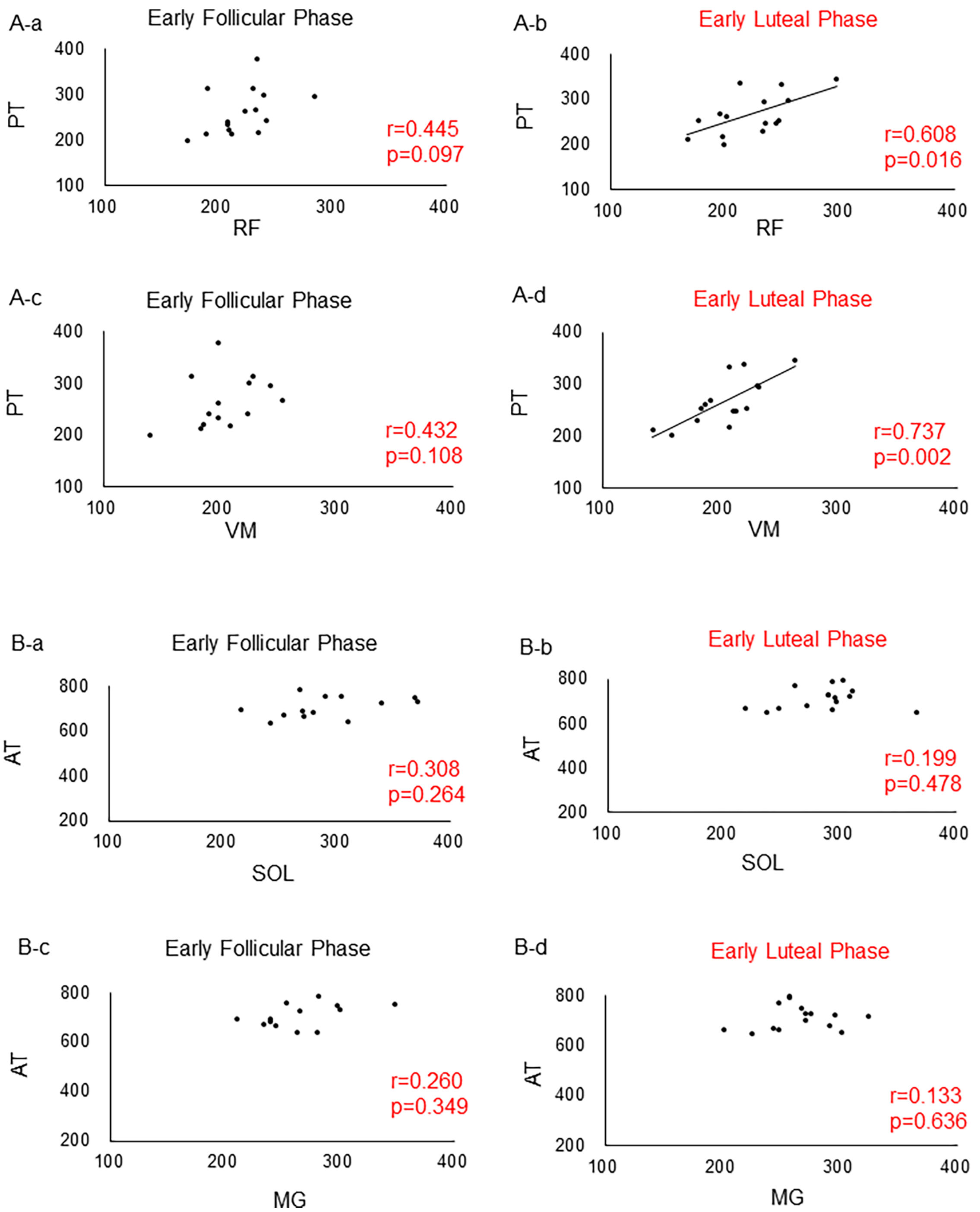

FIGURE 4
Correlations between the lower leg and thigh muscles and tendons. (A) Correlations between muscles and tendons in the thigh. (A) Between RF and PT in the early follicular phase. (B) Between RF and PT in the ovulatory phase. (C) Between VM and PT in the early follicular phase. (D) Between VM and PT in the ovulatory phase. (B) Correlations between muscles and tendons in the lower leg. (A) Between SOL and AT in the early follicular phase. (B) Between SOL and AT in the ovulatory phase. (C) Between MG and AT in the early follicular phase. (D) Between MG and AT in the ovulatory phase. RF, rectus femoris; VM, vastus medialis; PT, patellar tendon; SOL, soleus; MG, medial head of the gastrocnemius; AT, achilles tendon.

Discussion

The present study aimed to aimed to investigate changes in the stiffness of the thigh and lower leg muscles and tendons during the early follicular and early luteal phases of the menstrual cycle among female university students using a noninvasive digital palpation device. To the best of our knowledge, no previous studies have provided a consistent view of changes in the stiffness of muscles and tendons during the menstrual cycle. Therefore, we believe that this constitutes the first study to investigate the relationships between the stiffness of muscles and tendons during the early follicular and early luteal phases of the menstrual cycle using a digital palpation device.

No significant differences in stiffness of any muscles or tendons were found between the early follicular and early luteal phases. In previous study, reported that anterior knee laxity increased during the ovulatory compared with the early follicular phase in females with genu recurvatum (26). In addition, another previous study reported that an increase in the elasticity of the plantar fascia during the ovulatory phase increased postural sway and the risk of falls (27), other previous study found that strenuous exercise resulted in a rapid increase in collagen synthesis in human tendon and muscle the stiffness (28). Although methodologies differ and it is difficult to draw a general conclusion, these previous findings suggest that increased joint laxity in the knee and ankle joints during the ovulatory phase increases the musculotendinous load to compensate for tissue instability and stiffening. In addition, a previous study reported that E2 may affect the degradation of type I collagen by MMPs, which are degrading enzymes during the menstrual cycle (7). Therefore, the action of E2 may counteract this response to maintain joint stability. In addition, although a digital palpation device was used in the present study, previous studies (9, 11) have used a variety of methods to evaluate the mechanical properties of the muscle-tendon units. Furthermore, because opinions differ on issues such as changes in skeletal muscle strength during the menstrual cycle (29–31), further validation is needed, including evaluations of skeletal muscle strength and neural activation levels and verification among various evaluation methods.

In the present study, we found significant positive correlations between the stiffness from the RF and PT and between the VM and PT during the ovulatory phase. A previous study using the same palpation device reported that the stiffness of the VM and ST in the thigh muscle group was higher during the ovulatory than during the luteal phase; however, no such changes were observed for the VL or BF (12). These findings suggest that the quadriceps muscle and the medial side of the hamstrings (ST) may be affected during the ovulatory phase, when the E2 concentration is higher (12). However, a previous study involving females who had undergone reconstruction of the ACL reported no differences in muscle stiffness (ST and BF) on both sides during the early follicular and ovulatory phases (20). In addition, no periodic changes in the stiffness of the PL, TA, or MG were seen in the lower leg muscle group (14). Therefore, the degree of change in the mechanical properties of the muscle-tendon complex remains controversial. However, interestingly, significant correlations were observed only between the stiffness of the RF and PT and between the VM and PT during ovulation in this study. This site-specific change may provide insight into the mechanism of muscle-tendon injury. Therefore, further research on this issue is warranted.

This study has several limitations. First, the sample size was small. A larger sample size is needed to confirm the changes in the mechanical properties of the thigh and lower leg musculotendons during the early follicular and ovulatory phases, and to determine correlations between muscles and tendons. Second, this study does not take into account subcutaneous fat thickness, which may affect measurement results. Third, the measurement limb position (ankle joint position) was not strictly defined. Fourth, the MyotonPRO used in the present study measures only the surface layer of the muscle and a portion of that muscle at rest, and thus, may not capture changes in the entire muscle or deeper layers. In addition, it is unclear whether tendon stiffness by MyotonPRO truly reflects the condition of the muscle when it is contracting. MyotonePro primarily measures oscillation in the transverse direction, not necessarily axial stiffness like shear wave elastography or dynamometry/ultrasound techniques might. Therefore, it may be difficult to apply MyotonPRO measurements to human performance injury. Fifth, this study that the reliability was measured with male subjects, and therefore the reliability obtained with female subjects may be smaller due to possible influences of the menstrual cycle. Finally, although the basal body temperature method, salivary hormone concentration analysis, and an ovulation prediction kit were used to classify the menstrual cycle, whether the early luteal phase was strictly defined remains unclear. Future studies should use blood sampling for more appropriate monitoring of the menstrual cycle.

In conclusion, in the present study, no significant difference in muscle or tendon stiffness was observed in the anterior or lower posterior thigh regions between the early follicular and early luteal phases of menstruation. Only the anterior thigh region showed a significant positive correlation with the early luteal phases. These results suggest that the stiffness of the muscle-tendon units of the anterior and posterior thighs do not change between the early follicular and early luteal phases, and that tendons may be stiffer in those who have stiffer anterior thigh muscles during the early luteal phases.

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 study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Ethics Committee of Niigata University of Health and Welfare (18,671—20 July 2021). The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

RS: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Validation, Visualization, Writing – original draft, Writing – review & editing. MS: Investigation, Writing – review & editing. YS: Investigation, Validation, Writing – review & editing. TH: Validation, Writing – review & editing. KK: Validation, Writing – review & editing. CS: Validation, Writing – review & editing. HY: Validation, Writing – review & editing. RH: Validation, Writing – review & editing. TI: Validation, Writing – review & editing. HA: Validation, Writing – review & editing. RT: Validation, Writing – review & editing. YY: Validation, Writing – review & editing. HO: Validation, Writing – review & editing. ME: Conceptualization, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Supervision, Validation, Writing – original draft, Writing – review & editing.

Funding

The authors declare financial support was received for the research, authorship, and/or publication of this article.

This study was supported by a Grant-in-Aid for Scientific Research (grant no. 22K19739) from the Japan Society for the Promotion of Science (JSPS), commissioned by the Japan Sports Agency (Female Athletes Development and Support Projects 2022, 2023), and a Grant-in-Aid program from Niigata University of Health and Welfare.

Acknowledgments

The researchers appreciate and thank all the female volunteers who took part in this study. The researchers would also like to thank FORTE (https://forte-science.co.jp/) for the English language review.

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.

Publisher's note

References

1. Beynnon BD, Johnson RJ, Braun S, Sargent M, Bernstein IM, Skelly JM, et al. The relationship between menstrual cycle phase and anterior cruciate ligament injury: a case-control study of recreational alpine skiers. *Am J Sports Med.* (2006) 34:757–64. doi: 10.1177/0363546505282624

2. Schaumberg MA, Jenkins DG, Janse De Jonge XAK, Emmerton LM, Skinner TL. Three-step method for menstrual and oral contraceptive cycle verification. *J Sci Med Sport.* (2017) 20:965–9. doi: 10.1016/j.jsams.2016.08.013

3. Janse DEJX, Thompson B, Han A. Methodological recommendations for menstrual cycle research in sports and exercise. *Med Sci Sports Exerc.* (2019) 51:2610–7. doi: 10.1249/MSS.0000000000002073

4. Martin D, Timmins K, Cowie C, Alty J, Mehta R, Tang A, et al. Injury incidence across the menstrual cycle in international footballers. *Front Sports Act Living.* (2021) 3:616999. doi: 10.3389/fspor.2021.616999

5. Wiik A, Ekman M, Morgan G, Johansson O, Jansson E, Esbjörnsson M. Oestrogen receptor beta is present in both muscle fibres and endothelial cells within human skeletal muscle tissue. *Histochem Cell Biol.* (2005) 124:161–5. doi: 10.1007/s00418-005-0030-z

6. Bridgeman JT, Zhang Y, Donahue H, Wade AM, Juliano PJ. Estrogen receptor expression in posterior tibial tendon dysfunction: a pilot study. *Foot Ankle Int.* (2010) 31:1081–4. doi: 10.3113/FAI.2010.1081

7. Leblanc DR, Schneider M, Angele P, Vollmer G, Docheva D. The effect of estrogen on tendon and ligament metabolism and function. *J Steroid Biochem Mol Biol.* (2017) 172:106–16. doi: 10.1016/j.jsbmb.2017.06.008

8. Fede C, Pirri C, Fan C, Albertin G, Porzionato A, Macchi V, et al. Sensitivity of the fasciae to sex hormone levels: modulation of collagen-I, collagen-III and fibrillin production. *PLoS One.* (2019) 14:e0223195. doi: 10.1371/journal.pone.0223195

9. Kubo K, Miyamoto M, Tanaka S, Maki A, Tsunoda N, Kanehisa H. Muscle and tendon properties during menstrual cycle. *Int J Sports Med.* (2009) 30:139–43. doi: 10.1055/s-0028-1104573

10. Ham S, Kim S, Choi H, Lee Y, Lee H. Greater muscle stiffness during contraction at menstruation as measured by shear-wave elastography. *Tohoku J Exp Med.* (2020) 250:207–13. doi: 10.1620/tjem.250.207

11. Saeki J, Ikezoe T, Yoshimi S, Nakamura M, Ichihashi N. Menstrual cycle variation and gender difference in muscle stiffness of triceps surae. *Clin Biomech.* (2019) 61:222–6. doi: 10.1016/j.clinbiomech.2018.12.013

12. Sung ES, Kim JH. The difference effect of estrogen on muscle tone of medial and lateral thigh muscle during ovulation. *J Exerc Rehabil.* (2018) 14:419–23. doi: 10.12965/jer.1836110.055

13. Khowailed IA, Lee H. Neuromuscular control of ankle-stabilizing muscles-specific effects of sex and menstrual cycle. *Int J Sports Med.* (2021) 42:270–6. doi: 10.1055/a-1236-3654

14. Khowailed IA, Lee Y, Lee H. Assessing the differences in muscle stiffness measured with shear wave elastography and myotonometer during the menstrual cycle in young women. *Clin Physiol Funct Imaging.* (2022) 42:320–6. doi: 10.1111/cpf.12763

15. Zinder SM, Padua DA. Reliability, validity, and precision of a handheld myometer for assessing in vivo muscle stiffness. *J Sport Rehabil.* (2011) 20(3):2010 0051. doi: 10.1123/jsr.2010-0051

16. Schneider S, Peipsi A, Stokes M, Knicker A, Abeln V. Feasibility of monitoring muscle health in microgravity environments using myoton technology. *Med Biol Eng Comput.* (2015) 53:57–66. doi: 10.1007/s11517-014-1211-5

17. Maruyama S, Sekine C, Shagawa M, Yokota H, Hirabayashi R, Togashi R, et al. Menstrual cycle changes joint laxity in females-differences between eumenorrhea and oligomenorrhea. *J Clin Med.* (2022) 11(11):3222. doi: 10.3390/jcm11113222

18. Shagawa M, Maruyama S, Sekine C, Yokota H, Hirabayashi R, Hirata A, et al. Comparison of anterior knee laxity, stiffness, genu recurvatum, and general joint laxity in the late follicular phase and the ovulatory phase of the menstrual cycle. *BMC Musculoskelet Disord.* (2021) 22:886. doi: 10.1186/s12891-021-04767-8

19. Edama M, Ohya T, Maruyama S, Shagawa M, Sekine C, Hirabayashi R, et al. Relationship between changes in foot arch and sex differences during the menstrual cycle. *Int J Environ Res Public Health.* (2022) 20(1):509. doi: 10.3390/ijerph20010509

20. Shagawa M, Maruyama S, Sekine C, Yokota H, Hirabayashi R, Togashi R, et al. Knee laxity in the menstrual cycle after anterior cruciate ligament reconstruction: a case series. *Int J Environ Res Public Health.* (2023) 20:2277. doi: 10.3390/ijerph20032277

21. Rasmussen OS. Sonography of tendons. *Scand J Med Sci Sports*. (2000) 10:360–4. doi: 10.1034/j.1600-0838.2000.010006360.x

22. Caresio C, Salvi M, Molinari F, Meiburger KM, Minetto MA. Fully automated muscle ultrasound analysis (MUSA): robust and accurate muscle thickness measurement. *Ultrasound Med Biol*. (2017) 43:195–205. doi: 10.1016/j.ultrasmedbio.2016.08.032

23. Chen G, Wu J, Chen G, Lu Y, Ren W, Xu W, et al. Reliability of a portable device for quantifying tone and stiffness of quadriceps femoris and patellar tendon at different knee flexion angles. *PLoS One*. (2019) 14(7):e0220521. doi: 10.1371/journal.pone.0220521

24. Landis JR, Koch GG. The measurement of observer agreement for categorical data. *Biometrics*. (1977) 33:159–74. doi: 10.2307/2529310

25. Cohen J. *Statistical Power Analysis for the Behavioral Sciences*. Mahwah, NJ: Lawrence Erlbaum Associates (1998).

26. Maruyama S, Yamazaki T, Sato Y, Suzuki Y, Shimizu S, Ikezu M, et al. Relationship between anterior knee laxity and general joint laxity during the menstrual cycle. *Orthop J Sports Med*. (2021) 9:2325967121993045. doi: 10.1177/2325967121993045

27. Petrofsky J, Lee H. Greater reduction of balance as a result of increased plantar fascia elasticity at ovulation during the menstrual cycle. *Tohoku J Exp Med*. (2015) 237:219–26. doi: 10.1620/tjem.237.219

28. Miller BF, Olesen JL, Hansen M, Døssing S, Crameri RM, Welling RJ, et al. Coordinated collagen and muscle protein synthesis in human patella tendon and quadriceps muscle after exercise. *J Physiol*. (2005) 567:1021–33. doi: 10.1113/jphysiol.2005.093690

29. Sarwar R, Niclos BB, Rutherford OM. Changes in muscle strength, relaxation rate and fatiguability during the human menstrual cycle. *J Physiol*. (1996) 493(Pt 1):267–72. doi: 10.1113/jphysiol.1996.sp021381

30. Janse De Jonge XA, Boot CR, Thom JM, Ruell PA, Thompson MW. The influence of menstrual cycle phase on skeletal muscle contractile characteristics in humans. *J Physiol*. (2001) 530:161–6. doi: 10.1111/j.1469-7793.2001.0161m.x

31. Fridén C, Hirschberg AL, Saartok T. Muscle strength and endurance do not significantly vary across 3 phases of the menstrual cycle in moderately active premenopausal women. *Clin J Sport Med*. (2003) 13:238–41. doi: 10.1097/00042752-200307000-00007

frontiers | Frontiers in Sports and Active Living

TYPE Original Research
PUBLISHED 25 April 2024
DOI 10.3389/fspor.2024.1380864

Characterization of the vastus lateralis torque-length, and knee extensors torque-velocity and power-velocity relationships in people with Parkinson's disease

Riccardo Magris*, Francesca Nardello, Federica Bombieri, Andrea Monte and Paola Zamparo

Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy

Check for updates

OPEN ACCESS

EDITED BY
Maria-Elissavet Nikolaidou,
National and Kapodistrian University of Athens, Greece

REVIEWED BY
Marco Romanato,
INSERM U1127 Institut du Cerveau et de la Moelle épinière (ICM), France
Adamantios Arampatzis,
Humboldt University of Berlin, Germany
Zhenya Smirnov,
Humboldt University of Berlin, Germany, in collaboration with reviewer [AA]

*CORRESPONDENCE
Riccardo Magris
✉ riccardo.magris@univr.it

RECEIVED 02 February 2024
ACCEPTED 10 April 2024
PUBLISHED 25 April 2024

CITATION
Magris R, Nardello F, Bombieri F, Monte A and Zamparo P (2024) Characterization of the vastus lateralis torque-length, and knee extensors torque-velocity and power-velocity relationships in people with Parkinson's disease.
Front. Sports Act. Living 6:1380864.
doi: 10.3389/fspor.2024.1380864

Introduction: Parkinson's disease (PD) is a prevalent neurodegenerative condition observed primarily in the elderly population that gives rise to motor and non-motor symptoms, one of which is muscle weakness. The aim of this study was to characterize the vastus lateralis torque-fascicle length (T-L) and the knee extensors torque-angular velocity (T-V) and power-angular velocity (P-V) relationships in PD patients and to investigate the influence of muscle geometry on muscle mechanics.
Methods: Participants (11 PD: patients, 9 CR: age matched healthy controls; 10 CY: young healthy controls) performed: (i) isometric contractions (e.g., MVC) to obtain the torque-angle and T-L relationships; (ii) isokinetic (e.g., iso-velocity) contractions to obtain the T-V and P-V relationships. During the experiments, the architecture of vastus lateralis (pennation angle, fascicle length, muscle thickness) was also determined by using an ultrasound apparatus.
Results: Significant differences were observed between PD patients and physically matched control groups (CR and CY) in terms of maximum isometric force (calculated as the apex of the T-L curve) and maximum mechanical power (apex of the P-V curve), but not in maximum shortening velocity. Among the mechanical variables investigated, mechanical power was able to identify differences between the less and the more affected side in PD patients, suggesting that this parameter could be useful for clinical evaluation in this population.
Conclusions: The observed results cannot be explained by differences in muscle geometry at rest (similar in the three cohorts), but rather by the muscle capacity to change in shape during contraction, that is impaired in PD patients.

KEYWORDS
force-velocity relationship, muscle disorders, muscle mechanics, muscle architecture, mechanical power

Introduction

The evaluation of the muscle's mechanical output by means of the torque-length (T-L), torque-angular velocity (T-V) and power-angular velocity (P-V) relationships has gained increasing attention in recent years due to their association with physical performance in populations ranging from young people (and athletes) to elderly people with or without pathological conditions (1, 2).

The T-V relationship illustrates the ability to produce torque at different contraction velocities and can be used to identify muscle weakness at different/specific torque or velocity levels; this allows for a better characterization of a subject's functional capacity during daily life activities (3) compared to standard isometric evaluations: the higher the torque produced at a given movement velocity, the higher the mechanical power generated. Mechanical power is negatively and independently associated with the risk of cognitive decline (4), mobility limitations, disability (5) and hospitalisation (6) in older adults. Therefore, the evaluation of the P-V relationship is also of scientific interest to better identify the loss of function in several pathologies. Last but not least, the T-L curve could also provide important information about people's physical fitness since it describes a muscle's capability to generate torque at any given fascicle length. Together, these three relationships could provide important insights into a specific population's muscle's mechanical capacity.

To our knowledge, a population that has never been characterised in terms of T-L, T-V and P-V relationships is that of subjects with Parkinson's disease (PD). Parkinson's disease is a prevalent neurodegenerative condition observed primarily in the elderly population, affecting approximately 1% of individuals aged 60 years and above (7). This pathological state gives rise to motor and non-motor symptoms, including bradykinesia, rigidity, tremor, and postural instability. These symptoms collectively contribute to a progressive deterioration of functional abilities, ultimately leading to disability. The decrease in functional performance observed in patients with Parkinson's disease (PD) can be attributed to various factors, one of which is muscle weakness (8). This weakness is linked to both central and peripheral alterations, such as a decrease in voluntary activation (9), altered spatial activation pattern (10), diminished fascicle shortening and tendon elongation (11), increased muscle stiffness (12) and decreased muscle's capability to change in shape (13). Although the effects of the central parameters in determining the loss of force in people with PD are well acknowledged in the literature [e.g., (14, 15)], the impact of peripheral variables remains unclear.

In this regard, from a neural/nervous point of view, a lower EMG amplitude during maximum explosive contractions was observed in PD patients compared to healthy age-matched control participants (13, 15). Furthermore, PD patients exhibit higher antagonist co-activation (16) and variable motor unit discharge rates (17) during maximal and submaximal isometric force reduction compared to healthy subjects, which in turn decrease force production capacity (18). From a peripheral point of view, alterations in muscle architecture could (at least partially) explain the observed loss of force (and hence of torque). For example, changes in muscle thickness and/or physiological cross-sectional area are related to a loss of maximum force in elderly people (19). On the other hand, decreases in pennation angle and fascicle length could be related to a decrease in muscle force and fascicle shortening velocity, respectively (20). These geometrical modifications could be associated to changes in the T-L, T-V and P-V relationships. For example, a decrease in fascicle length is related to a reduction in the maximum shortening velocity and a decrease in the fascicle operating length along its T-L relationship (21, 22).

Resting muscle geometry was observed to be similar between PD patients and age and physically-matched control groups (13, 15) whereas fascicle and muscle behaviour were found to differ in dynamic conditions (e.g., during explosive contractions; (13). Indeed, PD patients showed higher values of MTU stiffness during contraction compared to healthy controls reducing the muscle's capability to change in shape and to increase force production rapidly (13). Therefore, the concomitant evaluation of muscle geometry at rest and during contraction, as well as that of the torque-length and power-velocity relationships, could represent an important screening for people with PD, providing information about the level/progression of this pathology.

Therefore, this study's primary aim was to evaluate the possible changes in muscle geometry at rest and during contraction and to investigate how these changes could affect the T-L, T-V and P-V relationships in PD patients. Furthermore, we compared data obtained in this specific population with those obtained in age and physically-matched cohorts and a young control group to better distinguish the changes imposed by PD from those typically related to the ageing process.

We hypothesized no significant differences in resting muscle geometry between PD patients and the age and physically-matched group, as previously reported by Martignon et al. (15) and Monte et al. (13). However, significant differences between PD patients and control groups were expected in terms of architectural behavior during the characterization of the T-L, T-V, and P-V relationships (as previously found during explosive contractions by (13). We then expected that the differences in muscle fascicle behaviour exhibited by PD patients could affect these relationships (e.g., lower torque, angular velocity and power in PD patients). Finally, since the onset of pathological symptoms in PD exhibits asymmetry (resulting in a higher degree of impairment of one limb) we hypothesised that the torque, angular velocity and power were reduced in the more affected limb compared to the less affected one.

Materials and methods

Participants

In this study we calculated the sample size *a-priori* with an effect size of 0.55 (e.g., obtained from the literature based on the differences in maximum voluntary torque between PD patients and healthy controls (15);, an alfa-level of 0.05, a statistical power of 0.8 and three groups of subjects (PD, CR and CY). The total sample size was 36 (12 participants for each group). For this study, we recruited 12 patients with mild to moderate Parkinson's disease (PD), 16 healthy age-matched controls (CR), and 14 healthy young controls (CY).

For PD patients, the inclusion criteria were a diagnosis of idiopathic PD performed by a neurologist, according to the London Brain Bank guidelines. All subjects were characterized by postural instability and a gait disorders phenotype (PIDG). The disease severity was calculated based on the modified Hoehn and Yahr scale, and only subjects up to stage 3 were recruited. All procedures were conducted during the medication "on" condition of the dopaminergic treatment. For all participants (PD, CR and CY), exclusion criteria were any type of dementia, inability to walk and muscular injuries. Subjects with cardiovascular, orthopaedic, or metabolic disorders were also excluded from the cohorts.

All participants received written and oral information and instructions before the study and gave their written informed consent to the experimental procedure. The experimental protocol was approved by the Ethical Committee of the University of Verona (protocol number: 2021-UNVRCLE-0450152) and was performed in accordance with the Helsinki Declaration.

Experimental design

Healthy subjects (CR and CY) participated in a single session during which only their dominant lower limb was tested whereas PD patients performed two sessions during which the more affected limb (PDA) and the less affected limb (PDNA) (as ranked by their neurologist) were both tested.

The entire protocol involved: (i) isometric contractions (e.g., maximum voluntary fixed-end contractions, MVC) to obtain the torque-angle and torque-fascicle length relationships; (ii) isokinetic (e.g., iso-velocity) contractions to obtain the torque-angular velocity and power-angular velocity relationships.

During the experiments, the vastus lateralis (VL) fascicles length was determined by using an ultrasound apparatus.

Data collection

T-Angle, T-L, T-V and P-V relationships

The participants were secured on a dynamometer (100 Hz; Cybex NORM, Computer Sports Medicine Inc., Stoughton, USA) using a trunk and pelvic strap. They were seated with their hip fixed at 85° of flexion (23) with the arms crossed in front of the chest. The participant's lower leg was fixed with Velcro straps around the mid-shank to a cushioned attachment connected to the dynamometer's lever arm. The rotational axis of the dynamometer was carefully aligned with the axis of rotation of the knee joint during a maximal contraction at 60° knee flexion (24). A standardized warm-up for each contraction type was conducted to familiarise the subjects with the task (see Supplementary Figure S1). Firstly, to account for the effects of gravity and passive joint torque on the net knee joint torque, three passive knee extensions were performed over the knee joint's range of motion, while participants were instructed to relax (25). Secondly, the subjects were instructed to push "as hard as possible" for 3–4 s (for the MVCs) or over the entire ROM (i.e., from 90° to 0°) (for the iso-velocity trials).

Participants performed three consecutive MVCs (knee extensions) at six knee joint angles: 15, 30, 45, 60, 75 and 90° (where 0° = knee fully extended) and three consecutive iso-velocity knee extensions at five angular speeds: 45, 90, 150, 210 and 250°/s. For each iso-velocity trial, participants were instructed to extend their knee as hard as possible starting ~0.5 s before the knee extension. This protocol pre-loaded the quadriceps muscles, facilitating maximal voluntary neuromuscular activation and maximum torque throughout the entire ROM (26, 27).

Between contractions, at least 2 min of rest were provided to minimize fatigue (28). A randomized order has been followed and the participants received strong verbal encouragement during each MVCs and each iso-velocity knee extension.

A 2D video analysis was utilized to verify the actual knee angle during all contractions based on the following markers: greater trochanter of the opposite side, the lower portion of the patella (patellar tendon origin), upper anterior surface of the tibia (patellar tendon insertion), knee center of rotation and medial malleolus. The marker positions were recorded by means of a Casio Exilim Camera (100 Hz, Casio Computer Co., Ltd., Tokyo, Japan) and analysed with a video processing software (Tracker v6.0.10). The camera was positioned on the opposite side of the subject, perpendicular to the thigh longitudinal axis.

During the MVCs, muscle architecture behaviour was recorded by means of a B-mode ultrasound apparatus (Telemed MicrUs EXT-1H rev.D, Lithuania) with a 6 cm linear array probe (with a depth and width of 60 and 60 mm, respectively) and a sampling frequency of 12 Hz. The probe was fixed to the skin by means of a plastic strap approximately at 50% of the femoral length, aligned on the muscle belly and corrected with respect to the superficial and deep aponeurosis to have a clear image of the perimysial connective intramuscular tissue, that it is indicative of the muscle fascicle structure. The probe was never removed during the entire experimental session. The muscle architecture parameters were calculated from the ultrasound videos (see Data Analysis below). All devices were synchronised with an external manual trigger (5 V) and data were collected by means of LabChart (V.6).

Questionnaires

Physical activity level was investigated by means of the International Physical Activity Questionnaire (29), while the cognitive function was assessed using the Mini Mental State Examination (MMSE) (30). The disease severity was classified according to the Hoehn and Yahr scale (31), while the assessment of the degree of motor and functional impairment was determined using part III of the Unified Parkinson's Disease Rating Scale (UPDRS) (32).

Data analysis

The offline analysis was conducted using custom-developed programs in Matlab (r2021a).

T-Angle, T-L, T-V and P-V relationships

The total torque generated by the knee extensors was corrected for the gravitational torque effect (determined during the passive joint rotation driven by the dynamometer) according to Kellis & Baltzopoulos (33), and finally normalized to the mass of the subjects (e.g., expressed in Nm/kg).

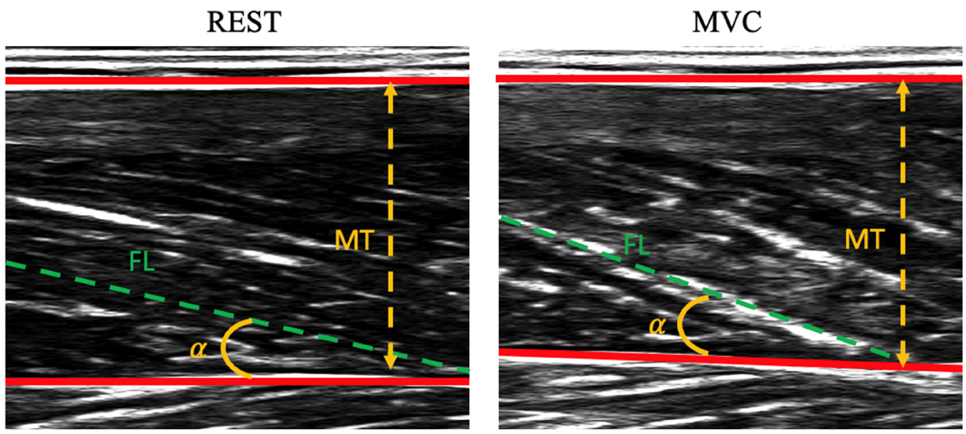

FIGURE 1
Representative ultrasound images of vastus lateralis at rest (panel on the left) and during contraction (panel on the right). Data refer to a patient with Parkinson's disease. MT, muscle thickness; α, pennation angle (PA); FL, fascicle length.

Ultrasound measurements during contraction were performed during the MVCs: on each ultrasound image, two points were manually digitalized on the superficial aponeurosis, two on the deep aponeurosis and two on a muscle fascicle through a Matlab custom script. The length of the muscle fascicles (FL) was defined as the distance along the fascicles between the deep and superficial aponeuroses; pennation angle (PA) was defined as the angle between the collagenous tissue and the deep aponeurosis; muscle thickness (MT) was defined as the perpendicular distance from the deep to the superficial aponeurosis in the middle of the field of view (34, 35) (see Figure 1). Since the VL fascicles could be longer than the field of view of the ultrasound probe, a linear extrapolation of the fascicle length was performed, as previously proposed in the literature (e.g., (22, 36–38). Representative ultrasound measurements were taken as a mean of three digitalized images. Changes in fascicle length (ΔFL), muscle thickness (ΔMT), and pennation angle (ΔPA) were calculated as the differences between the values determined during contraction and in the rest condition.

The values of torque at plateau (registered in the three MVCs at each knee joint angle), the corresponding values of fascicle length (FL at torque plateau) and the knee joint angles (obtained from the video analysis) were averaged and used to determine the torque-angle and T-L relationships. These relationships were then fitted with second-order polynomial functions based on which we calculated: (i) the maximum voluntary torque (T_{max}) as the apex of the T-L relationship, (ii) the optimal knee joint angle as the angle at which T_{max} occurs and (iii) the optimal fascicle length (L_{opt}) as the fascicle length at which T_{max} occurs.

Regarding the T-V relationship, the values of (peak) torque were calculated within the iso-velocity phase (during each isokinetic contraction); the acceleration and deceleration phases were thus excluded from data analysis [e.g., (39, 40)] (see Supplementary Figure S2).

Peak torque values were averaged among trials at the same angular velocity and used to reconstruct the T-V relationship (the values of angular velocity being determined by the Cybex software). Maximal angular velocity was determined as the intercept value on the abscissa of the T-V curve (fitted with a linear relationship).

Finally, the mechanical power values were calculated as the product between torque and angular velocity. The P-V curve was obtained by fitting the power and angular velocity data with a second-order polynomial function (41). Maximum mechanical power was calculated as the apex of the P-angular velocity curve. Power data are reported normalized for body mass (e.g., expressed in W/kg).

Statistics

Data normality was assessed using a Shapiro-Wilks test. A one-way ANOVA was used to investigate differences in physical activity levels and anthropometric characteristics.

To accentuate the effect of the pathology, a one-way ANOVA was employed to delineate significant differences among the more affected limb of PD patients, and the CR and CY group in optimal knee angle, L_{opt}, T_{max}, V_{max} and P_{max}. A two-way ANOVA was used to compare the groups cited above at the different knee angular positions or angular velocities to evaluate possible differences in specific portions of the T - L, T -V and P -V relationships. A *post hoc* Tukey test (with Bonferroni correction) was used. A paired *t*-test was then utilized to assess differences between PD patients' more affected and less affected limb.

Correlations between variables were calculated according to the Pearson correlation coefficient. In case the normality of the data was not verified, a Kruskal-Wallis test was performed and Spearman correlation coefficients were determined instead. Complete data are reported for 11 PD patients, 9 elderly controls and 10 young controls; due to the revised sample size, the α level was adjusted, as suggested by Gómez-de-Mariscal et al. (42). *A-posteriori* sample size analysis with the actual sample size (30 participants) indicated a *p*-value of 0.09; in order to reduce the Type1 error, the *p*-value was finally set as 0.08.

TABLE 1 Anthropometric characteristics, level of physical activity (IPAQ), cognitive status (MMSE), UPDRS-motor scale score, and H&Y score.

	Age (years)	Body mass (kg)	Stature (cm)	IPAQ (MET/week)	MMSE score	UPDRS score	H&Y score	Disease duration (years)
CY (n = 10; 4F/6M)	25.1 ± 1.73*	74.9 ± 14.9	176.2 ± 10.2	2,978 ± 1,657	29.4 ± 0.52	–	–	–
CR (n = 9; 1F/8M)	68.3 ± 5.36	78.7 ± 10.1	173.4 ± 8.10	3,916 ± 1,856	28.6 ± 1.67	–	–	–
PD (n = 11; 2F/9M)	70.5 ± 5.28	78.4 ± 12.5	172.5 ± 8.18	3,493 ± 1,230	28.6 ± 1.43	27.3 ± 13.2	1.50 ± 0.65	6.1 ± 4.36

CY: young subjects; CR: elderly control subjects; PD: subjects with Parkinson's disease.
IPAQ, International Physical Activity Questionnaire; MMSE, mini mental state examination; UPDRS, Unified Parkinson's Disease Rating Scale; H&Y, Hoehn and Yahr scale.
*p < 0.001 between CR and CY.

Statistical analysis was performed using Jamovi (v2.4.11) and SPSS 23 (IBM Corp., Armonk, NY, USA).

Results

One PD patient, 7 elderly controls and 4 young controls were excluded from the final cohorts since they were not able to complete the experimental protocol or to modulate force in either the isometric and/or isokinetic trials; in these cases, the R^2 of the T-L or T-V relationship was rather low (ranging from 0.40 to 0.57) whereas R^2 was higher than 0.85 in all other cases/participants. Complete data are, thus, reported for 11 PD patients, 9 elderly controls and 10 young controls. As shown in Table 1, no differences were observed among groups regarding body mass and stature, as well as in physical activity level and cognitive function; according to the IPAQ scores, all participants were physically active. In patients, the UPDRS score and the H&Y score indicate a mild to moderate stage of the pathology (Table 2).

Results are presented by considering the effect of ageing (CY vs. CR), pathology (CR vs. PDA) and the intra-subject effect of pathology (PDNA vs. PDA). In tables, data are reported as mean and standard deviation; in figures, data are reported as mean and standard error.

T-Angle, T-L, T-V and P-V relationships

In Figure 2 the torque-fascicle length and the torque-angle relationships for the three groups are reported. T_{max} was higher in CY compared to CR (p = 0.037). Significant differences were also observed between CY and PDA (P < 0.001) and CR and PDA (P = 0.031). No significant effect of ageing or pathology was observed in optimal knee joint angle (CY vs. CR: p = 0.789; CR vs. PDA: p = 0.703) or optimal fascicle length (CY vs. CR: p = 0.879; CR vs. PDA: p = 0.632).

No significant differences were observed between PDA and PDNA in optimal fascicle length (p = 0.399), optimal knee joint angle (p = 0.767) and T_{max} (p = 0.153).

In Figure 3 the T-V and P-V relationships for the three groups are reported. No significant effect of ageing or pathology in maximum angular velocity (V_{max}) among groups (CY vs. CR = 0.845; CR vs. PDA = 0.153) was observed. Significant differences were observed in P_{max}: CY showed higher values of P_{max} compared to CR (p = 0.05) and PDA (p < 0.001), while CR exhibited higher values of P_{max} compared to PDA (P = 0.023).

No significant differences were observed between PDA and PDNA in V_{max} (p = 0.485), but a significant difference was observed between PDA and PDNA in P_{max} (P = 0.064).

Ultrasound measurement

Data of fascicle length, pennation angle and muscle thickness at rest, and their changes from force onset till plateau during the MVCs, are reported in Figure 4.

Muscle thickness, pennation angle and fascicle length at rest showed no significant differences among groups. Changes in muscle thickness showed no significant differences between CY and CR (p = 0.285), while a significant effect of pathology was

TABLE 2 Characteristics of subjects with Parkinson's disease.

Gender (M/F)	Age (years)	Body mass (kg)	Stature (cm)	Disease duration (years)	H&Y score	UPDRS score	MMSE score	Medication	Side of appearance of the first motor symptom
M	77	90	176	5	1	36	29	Madopar, Sinemed	Left
M	70	82	165	5	1	14	30	Madopar	Right
M	75	76	183	15	1	22	30	Mirapixin, Sinemed, Oxyen, Sirio	Right
M	65	67.5	178	2	1.5	31.5	27	Sinemed, Rantal	Right
M	73	72	180	7	1.5	20.5	30	Sinemed	Left
M	65	85	179	2	0.5	9	28	Madopar, Sinemed	Right
M	80	62	168	12	2	46	29	Mirapixin, Sinemed	Left
F	68	60	156	6	3	52	29	Sinemed, Mirapixin, Sirio	Left
F	66	78	165	0.2	1	21	30	None	Left
M	72	100	176	5	2	28	27	Sinemed, Ropinirol	Left
M	65	90	172	8	1.5	19	26	Sinemed, Requir	Right

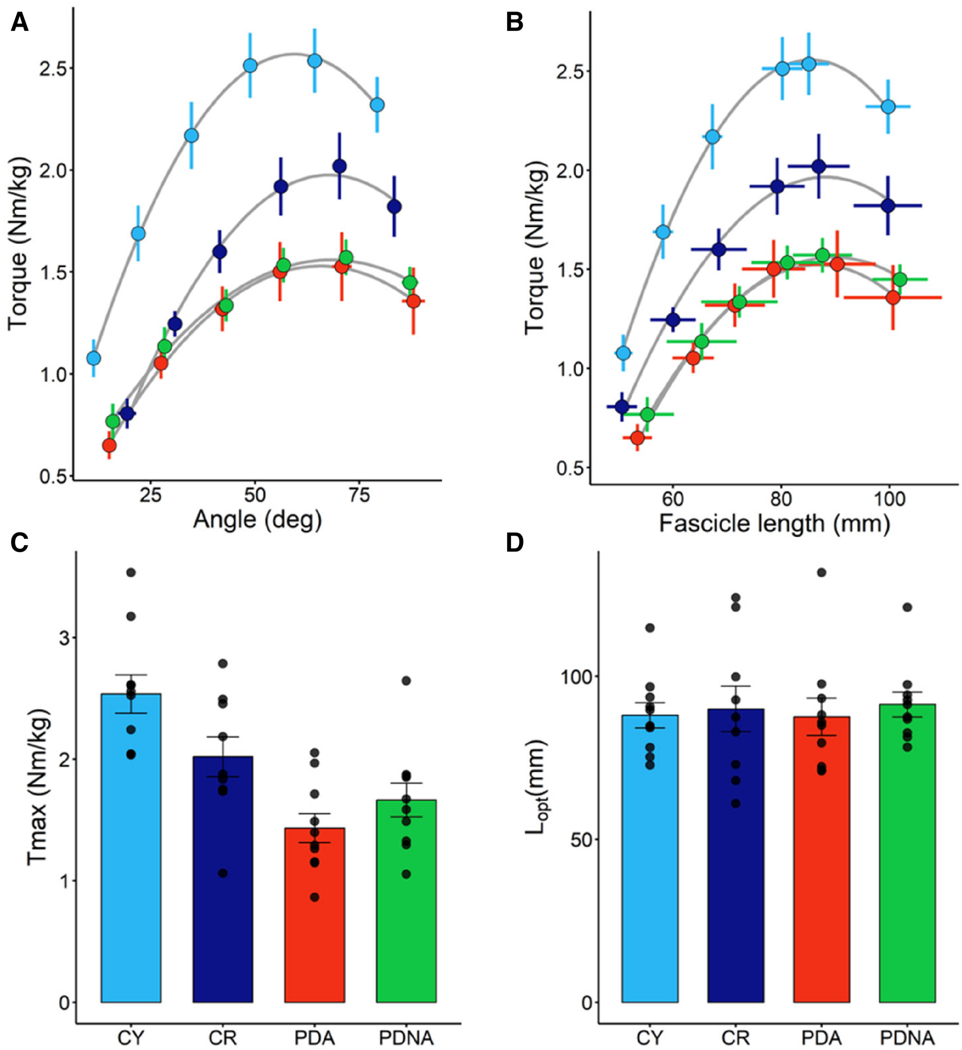

FIGURE 2
Torque-angle (panel A) and torque-fascicle length (panel B) relationships (fascicle data refer to VL); data points are means ± SE. The optimal length of the VL fascicles (L_{opt}) and maximal torque (T_{max}), as calculated from the torque-fascicle length relationship, are reported in panels D and C, respectively; data points represent individual data (means ± SE are represented as well). Light blue and dark blue data refer to control groups (young and middle-aged, respectively: CY and CR); red and green data refer to PD patients (more affected and less affected limb, respectively: PDA and PDNA). No differences were observed in L_{opt} among groups. For T_{max}: CY vs. CR = 0.037; CY vs. PDA <0.001; CR vs. PDA = 0.031.

observed (CR vs. PDA: $p = 0.022$); the smallest changes were observed in PDA. No differences between PDA and PDNA were appreciated ($p = 0.508$).

Pennation angle was not affected by ageing (CY vs. CR: $p = 0.110$) or pathology (CR vs. PDA: $p = 0.431$) and no differences between PDA and PDNA were appreciated ($p = 0.901$).

Significant differences were observed in fascicle length as a function of age (CY vs. CR: $p = 0.025$) and pathology (CR vs. PDA: $p = 0.069$), as well as between PDA and PDNA ($p = 0.077$); the smallest changes were observed in PDA.

Correlations between parameters

No significant correlations were observed between architectural parameters (at rest or changes) and T_{max}, V_{max} or P_{max}.

Moreover, no significant correlations were appreciated between each of the investigated parameters and the UPDRS and H&Y scores.

Discussion

This study aimed to determine the effects of Parkinson's disease on muscle's mechanical output (T-L, T-V and P-V relationships) and its determinants (muscle geometry). For this reason, we recruited three groups of participants, matched for physical activity, to better distinguish the effect of ageing from the impairment imposed by the pathology itself. We observed an effect of age and pathology on maximal isometric force and maximal mechanical power: elderly participants had lower values of force and power compared to their younger counterparts, and Parkinson's disease exacerbated the loss of function (by

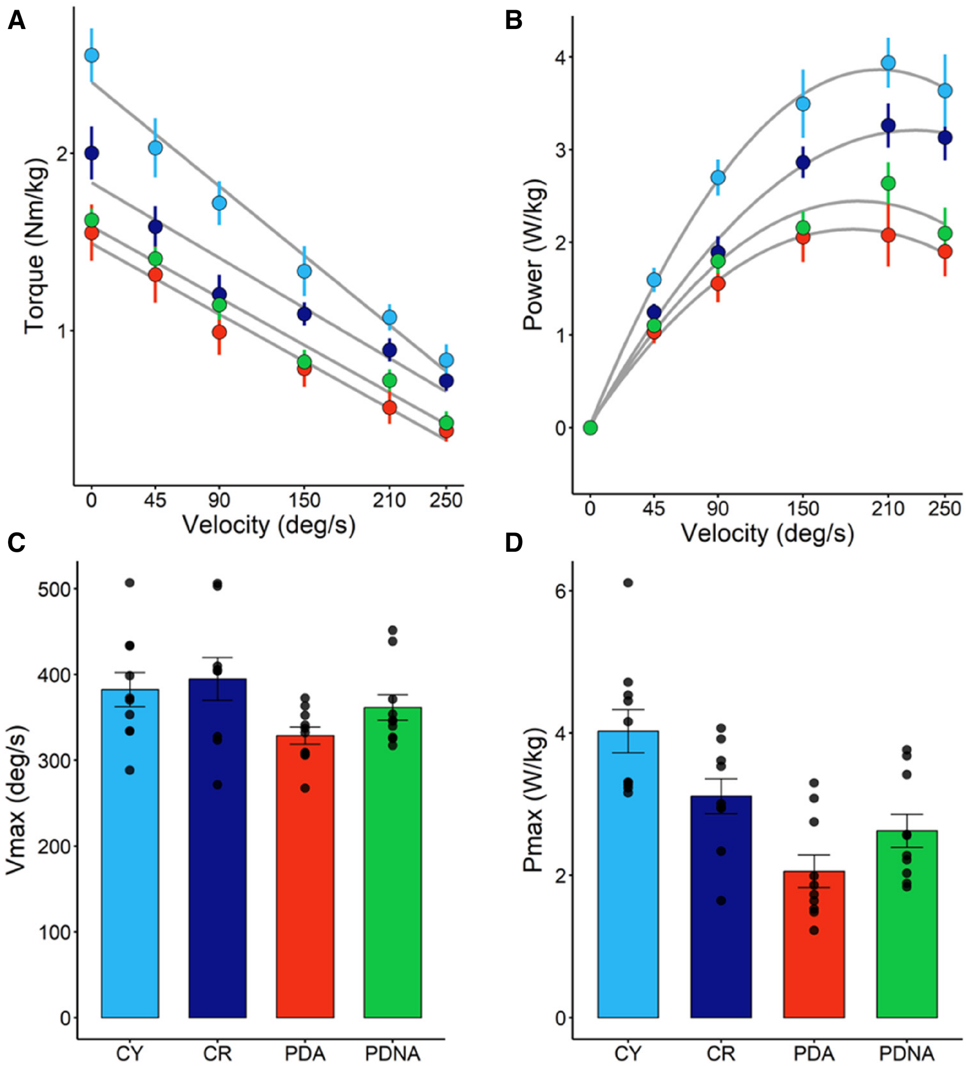

FIGURE 3
Torque-angular velocity (panel **A**) and power-angular velocity (panel **B**) relationships; data points are means ± SE. Maximal angular speed (V_{max}) and maximal power output (P_{max}), as calculated from the T-V and P-V relationships, are reported in panels **C** and **D**, respectively; data points represent individual data (means ± SE are represented as well). Light blue and dark blue data refer to control groups (young and middle-aged, respectively: CY and CR); red and green data refer to PD patients (more affected and less affected limb, respectively: PDA and PDNA). No differences were observed in V_{max} among groups. For P_{max}: CY vs. CR = 0.05; CY vs. PDA <0.001; CR vs. PDA = 0.023; PDA vs. PDNA = 0.061.

comparing PD patients with healthy subjects of the same age). In addition, we observed differences in maximum mechanical power between the more and the less-affected side in PD patients. Last but not least, while muscle geometry at rest was similar among groups, significant differences were observed when investigating dynamic muscle shape changes in these populations.

Muscle weakness is one of the more critical factors that characterise functional disability in elderly people as well as in Parkinson's patients. Indeed, a decrease of muscle strength as a function of age is well established in the literature (see (43, 44). Our data support these findings, showing a downward shift of the T-L and T-V curves in elderly subjects compared to the young control group. We also observed a further reduction in T_{max} in people with Parkinson's disease, suggesting that maximum isometric torque, calculated as the apex of the T-L curve, could be a useful indicator of functional disability in this population.

Maintaining the ability to generate (and control) high-velocity movements is quite challenging in the elderly and, more so, in PD patients. The difficulty in generating high-speed movements could lead to functional disability and could affect the patient's quality of life (e.g., increasing the risk of falls) (45). Our data showed no significant differences among groups in V_{max} (maximum knee joint angular velocity); since muscle fascicle shortening velocity and knee angular velocity are likely correlated (46, 47), these data suggest that maximum contraction speed is preserved (in our experimental cohorts) regardless of age and pathology. Other studies showed a significant reduction in V_{max} as a function of age, but with differences that became evident after the 7th decade (2, 43). For

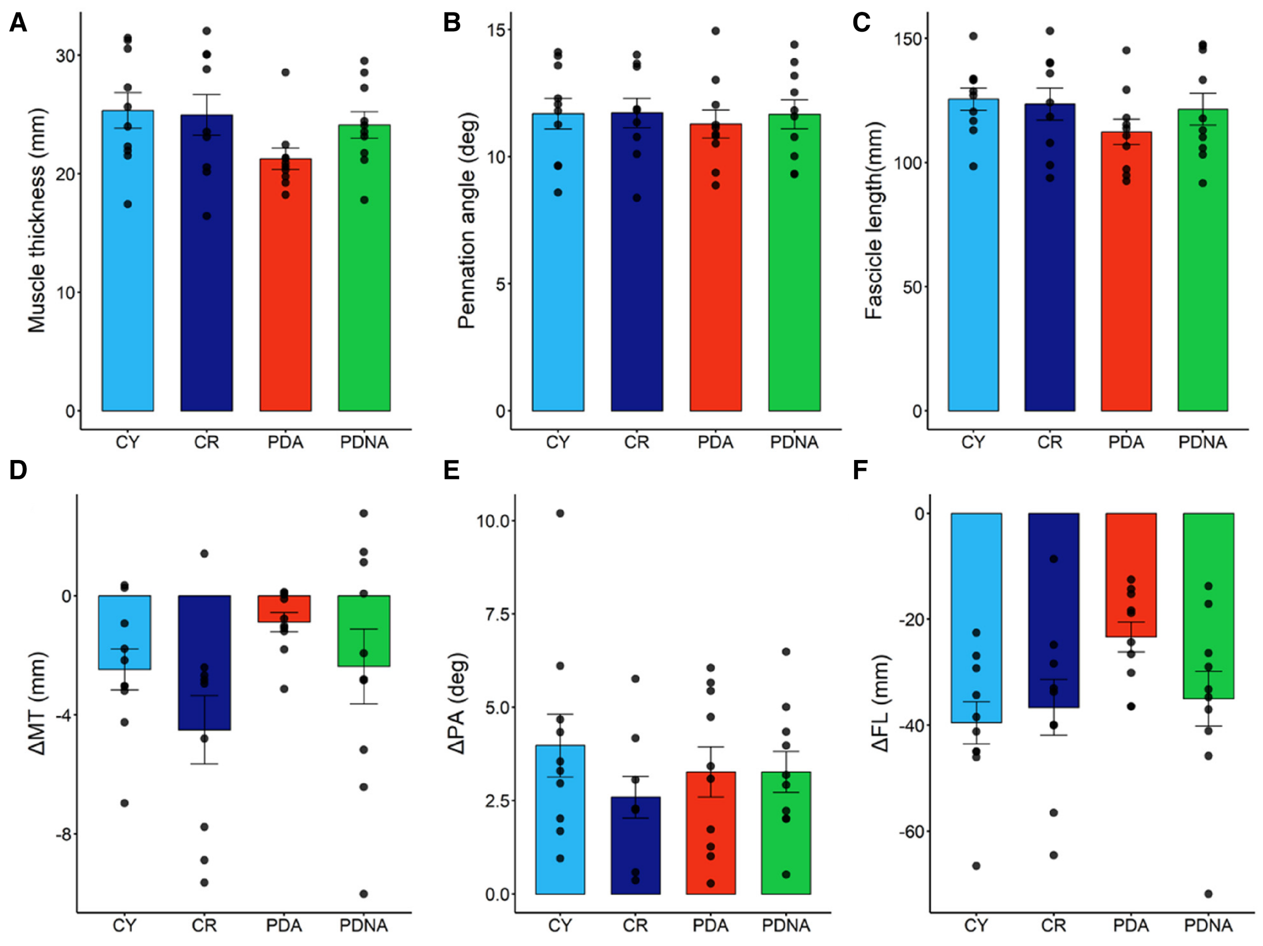

FIGURE 4
Fascicle length, muscle thickness, and pennation angle at rest (panels A–C, respectively). Dynamic changes (contraction-rest) are reported in panels D–F, respectively: fascicles shorten, muscle thickness decreases, and pennation angle increases. Data points represent individual data (means ± SE are represented as well). Light blue and dark blue data refer to control groups (young and middle-aged, respectively: CY and CR); red and green data refer to PD patients (more affected and less affected limb, respectively: PDA and PDNA). No differences were observed in ΔPA among groups. For ΔMT: CR vs. PDA = 0.022; for ΔFL: CY vs. CR = 0.025; CR vs. PDA = 0.069; PDA vs. PDNA = 0.077.

example, Alcazar and colleagues (43) observed a non-significant difference in V_{max} (angular velocity) of the quadriceps muscles between young and middle-aged (40–60 years) subjects. The CR and PD participants of this study had an average age of 71 and 68 years, respectively, and, more importantly, were matched for physical activity, that could counteract the loss of function that occurs with age. To our knowledge, this is the first study that investigates the effects of PD on V_{max} and our data indicate that an active lifestyle could (at least) partially counteract the loss of V_{max} in this population.

Muscle power is one of the most important indices of functional capacity of an individual (5). In accordance with the literature, we observed an age-related decline in P_{max} (43, 48), which essentially reflects the loss in T_{max}. Furthermore, our data indicate a further decrease in P_{max} in PD patients, suggesting that this variable could be another important indicator (along with T_{max}) for investigating the functional capacity of this population. In this regard, it is important to note that P_{max} was the only variable able to identify differences between the less and the more-affected side in PD patients; the small differences between PDA and PDNA in V_{max} and T_{max} "sum up" and became significant when power, and hence P_{max}, is calculated.

Optimal knee joint angle and optimal fascicle length were unaffected by age and pathology: the T-L and T-A curves showed a down-shift without any significant change in the fascicle's operating range. These findings could be primarily attributed to the similar values of muscle geometry in our groups at rest. Indeed, fascicle length and optimal knee joint angle depend on the operating length (49): increasing fiber length increases the absolute muscle active operating length as well as the knee joint range of motion. Moreover, a longer fascicle length is also related to higher values of maximum shortening velocity (50).

Previous studies did not show any difference in architectural parameters at rest between PD patients and age and physically-matched controls [e.g., (13, 15)], and this was also observed in the present study, reinforcing the idea that an active lifestyle is a key factor in preserving muscle structure in this population. However, significant differences were observed between PD patients and the other groups in terms of dynamic muscle changes (from rest to contraction). Indeed, the smaller changes in thickness and fascicle length (from rest to contraction) were observed in the vastus lateralis of PD patients (affected side). Similar results were reported by Monte et al. (13) during explosive contractions, where the authors indicate that reduced muscle's capability to change in shape in PD patients is associated with an increase in their muscle-tendon unit stiffness. The differences in muscle fascicle behavior observed in this study between PD patients and the other cohorts could, thus, be related to hypertonia of the musculoskeletal system (12) (or to other mechanical factors that could not be appreciated with a simple geometrical analysis at rest). Hence, these data underline, again (13), that the analysis of muscle and fascicle behaviour during contraction could provide further insight into the mechanical alteration imposed by PD, compared to standard imaging techniques at rest.

Limitations and further considerations

The present study assessed knee extension muscle function, so these findings apply to this muscle action only. Evaluation of other muscles, such as the hip and plantar flexor muscles, cannot be performed with sufficient methodological rigor in the elderly/patients (e.g., for the difficulty of remaining in a prone lying position). Hence, even if other lower-limb muscles are more compromised in PD (51, 52), testing the knee extensors represents the best solution to evaluate muscle's mechanical alterations in these patients.

In this study, we did not investigate the co-activation of the antagonist muscles, but muscle co-activation might be altered in Parkinson's patients [e.g., (53–55)] owing the deterioration of motor control. Activation in all muscles that might act as antagonists is worth being assessed in future studies to better elucidate their contribution to maximum torque/power capacity.

Although we corrected the *p*-value for our sample size, our cohorts are limited in number (and further studies with larger groups should be performed). To note, we excluded several subjects for poor-quality data not related to data collection or analysis, but to the subject inability to correctly perform the assigned task. Thus, it is suggested, with these patients (and with elderly subjects in more general terms), to perform several familiarization sessions and to utilize real-time feedback to control the data quality during data collection.

In the present study, all participants were tested during the medication "on" condition of the dopaminergic treatment. Levodopa alleviates rigidity, rest tremors, and bradykinesia but could also affect EMG activity, force production and passive muscle tone in these patients (56–59). The "increase" in muscle tone during the "off" phase of the pharmacological treatment is expected to exacerbate the impairment in muscle's contractile capacity; on the other hand, the symptoms that are alleviated by levodopa could interfere with the evaluation of the muscle's contractile capacity.

Different fitting procedures (for the T-V relationship) provide different estimates of V_{max} (e.g., (60). We tested both the linear fitting and the linear-hyperbolic fitting (as proposed by (41), but no significant differences were appreciated in V_{max} comparing these two modalities. Hence, we opted for the linear fitting, due to its simplicity.

Due to the age-dependent reduction in the quality of ultrasound images (61) and the low sampling rate it was not possible to obtain high-quality data during the dynamic contraction phase in the MCVs; for this reason, only values of MT, FL and PA at rest and during contraction (at steady state) are reported in this study.

Last but not least, we focused on the peripheral factors known to affect muscle function (dynamic muscle shape changes), but the differences we observed could also be attributed to neural/nervous factors not investigated in this study.

Conclusions

The present study revealed significant differences between PD patients and age and physically matched control groups in terms of maximum isometric torque (calculated as the apex of the T-L curve) and maximum mechanical power (apex of the P-V curve), while the maximum angular velocity of the knee extensor muscles was not affected by the pathology. Among the mechanical variables investigated, mechanical power was able to identify differences between the less and the more affected side of the body in PD patients, suggesting that this parameter could be useful for clinical evaluation in these patients. The observed results cannot be explained by differences in muscle geometry at rest, but rather by the muscle capacity to change in shape during contraction, that is affected in PD patients.

Data availability statement

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

Ethics statement

All participants received written and oral information and instructions before the study and gave their written informed consent to the experimental procedure. The experimental protocol was approved by the Ethical Committee of the University of Verona (protocol number: 2021-UNVRCLE-0450152) and was performed in accordance with the Helsinki Declaration.

Author contributions

RM: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Software, Validation, Writing – original draft, Writing – review & editing. FN: Conceptualization, Data curation, Investigation, Methodology, Writing – original draft, Writing – review & editing. FB: Project administration, Supervision, Writing – original draft, Writing – review & editing, Methodology, Validation. AM: Conceptualization, Data curation, Formal Analysis, Methodology, Supervision, Validation, Writing – original draft, Writing – review & editing. PZ: Conceptualization, Data curation, Funding acquisition, Methodology, Project administration, Resources, Validation, Visualization, Writing – original draft, Writing – review & editing.

Funding

The author(s) declare that no financial support was received for the research, authorship, and/or publication of this article.

Acknowledgments

We thank the "fondo straordinario di Ateneo per la pubblicazione in Open Access" for paying the Open Access.

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.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2024.1380864/full#supplementary-material

References

1. Bohm S, Mersmann F, Santuz A, Schroll A, Arampatzis A. Muscle-specific economy of force generation and efficiency of work production during human running. *eLife*. (2021) 10:e67182. doi: 10.7554/eLife.67182

2. Raj IS, Bird SR, Shield AJ. Aging and the force–velocity relationship of muscles. *Exp Gerontol*. (2010) 45(2):81–90. doi: 10.1016/j.exger.2009.10.013

3. Van Roie E, Verschueren SM, Boonen S, Bogaerts A, Kennis E, Coudyzer W, et al. Force-velocity characteristics of the knee extensors: an indication of the risk for physical frailty in elderly women. *Arch Phys Med Rehabil*. (2011) 92(11):1827–32. doi: 10.1016/j.apmr.2011.05.022

4. Haeger A, Costa AS, Schulz JB, Reetz K. Cerebral changes improved by physical activity during cognitive decline: a systematic review on MRI studies. *NeuroImage Clin*. (2019) 23:101933. doi: 10.1016/j.nicl.2019.101933

5. Alcazar J, Alegre LM, Suetta C, Júdice PB, Van Roie E, González-Gross M, et al. Threshold of relative muscle power required to rise from a chair and mobility limitations and disability in older adults. *Med Sci Sports Exerc*. (2021) 53 (11):2217–24. doi: 10.1249/MSS.0000000000002717

6. Losa-Reyna J, Alcazar J, Carnicero J, Alfaro-Acha A, Castillo-Gallego C, Rosado-Artalejo C, et al. Impact of relative muscle power on hospitalization and all-cause mortality in older adults. *J Gerontol*. (2022) 77(4):781–9. doi: 10.1093/gerona/glab230

7. Balestrino R, Schapira AHV. Parkinson Disease. *Eur J Neurol*. (2020) 27(1):27–42. doi: 10.1111/ene.14108

8. Cano-De-La-Cuerda R, Pérez-De-Heredia M, Miangolarra-Page JC, Muñoz-Hellín E, Fernández-De-Las-Peñas C. Is there muscular weakness in Parkinson's disease? *Am J Phys Med Rehabil*. (2010) 89(1):70–6. doi: 10.1097/PHM.0b013e3181a9ed9b

9. Stevens-Lapsley J, Kluger BM, Schenkman M. Quadriceps muscle weakness, activation deficits, and fatigue with Parkinson disease. *Neurorehabil Neural Repair*. (2012) 26(5):533–41. doi: 10.1177/1545968311425925

10. Nishikawa Y, Watanabe K, Takahashi T, Orita N, Kimura H, Matsumoto M, et al. Spatial electromyography distribution pattern of the vastus lateralis muscle during ramp up contractions in Parkinson's disease patients. *J Electromyogr Kinesiol*. (2017) 37:125–31. doi: 10.1016/j.jelekin.2017.10.003

11. Smart RR, Richardson CM, Wile DJ, Dalton BH, Jakobi JM. Importance of maximal strength and muscle-tendon mechanics for improving force steadiness in persons with Parkinson's disease. *Brain Sci*. (2020) 10(8):1–12. doi: 10.3390/brainsci10080471

12. Marusiak J, Kisiel-Sajewicz K, Jaskólska A, Jaskólski A. Higher muscle passive stiffness in Parkinson's disease patients than in controls measured by myotonometry. *Arch Phys Med Rehabil*. (2010) 91(5):800–2. doi: 10.1016/j.apmr.2010.01.012

13. Monte A, Magris R, Nardello F, Bombieri F, Zamparo P. Muscle shape changes in Parkinson's disease impair function during rapid contractions. *Acta Physiol*. (2023) 238(1):e13957. doi: 10.1111/apha.13957

14. David FJ, Rafferty MR, Robichaud JA, Prodoehl J, Kohrt WM, Vaillancourt DE, et al. Progressive resistance exercise and Parkinson's disease: a review of potential mechanisms. *Parkinson's Disease*. (2011) 2012:e124527. doi: 10.1155/2012/124527

15. Martignon C, Ruzzante F, Giuriato G, Laginestra FG, Pedrinolla A, Di Vico IA, et al. The key role of physical activity against the neuromuscular deterioration in patients with Parkinson's disease. *Acta Physiol*. (2021) 231(4):e13630. doi: 10.1111/apha.13630

16. Rose MH, Løkkegaard A, Sonne-Holm S, Jensen BR. Tremor irregularity, torque steadiness and rate of force development in Parkinson's disease. *Motor Control*. (2013) 17(2):203–16. doi: 10.1123/mcj.17.2.203

17. Glendinning DS, Enoka RM. Motor unit behavior in Parkinson's disease. *Phys Ther*. (1994) 74(1):61–70. doi: 10.1093/ptj/74.1.61

18. Narici MV, Maffulli N, Maganaris CN. Ageing of human muscles and tendons. *Disabil Rehabil*. (2008) 30(20–22):1548–54. doi: 10.1080/09638280701831058

19. Venturelli M, Saggin P, Muti E, Naro F, Cancellara L, Toniolo L, et al. In vivo and in vitro evidence that intrinsic upper- and lower-limb skeletal muscle function is unaffected by ageing and disuse in oldest-old humans. *Acta Physiol*. (2015) 215 (1):58–71. doi: 10.1111/apha.12524

20. Campbell EL, Seynnes OR, Bottinelli R, McPhee JS, Atherton PJ, Jones DA, et al. Skeletal muscle adaptations to physical inactivity and subsequent retraining in young men. *Biogerontology*. (2013) 14(3):247–59. doi: 10.1007/s10522-013-9427-6

21. Beck ON, Trejo LH, Schroeder JN, Franz JR, Sawicki GS. Shorter muscle fascicle operating lengths increase the metabolic cost of cyclic force production. *J Appl Physiol*. (2022) 133(3):524–33. doi: 10.1152/japplphysiol.00720.2021

22. Nikolaidou ME, Marzilger R, Bohm S, Mersmann F, Arampatzis A. Operating length and velocity of human M. vastus lateralis fascicles during vertical jumping. *R Soc Open Sci*. (2017) 4(5):170185. doi: 10.1098/rsos.170185

23. Herzog W, Abrahamse SK, ter Keurs HEDJ. Theoretical determination of force-length relations of intact human skeletal muscles using the cross-bridge model. *Pflügers Arch Eur J Physiol*. (1990) 416(1–2):113–9. doi: 10.1007/BF00370231

24. Arampatzis A, Karamanidis K, De Monte G, Stafilidis S, Morey-Klapsing G, Brüggemann GP. Differences between measured and resultant joint moments during voluntary and artificially elicited isometric knee extension contractions. *Clin Biomech*. (2004) 19(3):277–83. doi: 10.1016/j.clinbiomech.2003.11.011

25. Bakenecker P, Raiteri B, Hahn D. Patella tendon moment arm function considerations for human vastus lateralis force estimates. *J Biomech*. (2019) 86:225–31. doi: 10.1016/j.jbiomech.2019.01.042

26. Tillin NA, Pain MTG, Folland JP. Short-term training for explosive strength causes neural and mechanical adaptations. *Exp Physiol*. (2012) 97(5):630–41. doi: 10.1113/expphysiol.2011.063040

27. Tillin NA, Pain MTG, Folland JP. Contraction speed and type influences rapid utilisation of available muscle force: neural and contractile mechanisms. *J Exp Biol*. (2018) 221:24. doi: 10.1242/jeb.193367

28. Monte A, Bertucco M, Magris R, Zamparo P. Muscle belly gearing positively affects the force–velocity and power–velocity relationships during explosive dynamic contractions. *Front Physiol*. (2021) 12:1–13. doi: 10.3389/fphys.2021.683931

29. Craig CL, Marshall AL, Sjöström M, Bauman AE, Booth ML, Ainsworth BE, et al. International physical activity questionnaire: 12-country reliability and validity. *Med Sci Sports Exercise*. (2003) 35(8):1381–95. doi: 10.1249/01.MSS.0000078924.61453.FB

30. Tombaugh TN, Mclntyre NJ. The Mini-mental state examination: a comprehensive review. *Prog Geriatr*. (1992) 40(9):922–35. doi: 10.1111/j.1532-5415.1992.tb01992.x

31. Hoehn MM, Yahr MD. Parkinsonism: onset, progression, and mortality. *Neurology*. (1967) 57(10 SUPPL. 3):427–42. doi: 10.1212/wnl.50.2.318

32. Richards M, Marder K, Cote L, Mayeux R. UPDRS interrater reliability. *Mov Disord*. (1994) 9(1):89–91. doi: 10.1002/mds.870090114

33. Kellis E, Baltzopoulos V. Gravitational moment correction in isokinetic dynamometry using anthropometric data. *Med Sci Sports Exercise*. (1996) 28 (7):900–7. doi: 10.1097/00005768-199607000-00018

34. Franchi MV, Raiteri BJ, Longo S, Sinha S, Narici MV, Csapo R. Muscle architecture assessment: strengths, shortcomings and new frontiers of in vivo imaging techniques. *Ultrasound Med Biol*. (2018) 44(12):2492–504. doi: 10.1016/j.ultrasmedbio.2018.07.010

35. Monte A, Zamparo P. Correlations between muscle-tendon parameters and acceleration ability in 20m sprints. *PLoS One*. (2019) 14(3):1–13. doi: 10.1371/journal.pone.0213347

36. Hahn D, Bakenecker P, Zinke F. Neuromuscular performance of maximal voluntary explosive concentric contractions is influenced by angular acceleration. *Scand J Med Sci Sports*. (2017) 27(12):1739–49. doi: 10.1111/sms.12812

37. Monte A, Baltzopoulos V, Maganaris CN, Zamparo P. Gastrocnemius medialis and vastus lateralis in vivo muscle-tendon behavior during running at increasing speeds. *Scand J Med Sci Sports*. (2020) 30(7):1163–76. doi: 10.1111/sms.13662

38. Randhawa A, Jackman ME, Wakeling JM. Muscle gearing during isotonic and isokinetic movements in the ankle plantarflexors. *Eur J Appl Physiol*. (2013) 113 (2):437–47. doi: 10.1007/s00421-012-2448-z

39. Iossifidou AN, Baltzopoulos V. Inertial effects on the assessment of performance in isokinetic dynamometry. *Int J Sports Med*. (1998) 19:567–73. doi: 10.1055/s-2007-971961

40. Iossifidou AN, Baltzopoulos V. Inertial effects on moment development during isokinetic concentric knee extension testing. *J Orthop Sports Phys Ther*. (2000) 30:317–23; discussion 324–327. doi: 10.2519/jospt.2000.30.6.317

41. Alcazar J, Pareja F, Carlos B, Lopez R, Reguero HG, Sanchez J, et al. A novel equation that incorporates the linear and hyperbolic nature of the force—velocity relationship in lower and upper limb exercises. *Eur J Appl Physiol*. (2022) 122 (10):2305–13. doi: 10.1007/s00421-022-05006-1

42. Gómez-de-Mariscal E, Guerrero V, Sneider A, Jayatilaka H, Phillip JM, Wirtz D, et al. Use of the p-values as a size-dependent function to address practical differences when analyzing large datasets. *Sci Rep*. (2021) 11:20942. doi: 10.1038/s41598-021-00199-5

43. Alcazar J, Rodriguez-Lopez C, Delecluse C, Thomis M, Van Roie E. Ten-year longitudinal changes in muscle power, force, and velocity in young, middle-aged, and older adults. *J Cachexia Sarcopenia Muscle*. (2023) 14(2):1019–32. doi: 10.1002/jcsm.13184

44. Narici MV, Maganaris CN, Reeves ND, Capodaglio P. Effect of aging on human muscle architecture. *J Appl Physiol*. (2003) 95(6):2229–34. doi: 10.1152/japplphysiol.00433.2003

45. Tinetti ME, Kumar C. The patient who falls: "it's always a trade-off". *JAMA*. (2010) 303(3):258. doi: 10.1001/jama.2009.2024

46. Hauraix H, Dorel S, Rabita G, Guilhem G, Nordez A. Muscle fascicle shortening behaviour of vastus lateralis during a maximal force–velocity test. *Eur J Appl Physiol*. (2017) 117:289–99. doi: 10.1007/s00421-016-3518-4

47. De Brito Fontana H, Roesler H, Herzog W. In vivo vastus lateralis force–velocity relationship at the fascicle and muscle tendon unit level. *J Electromyogr Kinesiol*. (2014) 24:934–40. doi: 10.1016/j.jelekin.2014.06.010

48. Allison SJ, Brooke-Wavell K, Folland JP. Multiple joint muscle function with ageing: the force–velocity and power–velocity relationships in young and older men. *Aging Clin Exp Res*. (2013) 25(2):159–66. doi: 10.1007/s40520-013-0024-y

49. Kruse A, Rivares C, Weide G, Tilp M, Jaspers RT. Stimuli for adaptations in muscle length and the length range of active force exertion—a narrative review. *Front Physiol*. (2021) 12:742034. doi: 10.3389/fphys.2021.742034

50. Lieber RL, Fridén J. Functional and clinical significance of skeletal muscle architecture. *Muscle Nerve*. (2000) 23(11):1647–66. doi: 10.1002/1097-4598(200011)23:11<1647::AID-MUS1>3.0.CO;2-M

51. Islam A, Alcock L, Nazarpour K, Rochester L, Pantall A. Effect of Parkinson's disease and two therapeutic interventions on muscle activity during walking: a systematic review. *Npj Parkinson's Disease*. (2020) 6:22. doi: 10.1038/s41531-020-00119-w

52. Zanardi APJ, da Silva ES, Costa RR, Passos-Monteiro E, dos Santos IO, Kruel LFM, et al. Gait parameters of Parkinson's disease compared with healthy controls: a systematic review and meta-analysis. *Sci Rep*. (2021) 11(1):1. doi: 10.1038/s41598-020-80768-2

53. Busse ME, Wiles CM, van Deursen RW. Co-activation: its association with weakness and specific neurological pathology. *J Neuroeng Rehabil*. (2006) 3(1):26. doi: 10.1186/1743-0003-3-26

54. Dietz V, Zijlstra W, Prokop T, Berger W. Leg muscle activation during gait in Parkinson's disease: adaptation and interlimb coordination. *Electroencephalogr Clin Neurophysiol*. (1995) 97(6):408–15. doi: 10.1016/0924-980x(95)00109-x

55. Moreno Catalá M, Woitalla D, Arampatzis A. Central factors explain muscle weakness in young fallers with Parkinson's disease. *Neurorehabil Neural Repair*. (2013) 27(8):753–9. doi: 10.1177/1545968313491011

56. Armstrong MJ, Okun MS. Diagnosis and treatment of Parkinson disease: a review. *JAMA*. (2020) 323(6):548–60. doi: 10.1001/jama.2019.22360

57. Nishikawa Y, Watanabe K, Takahashi T, Kimura H, Maruyama H. The effect of medication on vastus lateralis muscle activation patterns in Parkinson's disease patients. *J Electromyogr Kinesiol*. (2018) 42:66–73. doi: 10.1016/j.jelekin.2018.06.014

58. Pedersen SW, Oberg B. Dynamic strength in Parkinson's disease. Quantitative measurements following withdrawal of medication. *Eur Neurol*. (1993) 33 (2):97–102. doi: 10.1159/000116913

59. Robichaud JA, Pfann KD, Comella CL, Corcos DM. Effect of medication on EMG patterns in individuals with Parkinson's disease. *Mov Disord*. (2002) 17 (5):950–60. doi: 10.1002/mds.10218

60. Alcazar J, Csapo R, Ara I, Alegre LM. On the shape of the force-velocity relationship in skeletal muscles: the linear, the hyperbolic, and the double-hyperbolic. *Front Physiol*. (2019) 10:769. doi: 10.3389/fphys.2019.00769

61. Arts IM, Pillen S, Schelhaas HJ, Overeem S, Zwarts MJ. Normal values for quantitative muscle ultrasonography in adults. *Muscle Nerve*. (2010) 41:32–41. doi: 10.1002/mus.21458

frontiers | Frontiers in Sports and Active Living

TYPE Brief Research Report
PUBLISHED 23 May 2024
DOI 10.3389/fspor.2024.1391868

OPEN ACCESS

EDITED BY
Adamantios Arampatzis,
Humboldt University of Berlin, Germany

REVIEWED BY
Žiga Kozinc,
University of Primorska, Slovenia
Valerio Giustino,
University of Palermo, Italy

*CORRESPONDENCE
Katharina Borgmann
✉ katharina.borgmann1@uni-due.de

RECEIVED 26 February 2024
ACCEPTED 14 May 2024
PUBLISHED 23 May 2024

CITATION
Borgmann K, Ferdenhert J, Neyses AC, Bauer J, Hill MW and Muehlbauer T (2024) Exploring how arm movement moderates the effect of lower limb muscle fatigue on dynamic balance in healthy youth.
Front. Sports Act. Living 6:1391868.
doi: 10.3389/fspor.2024.1391868

Exploring how arm movement moderates the effect of lower limb muscle fatigue on dynamic balance in healthy youth

Katharina Borgmann[1]*, Jendrik Ferdenhert[1], Alexandra C. Neyses[1], Julian Bauer[2], Mathew W. Hill[3] and Thomas Muehlbauer[1]

[1]Division of Movement and Training Sciences/Biomechanics of Sport, University of Duisburg-Essen, Essen, Germany, [2]Department of Sport Science, Human Performance Research Centre, University of Konstanz, Konstanz, Germany, [3]Centre for Physical Activity, Sport and Exercise Sciences, Coventry University, Coventry, United Kingdom

Background: In young adults, there is evidence that free arm movements do not help to compensate muscle fatigue-induced deteriorations in dynamic balance performance. However, the postural control system in youth is immature, and as a result, the use of arm movements may provide a compensatory "upper body strategy" to correct fatigue-related balance impairments. Thus, the purpose of the present study was to compare the effects of free vs. restricted arm movement on dynamic balance performance prior and following exercise-induced muscle fatigue.

Methods: Forty-three healthy youth (19 females; mean age: 12.8 ± 1.9 years) performed the Y Balance Test–Lower Quarter before and immediately after a fatiguing exercise (i.e., repetitive vertical bipedal box jumps until failure) using two different arm positions: free (move the arms freely) and restricted (keep the arms akimbo) arm movement.

Results: Muscle fatigue ($p \le 0.033$; $0.10 \le \eta_p^2 \le 0.33$) and restriction of arm movement ($p \le 0.005$; $0.17 \le \eta_p^2 \le 0.46$) resulted in significantly deteriorated dynamic balance performance. However, the interactions between the two did not reach the level of significance ($p \ge 0.091$; $0.01 \le \eta_p^2 \le 0.07$).

Conclusion: Our findings indicate that the use of an "upper body strategy" (i.e., free arm position) has no compensatory effect on muscle fatigue-induced dynamic balance deteriorations in healthy youth.

KEYWORDS
postural control, upper body strategy, arm position, lower extremities, reaching movement, exhaustion, youth

1 Introduction

The negative influence of motor performance fatigue (i.e., reversible exercise-induced reduction in neuromuscular performance) on postural control in healthy youth is firmly established (1–5). For example, Steinberg et al. (3) investigated fatigue-induced performance changes in boys ($N = 13$; mean age: 11.5 ± 1.8 years) and girls ($N = 17$; mean age: 13.8 ± 2.9 years). The authors reported that postural sway was significantly increased while standing on both legs immediately after a 20-m shuttle-run aerobic

Abbreviations
ANOVA, analysis of variance; AT, anterior; CS, composite score; LL, leg length; PM, posteromedial; PL, posterolateral; SD, standard deviation; YBT–LQ, Y balance test–lower quarter.

fatigue test. Moreover, Pau et al. (2) compared balance performance before and after a repeated sprint ability test (i.e., 6 repetitions of maximal 2×15-m shuttle sprints) in adolescents ($N = 21$; mean age: 14.5 ± 0.2 years) and showed significantly increased sway values during single and double leg stance. Lastly, Guan et al. (4) tested 13 children aged between 9 and 11 years that performed a fatigue protocol consisting of two sets of 30-s double leg kicks at the maximum frequency and consecutive frog jumps until exhaustion. After finishing the protocol, reach distances for the Y Balance Test–Lower Quarter (YBT–LQ) were significantly reduced.

At the same time, there is growing evidence that arm movements can contribute to stabilise balance in children and adolescents (6, 7). For instance, Hill et al. (6) investigated 29 children (mean age: 10.6 ± 0.5 years) that completed several dynamic balance assessments with free and restricted arm movement. The free use of arm movement resulted in significantly larger YBT–LQ reach distances and shorter durations for the timed balance beam walking test. Further, Muehlbauer et al. (7) studied 40 children (mean age: 11.5 ± 0.6 years) and 30 adolescents (mean age: 14.0 ± 1.1 years) that performed static and dynamic balance tests under free vs. restricted arm movement conditions. The results showed better performance values for the single leg stance test, the YBT–LQ, and the 3-m beam walking backward test when participants were instructed to move their arms freely instead of to claps their hands in front of the body at the waist during the balance tasks.

Although the benefit of free arm movement on postural control is well known, the possibility of a compensatory effect on muscle fatigue-induced impairments in balance performance has hardly been investigated so far. To date, there is only one study[1] from our lab that examined the role of arm movements during dynamic balance testing before and after lower limb muscle fatigue. Specifically, healthy young adults ($N = 52$; mean age $=$ 22.6 ± 1.6 years) performed the YBT–LQ with free and restricted arm movements before and immediately after a fatiguing exercise (i.e., repetitive vertical bipedal box jumps until failure). We found that restriction of arm movement and application of fatigue independently, but not the interaction between the two, resulted in significantly deteriorated lower limb reach distances. Accordingly, in young adults, free arm movement does not seem to compensate for muscle fatigue-induced dynamic balance deterioration. However, a direct transfer of these findings to healthy youth is not possible because the maturation of postural control mechanisms is still incomplete (8–10). Specifically, deficits in static and dynamic balance performance are evident in children and adolescents when compared to young adults (11, 12), indicating different strategies for balance control (8, 13). Therefore, free arm movements may have a significant role in compensating for muscle fatigue-induced performance deteriorations in dynamic balance in youth. Thus, exploring how arm movement moderates the effect of lower limb muscle fatigue on measures of dynamic balance in healthy children and adolescents will enable us to better understand how the upper body is used to control posture in youth. From a practical perspective, the present study could add insights on how to effectively design balance training programmes. Precisely, allowing free arm movements would be a relatively simple task manipulation that could be used in long-lasting programmes to continue balance training despite exercise-induced muscle fatigue.

The current study explored how arm movement moderates the effect of lower limb muscle fatigue on dynamic balance in healthy youth. To reject or confirm previously reported effects of muscle fatigue on dynamic balance performance in healthy young adults[1], we applied the same methodology in terms of experimental procedure, fatigue protocol, and dynamic balance assessment. Based on previous work (4, 6), we hypothesised that lower limb muscle fatigue and restricted arm movement would lead to impaired dynamic balance performance, but performance impairment when fatigued would be less evident when the participants are allowed to use their arms for postural control during balance testing.

2 Material and methods

2.1 Participants and sample size estimation

Forty-three healthy, physically active subjects voluntarily participated in the present study. Their characteristics are shown in Table 1. With the help of G*Power 3.1.9.8 (14), an *a priori* power analysis ($f = 0.25$, $\alpha = 0.05$, $1\text{-}\beta = 0.80$, number of groups: $n = 1$, number of measurements: $n = 4$) was performed for measures of dynamic balance performance (15, 16). The analysis revealed that $N = 41$ participants would be sufficient to find statistically significant repeated measures analysis of variance (ANOVA) effects. The participants were recruited via an information event from public primary and secondary schools in the Ruhr area of North Rhine-Westphalia, Germany. Inclusion criteria were willingness to participate and age 10–16 years. We excluded participants suffering from any problems that may interact with postural control including musculoskeletal dysfunction, neurological impairment, orthopaedic pathology, or an injury during the last three months. Participant's assent and written informed consent of the parents or legal guardians were obtained before the start of the study. The Human Ethics Committee at the University of Duisburg-Essen, Faculty of Educational Sciences approved the study protocol.

2.2 Experimental procedure

A single-group repeated-measures design that included two sessions separated by one week was used to assess the effects

[1]Borgmann K, Brinkmann R, Bauer J, Hill MW, Muehlbauer T. Effect of lower limb muscle fatigue on dynamic balance performance in healthy young adults: role of arm movement. *Sci Rep*. (2024).

TABLE 1 Participant characteristics.

Characteristic	Youth ($n = 43$)
Sex (females; n)	19
Mean age (years)	12.8 ± 1.9
Age range (years)	10–16
Body height (cm)	159.7 ± 10.2
Body mass (kg)	52.8 ± 12.5
Body mass index (kg/m^2)	19.6 ± 5.5
Leg length (cm)	89.3 ± 7.2

Values are means ± standard deviations.

of exercise-induced lower limb muscle fatigue on measures of dynamic balance performance (Figure 1). At the beginning of each testing session, participants received instructions on the specific procedure. Afterwards, a standardised warm-up protocol was conducted that consisted of three minutes of rope skipping, two minutes of active stretching exercises for the lower body (i.e., calf muscles, quadriceps muscles, hamstring muscles, hip muscles), and two minutes of a familiarisation phase with submaximal single leg reaching movements. Thereafter, the participants executed the YBT–LQ that was followed by a rating of their initial perceived exertion. Afterwards, they performed the fatigue protocol, followed by another rating of their perceived execution. Immediately afterwards, the YBT–LQ was carried out again and the experimental procedure ended. The same procedure was repeated one week later. The permission to use (free) or not to use (restricted) arm movements while performing the single leg reaching movements prior and following the fatigue protocol was randomised between participants to avoid potential bias. Precisely, the free source Research Randomizer (www.randomizer.org) was used to randomly assign the participants to the experimental conditions. For the free arm movement condition, participants were instructed to move their arms freely and to their advantage. For the restricted arm movement condition, participants were asked to keep their arms akimbo and compliance was visually monitored. The experimental procedure described above was carried out in the morning (between 9 and 11 a.m.) at a room temperature of approx. 20°C in the gym hall of the respective school.

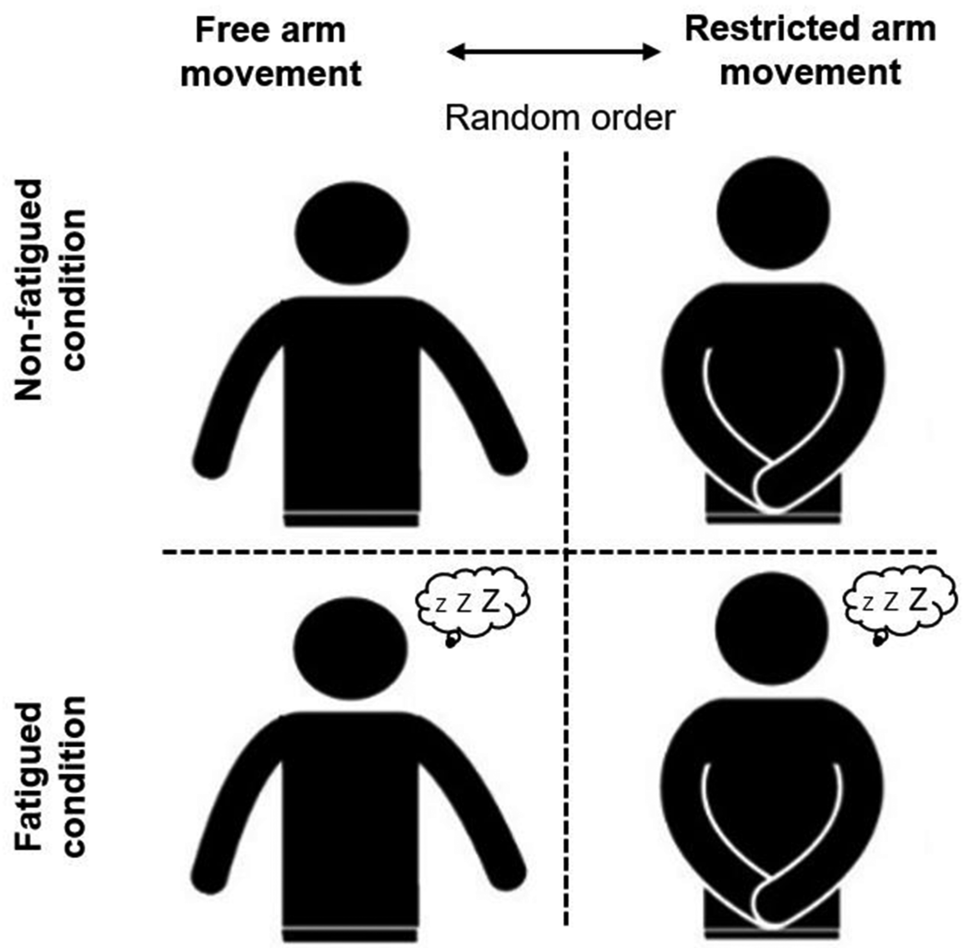

FIGURE 1
Schematic diagram of the different experimental conditions.

2.3 Anthropometric measurements

The participants were asked to stand straight and upright without shoes while their body height was measured to the nearest 0.1 cm using a Seca 217 linear measurement scale (Seca, Basel, Switzerland). Participants wore light sportswear but no shoes when their body mass was measured to the nearest 100 g using a Seca 803 electronic scale (Seca, Basel, Switzerland). The body mass index was calculated by dividing the body mass by the body height squared (kg/m^2).

2.4 Fatigue protocol

Lower limb muscle fatigue was individually induced by repeated vertical bipedal box jumps. Participants performed as many metronome-paced (70 beats per minute) box jumps (box height: 28 cm) as possible until failure (17). Failure was defined as the time when the participants could no longer follow the pace of the metronome. The number of repetitions in the first set represented the reference (i.e., 100%) for the following set. During this set, participants were again instructed to perform as many repetitions as possible until failure. If at least 60% of the first set was achieved (18), another set followed, otherwise the fatigue protocol ended. The rest period between sets amounted to one minute. The number of sets as well as repetitions per set was manually recorded.

2.5 Dynamic balance assessment

The YBT–LQ was administered using the Y Balance Test Kit (Functional Movement Systems, Chatham, USA). The device consists of a central standing platform on which three tubes are attached in different directions. These indicate the reach directions anterior (AT), posteromedial (PM), and posterolateral (PL) and are marked in 1.0 cm increments for measurement purposes. The three tubes were equipped with a movable reach indicator. Participants were instructed to stand without shoes on the central platform with their dominant leg. To determine the dominant leg, the participants were asked, "Which foot do you use to kick a ball?". In three practice trials, followed by three data collection trials for each reach direction, participants were asked to use their free leg to push the reach indicator as far as possible in the AT, PM, and PL directions. A trial was discarded and repeated if a participant (a) failed to maintain the one-legged stance (i.e., touched the ground with the reach leg), (b) raised the supporting leg from the central platform, (c) used the reach indicator for support of body weight, (d) failed to maintain contact with the reach indicator at the most distal point, (e) failed to return the reach leg to the centre of the central platform, or (f) released the arms from the hips with limited arm movement has. The trial with the greatest reach distance (measured in cm) per reach direction was used for the subsequent analyses. More specifically, the distance per reach direction was normalized [% leg length (LL)] by dividing the maximum reach distance (measured in cm) by the dominant lower limb length (measured in cm) and multiplying by 100. Additionally, the composite score (CS) was calculated (%LL) (19). This represents the sum of the maximum range (cm) per range direction divided by three times LL (cm). The result is then multiplied by 100. Lower limb length (measured in cm) was recorded from the anterior superior iliac spine to the most distal part of the medial malleolus (20). The YBT–LQ is a valid (AUC-values: ≥74%) and reliable (ICC-values: 0.40–0.96) tool to assess dynamic balance performance in children and adolescents (21, 22).

2.6 Rating of perceived exertion

We used a 6–20 Borg scale (23) to assess the level of subjectively perceived and verbally expressed exertion prior and following the fatigue protocol with 6 indicating "no exertion at all" at all and 20 indicating "maximal exertion".

2.7 Statistical analyses

Descriptive statistics were presented as group mean value ± standard deviation (*SD*). Before conducting inference statistics, normal distribution (Shapiro–Wilk Test) and variance homogeneity (Mauchly Test) were checked and confirmed. In terms of dynamic balance performance, series of 2 (fatigue level: non-fatigued, fatigued) × 2 (arm movement: free, restricted) repeated measures ANOVA were performed. Regarding perceived exertion, the Wilcoxon signed rank test was used to detect differences prior vs. immediately after the fatigue protocol. The significance level was *a priori* set at $\alpha < 0.05$. For the repeated measures ANOVA, partial eta-squared (η_p^2) was calculated and reported as small ($0.02 \leq \eta_p^2 \leq 0.12$), medium ($0.13 \leq \eta_p^2 \leq 0.25$), or large ($\eta_p^2 \geq 0.26$) (24). All analyses were performed using SPSS version 28.0 (IBM Inc., Chicago, IL).

3 Results

3.1 Trial-by-trial reliability

Irrespective of test condition and reach direction, trial-by-trial reliability was predominately "excellent" (i.e., ICC > 0.75) (data not shown).

3.2 Rating of perceived exertion

Participants performed between 2 and 5 sets of bipedal box jumps until failure, with an average jump number of 29.8 ± 16.9, 25.6 ± 16.1, 23.9 ± 14.2, and 21.0 ± 9.6 for the second, third, fourth, and fifth set, respectively. As a result, we detected significantly increased levels of perceived exertion in both arm movement conditions from "very light exertion" to "hard exertion" (free: non-fatigued = 8.9 ± 2.6; fatigued = 14.8 ± 2.4; $Z = -5.735$, $p < 0.001$; restricted: non-fatigued = 8.8 ± 2.2, fatigued = 15.1 ± 2.2; $Z = -5.724$, $p < 0.001$).

TABLE 2 Descriptive and inference statistics of dynamic balance performance by fatigue (non-fatigued vs. fatigued) and arm movement (free vs. restricted) conditions.

Outcome	Non-fatigued		Fatigued		Main effect: fatigue	Main effect: arm movement	Interaction effect: fatigue × arm movement
	Free	Restricted	Free	Restricted	p-value (η_p^2)		
AT (% LL)	70.5 ± 7.1	68.2 ± 8.3	68.4 ± 7.7	66.5 ± 7.4	**.004 (.18)**	**.005 (.17)**	.688 (.01)
PM (% LL)	102.0 ± 11.9	97.7 ± 11.7	98.8 ± 11.0	96.9 ± 9.1	**.033 (.10)**	**<.001 (.40)**	.091 (.07)
PL (% LL)	102.8 ± 11.6	97.3 ± 9.8	98.8 ± 10.7	94.0 ± 9.4	**<.001 (.29)**	**<.001 (.43)**	.609 (.01)
CS (% LL)	91.8 ± 9.1	87.7 ± 8.7	88.7 ± 8.5	85.8 ± 7.4	**<.001 (.33)**	**<.001 (.46)**	.129 (.05)

Bold values indicate statistically significant main effects ($p < .05$). Threshold values for the η_p^2-value were.02 ≤ η_p^2 ≤ .12 = small, .13 ≤ η_p^2 ≤ .25 = medium, and η_p^2 ≥ .26 = large. AT = anterior; CS, composite score; LL, leg length; PM, posteromedial; PL, posterolateral.

3.3 Dynamic balance performance

The results of the descriptive and inference statistics are shown in Table 2. We detected significant main effects of fatigue ($p \leq 0.033$; $0.10 \leq \eta_p^2 \leq 0.33$) and arm movement ($p \leq 0.005$; $0.17 \leq \eta_p^2 \leq 0.46$), indicating poorer dynamic balance performances in the fatigued compared to the non-fatigued condition and for the restricted compared to the free arm movement position (Figures 2A–D). However, we did not detect significant fatigue × arm movement interactions.

4 Discussion

The purpose of this study was to explore the effect of lower limb muscle fatigue on dynamic balance performance in free and restricted arm movement conditions in healthy youth. Our

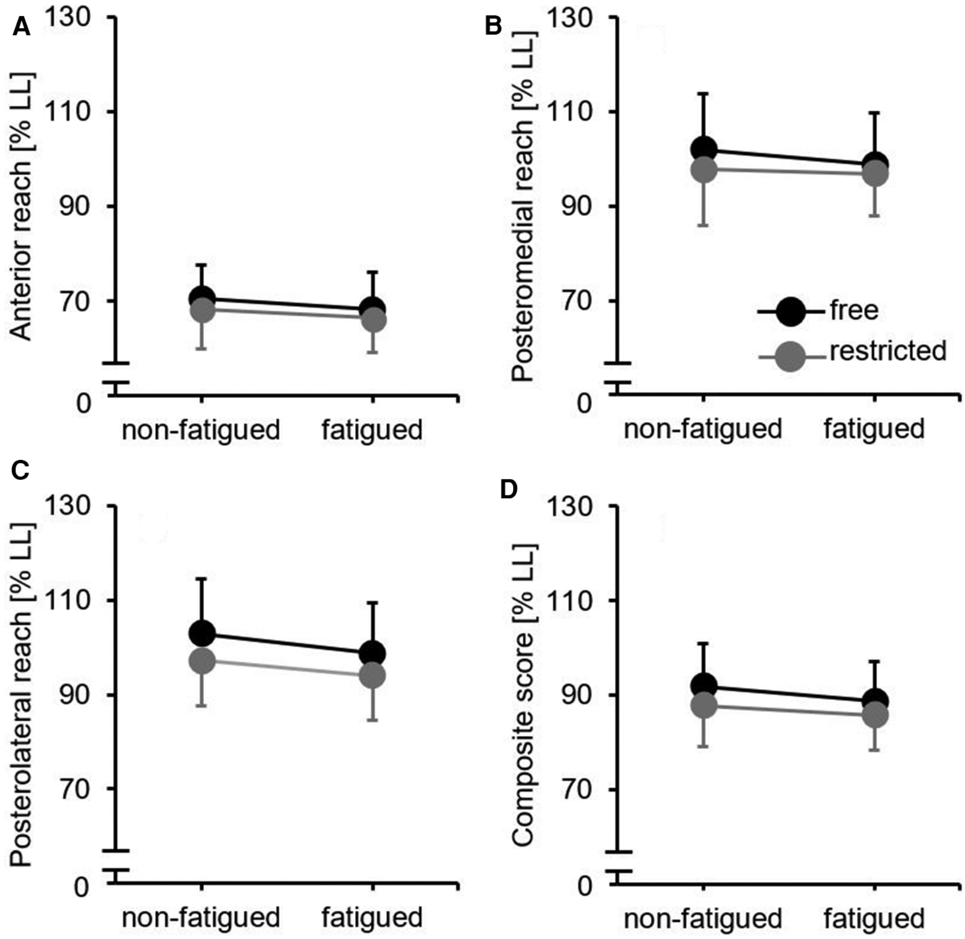

FIGURE 2
Dynamic balance performance by fatigue (non-fatigued vs. fatigued) and arm movement (free vs. restricted) conditions for (A) anterior reach direction, (B) posteromedial reach direction, (C) posterolateral reach direction, and (D) composite score. Black filled circles mean free arm movement condition and grey filled circles mean restricted arm movement condition. LL, leg length.

investigation builds on a recent study[1] with the following major findings: (a) in line with the first part our hypothesis, we detected deteriorations in dynamic balance performance under fatigued compared to non-fatigued conditions and with restricted compared to free arm movements; (b) contrary to the second part of our hypothesis, we did not observe a compensatory effect of free arm movements on muscle fatigue-induced impairments in dynamic balance performance (i.e., no fatigue by arm movement interaction).

4.1 Influence of lower limb muscle fatigue on dynamic balance performance

As expected, and in agreement with previous literature in youth (4) motor performance fatigue resulted in significantly reduced lower limb reach distances and the CS under restricted and free arm movement conditions. In this regard, Guan et al. (4) compared lower limb reach distances before and after exercised-induced muscle fatigue in children (mean age: 9.9 ± 0.8 years) and observed significantly degraded lower limb reach distances (except for the PL reach direction) and the CS. The impaired performance in dynamic balance after the fatigue protocol can be explained, among other things, by the fact that metabolic changes like the accumulation of lactic acid led to a desensitization of the muscle spindles (25). Additionally, these metabolic by-products may have been distributed to remote muscles, decreasing their voluntary activation (5) and hindering their contractile ability (26). In this context, Alderman (27) stated that a high level of fatigue can impair the accuracy of neuromuscular coordination tasks.

4.2 Contribution of arm movement to control dynamic balance

As predicted, and consistent with previous studies (6, 7), restricting arm movements resulted in significantly impaired lower limb reach distances and the CS under non-fatigued and fatigued conditions. In this context, Hill et al. (6) examined the effects of arm movements on the performance of dynamic postural tasks in 14 healthy boys and 10 girls. Restricting arm movements elicited significant deteriorations in YBT–LQ reach distances and 2-m tandem walk time on a balance beam. Furthermore, Muehlbauer et al. (7) explored the role of arm movements during dynamic balance tasks in children ($n = 40$) and adolescents ($n = 30$). Again, the restriction compared to the free use of arm movements resulted in smaller YBT–LQ reach distances and less steps while walking backward. Different mechanisms can be attributed to explain the deteriorations in dynamic balance during restricted arm movement conditions. Firstly, the arms cannot be used as a counterweight to shift the body centre of mass away from the direction of instability (28). Secondly, restoring torques to reduce angular momentum of the body (29) cannot be generated. Thirdly, the arms cannot be applied to increase the moment of inertia (6).

4.3 Compensatory effect of arm movement on fatigue-induced impairments in dynamic balance

Additionally, we further assumed that the fatigue-induced decrements in dynamic balance would be lowered when the participants are allowed to use their arms for postural control. However, we did not detect a fatigue by arm movement interaction. This result is contrary to our hypothesis but confirms the findings of our previous study in young adults[1]. In sum, this indicates that the use of an “upper body strategy” has no compensatory effect on fatigue-induced dynamic balance impairments in young adults as well as in youth. Therefore, practitioners are advised to deal with fatigue-related dynamic balance impairments by providing a rest period rather than instructions on the use of free arm movement. In this regard, Johnston et al. (15) applied the YBT–LQ before and after (0, 10, and 20 min) a fatigue protocol (i.e., modified 60-s Wingate protocol). They found that the AT reach direction returned to pre-fatigue level within 10 min and the PM reach direction within 20 min, while the PL reach direction did not return within this time. Future studies could examine whether these rest periods also apply to our fatigue protocol (i.e., repetitive vertical bipedal box jumps until failure) and can be reduced by using arm movements.

One possible reason for the absence of a fatigue by arm movement interaction could be that the YBT–LQ requires not only balance but also lower-extremity muscle strength (30), flexibility (31), and core control (32). Therefore, these additional factors could have contributed that the assumed compensatory effect of arm movement on fatigue-induced impairments in dynamic balance was not evident.

4.4 Limitations

The present study has some limitations. Firstly, we only measured behavioural data (i.e., reach distances) but no kinematic (e.g., joint angles) or electromyographical (i.e., muscle activation) data, which limits our understanding about the role of arm movements on postural control following lower limb muscle fatigue. Secondly, muscle fatigue was only assessed subjectively (i.e., 6–20 Borg scale) but not objectively (e.g., blood lactate). Thirdly, our study was limited to healthy youth and the findings cannot be directly generalized to older adults whose neuromuscular system is influenced by ageing processes, which may make them more dependent on arm movements for postural control following lower limb muscle fatigue.

5 Conclusions

The result of performance impairments following exercise-induced lower limb muscle fatigue and while performing the dynamic balance task with restricted arm movements but not the combination of both factors indicates that the "upper body strategy" (i.e., free arm position) has no compensatory effect in healthy youth. Therefore, teachers and coaches are advised to provide sufficient rest periods for neuromuscular recovery rather than to ask young individuals to use their arms freely when the goal is to compensate lower limb muscle fatigue-induced deteriorations in dynamic balance performance.

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 humans were approved by Human Ethics Committee at the University of Duisburg-Essen, Faculty of Educational Sciences. The studies were conducted in accordance with the local legislation and institutional requirements. Written informed consent for participation in this study was provided by the participants' legal guardians/next of kin. Written informed consent was obtained from the individual(s), and minor(s)' legal guardian/next of kin, for the publication of any potentially identifiable images or data included in this article.

Author contributions

KB: Conceptualization, Data curation, Formal Analysis, Methodology, Writing – original draft, Writing – review & editing. JF: Data curation, Methodology, Writing – review & editing. AN: Data curation, Methodology, Writing – review & editing. JB: Conceptualization, Methodology, Writing – review & editing. MH: Formal Analysis, Writing – review & editing. TM: Formal Analysis, Writing – original draft, Writing – review & editing.

Funding

The author(s) declare financial support was received for the research, authorship, and/or publication of this article.

Open Access funding enabled and organized by Projekt DEAL. We acknowledge support by the Open Access Publication Fund of the University of Duisburg-Essen organised by the project DEAL. The funding body is independent of the design of the study and collection, analysis, and interpretation of data and in writing the manuscript.

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.

Publisher's note

References

1. Ratel S, Duche P, Williams CA. Muscle fatigue during high-intensity exercise in children. *Sports Med.* (2006) 36(12):1031–65. doi: 10.2165/00007256-200636120-00004

2. Pau M, Ibba G, Attene G. Fatigue-induced balance impairment in young soccer players. *J Athl Train.* (2014) 49(4):454–61. doi: 10.4085/1062-6050-49.2.12

3. Steinberg N, Eliakim A, Zaav A, Pantanowitz M, Halumi M, Eisenstein T, et al. Postural balance following aerobic fatigue tests: a longitudinal study among young athletes. *J Mot Behav.* (2016) 48(4):332–40. doi: 10.1080/00222895.2015.1095153

4. Guan Y, Bredin S, Jiang Q, Taunton J, Li Y, Wu N, et al. The effect of fatigue on asymmetry between lower limbs in functional performances in elite child taekwondo athletes. *J Orthop Surg Res.* (2021) 16(1):33. doi: 10.1186/s13018-020-02175-7

5. Behrens M, Gube M, Chaabene H, Prieske O, Zenon A, Broscheid KC, et al. Fatigue and human performance: an updated framework. *Sports Med.* (2023) 53 (1):7–31. doi: 10.1007/s40279-022-01748-2

6. Hill MW, Wdowski MM, Pennell A, Stodden DF, Duncan MJ. Dynamic postural control in children: do the arms lend the legs a helping hand? *Front Physiol.* (2019) 9:1932. doi: 10.3389/fphys.2018.01932

7. Muehlbauer T, Hill MW, Heise J, Abel L, Schumann I, Brueckner D, et al. Effect of arm movement and task difficulty on balance performance in children, adolescents, and young adults. *Front Hum Neurosci.* (2022) 16:854823. doi: 10.3389/fnhum.2022.854823

8. Shumway-Cook A, Woollacott MH. The growth of stability: postural control from a development perspective. *J Mot Behav.* (1985) 17(2):131–47. doi: 10.1080/00222895.1985.10735341

9. Hirabayashi S, Iwasaki Y. Developmental perspective of sensory organization on postural control. *Brain Dev.* (1995) 17(2):111–3. doi: 10.1016/0387-7604(95)00009-Z

10. Giustino V, Messina G, Alesi M, La Malta L, Palma A, Battaglia G. Study of postural control and body balance in subjects with down syndrome. *Hum Mov.* (2021) 22(1):66–71. doi: doi:10.5114/hm.2021.98466

11. Hytonen M, Pyykko I, Aalto H, Starck J. Postural control and age. *Acta Otolaryngol.* (1993) 113(2):119–22. doi: 10.3109/00016489309135778

12. Oberg T, Karsznia A, Oberg K. Basic gait parameters: reference data for normal subjects, 10–79 years of age. *J Rehabil Res Dev.* (1993) 30(2):210–23.

13. Woollacott MH, Shumway-Cook A. Changes in posture control across the life span–a systems approach. *Phys Ther.* (1990) 70(12):799–807. doi: 10.1093/ptj/70.12.799

14. Faul F, Erdfelder E, Lang AG, Buchner A. G*power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behav Res Methods.* (2007) 39(2):175–91. doi: 10.3758/BF03193146

15. Johnston W, Dolan K, Reid N, Coughlan GF, Caulfield B. Investigating the effects of maximal anaerobic fatigue on dynamic postural control using the Y-balance test. *J Sci Med Sport.* (2018) 21(1):103–8. doi: 10.1016/j.jsams.2017.06.007

16. Hill M, Rosicka K, Wdowski M. Effect of sex and fatigue on quiet standing and dynamic balance and lower extremity muscle stiffness. *Eur J Appl Physiol.* (2022) 122 (1):233–44. doi: 10.1007/s00421-021-04831-0

17. Wadden KP, Button DC, Kibele A, Behm DG. Neuromuscular fatigue recovery following rapid and slow stretch-shortening cycle movements. *Appl Physiol Nutr Metab.* (2012) 37(3):437–47. doi: 10.1139/h2012-020

18. Lesinski M, Prieske O, Demps M, Granacher U. Effects of fatigue and surface instability on neuromuscular performance during jumping. *Scand J Med Sci Sports*. (2016) 26(10):1140–50. doi: 10.1111/sms.12548

19. Picot B, Terrier R, Forestier N, Fourchet F, McKeon PO. The star excursion balance test: an update review and practical guidelines. *Int J Athl Ther Train*. (2021) 26(6):285–93. doi: 10.1123/ijatt.2020-0106

20. Plisky PJ, Gorman PP, Butler RJ, Kiesel KB, Underwood FB, Elkins B. The reliability of an instrumented device for measuring components of the star excursion balance test. *N Am J Sports Phys Ther*. (2009) 4(2):92–9.

21. Schwiertz G, Brueckner D, Schedler S, Kiss R, Muehlbauer T. Performance and reliability of the lower quarter Y balance test in healthy adolescents from grade 6 to 11. *Gait Posture*. (2019) 67:142–6. doi: 10.1016/j.gaitpost.2018.10.011

22. Schwiertz G, Beurskens R, Muehlbauer T. Discriminative validity of the lower and upper quarter Y balance test performance: a comparison between healthy trained and untrained youth. *BMC Sports Sci Med Rehabil*. (2020) 12(1):73. doi: 10.1186/s13102-020-00220-w

23. Borg G. Perceived exertion as an indicator of somatic stress. *Scand J Rehabil Med*. (1970) 2(2):92–8. doi: 10.2340/1650197719702239298

24. Cohen J. *Statistical Power for the Behavioral Sciences*. Hillsdale, NJ: Erlbaum (1988).

25. Pedersen J, Lonn J, Hellstrom F, Djupsjobacka M, Johansson H. Localized muscle fatigue decreases the acuity of the movement sense in the human shoulder. *Med Sci Sports Exerc*. (1999) 31(7):1047–52. doi: 10.1097/00005768-199907000-00019

26. Halperin I, Chapman DW, Behm DG. Non-local muscle fatigue: effects and possible mechanisms. *Eur J Appl Physiol*. (2015) 115(10):2031–48. doi: 10.1007/s00421-015-3249-y

27. Alderman RB. Influence of local fatigue on speed and accuracy in motor learning. *Res Q*. (1965) 36:131–40.

28. Marigold DS, Bethune AJ, Patla AE. Role of the unperturbed limb and arms in the reactive recovery response to an unexpected slip during locomotion. *J Neurophysiol*. (2003) 89(4):1727–37. doi: 10.1152/jn.00683.2002

29. Patel M, Buckwell D, Hawken M, Bronstein AM. Does outstretching the arms improve postural stability? *Neurosci Lett*. (2014) 579:97–100. doi: 10.1016/j.neulet.2014.07.010

30. Ambegaonkar JP, Mettinger LM, Caswell SV, Burtt A, Cortes N. Relationships between core endurance, hip strength, and balance in collegiate female athletes. *Int J Sports Phys Ther*. (2014) 9(5):604–16.

31. Overmoyer GV, Reiser RF 2nd. Relationships between lower-extremity flexibility, asymmetries, and the Y balance test. *J Strength Cond Res*. (2015) 29(5):1240–7. doi: 10.1519/JSC.0000000000000693

32. Kang MH, Kim GM, Kwon OY, Weon JH, Oh JS, An DH. Relationship between the kinematics of the trunk and lower extremity and performance on the Y-balance test. *PM R*. (2015) 7(11):1152–8. doi: 10.1016/j.pmrj.2015.05.004

frontiers | Frontiers in Sports and Active Living

TYPE Systematic Review
PUBLISHED 25 July 2024
DOI 10.3389/fspor.2024.1414633

OPEN ACCESS

EDITED BY
Lida Mademli,
Aristotle University of Thessaloniki, Greece

REVIEWED BY
Luciana Labanca,
Rizzoli Orthopedic Institute (IRCCS), Italy
Maria-Elissavet Nikolaidou,
National and Kapodistrian University of Athens, Greece

*CORRESPONDENCE
Myoung-Hwee Kim
zokimro@gmail.com

RECEIVED 09 April 2024
ACCEPTED 10 July 2024
PUBLISHED 25 July 2024

CITATION
Kim MH, Martin W, Quarmby A, Stoll J, Engel T and Cassel M (2024) Effects of sensorimotor training on functional and pain outcomes in achilles tendinopathy: a systematic review. Front. Sports Act. Living 6:1414633. doi: 10.3389/fspor.2024.1414633

Effects of sensorimotor training on functional and pain outcomes in achilles tendinopathy: a systematic review

Myoung-Hwee Kim*, Wille Martin, Andrew Quarmby, Josefine Stoll, Tilman Engel and Michael Cassel

University Outpatient Clinic, Sports Medicine & Sports Orthopaedics, University of Potsdam, Potsdam, Germany

Background: Considering the neuromuscular alterations in Achilles tendinopathy (AT), sensorimotor training (SMT) might be beneficial to restore the neuromuscular capacity of the muscle-tendon complex and thereby improve patients' functions and alleviate symptoms. However, there is still a lack of knowledge concerning the effects of SMT on improving functional (e.g., strength) and pain outcomes in this population. Thus, the purpose of this study was to synthesize current evidence to analyze the efficacy of SMT in people with AT.
Methods: A systematic electronic search was performed in PubMed, Web of Science, and Cochrane Central Register of Controlled Trials from inception to December 2023. Studies applying SMT in people with AT investigating functional or clinical pain outcomes were considered. Protocols had to incorporate balance, stabilization, proprioception, or vibration training. Patients with insertional or mid-portion AT (≥18 years age) diagnosed with clinical or sonographic evaluation were included.
Results: The search yielded 823 records. A total of three randomized controlled trials were considered eligible for the analysis. Each trial used a different SMT protocol: balance training, balance with stabilization training, or whole-body vibration training (WBVT) with other co-interventions. Most functional and pain parameters improved compared to baseline. The first study reported a decrease in pain and an increase in performance (i.e., countermovement jump height) and endurance (i.e., number of heel-raises) by 12-week use of a balance training in addition to isometric, concentric/eccentric, and eccentric exercises. The second study evaluated the four weeks effect of SMT (balance and stabilization training plus eccentric exercises) in addition to passive physiotherapy (deep frictions, ice, ultrasound), resulting in an increased plantarflexion peak torque and reduced pain levels. The third study investigating WBVT reported at 12 weeks an increase in flexibility and a decrease in tendon pain.
Discussion: SMT in addition to other co-interventions (i.e., eccentric, isometric, concentric/eccentric training, physiotherapy) showed improvements in strength, performance, muscle flexibility, and alleviated clinical outcomes of pain. SMT might therefore be useful as part of a multimodal treatment strategy protocol in patients suffering from AT. However, due to the small number of studies included and the diversity of SMT protocols, the current evidence is weak; its additional effectiveness should be evaluated.

Systematic Review Registration: https://www.crd.york.ac.uk/PROSPERO/display_record.php?RecordID=467698, Identifier CRD42023467698.

KEYWORDS

sensorimotor training, sensorimotor exercise, balance, stabilization, vibration, pain reduction, function enhancement, achilles tendinopathy management

Introduction

Achilles tendinopathy (AT) is a chronic overuse condition that causes localized pain in the Achilles tendon with functional limitations when engaging in activities (1). It is problematic for both non-athletic and athletic populations. Epidemiological studies estimated a prevalence of 2.16 cases per 1,000 patient years in the general population and up to 36% in marathon runners (2). Tendinopathy has several anatomical (e.g., tenocytes) and biomechanical pain generators (e.g., mechanosensitive receptors and ion channels) that may be involved in nociception (3). In addition, recent research revealed the neuromuscular system associated with the condition (4). Altered muscle activity patterns of plantar flexors and a lowered temporal efficiency between muscle activation and mechanical force were reported in people with AT (4).

One of the most effective management strategies for AT has been recognized as eccentric calf exercise treatment, mostly conducted by use of the Alfredson loading protocol (5). It is theorized to increase tendon volume and tensile strength by influencing the production of type I collagen (6). Eccentric training may also reduce neovascularization and associated nerve ingrowth that accompanies the emergence of pain in some cases (7). Nevertheless, a previous systematic review suggested the effect of this training may be inferior to other heavy slow exercise interventions (8). Moreover, since the aim of the exercise treatment is to generate mechanical loading for improvement in muscle-tendon strength and endurance, the protocol may not address the issues of central motor impairment that were seen in AT (9). It was found that nearly 40% of people with AT still experience pain following the eccentric exercise protocol, meaning alternative interventions may be required for persistent cases (10).

Recently, Rio et al. focused on tendon neuroplastic training (TNT) to target the motor control impairments seen in AT (11). By use of TNT, isometric or isotonic strength training is combined with an externally paced audio or visual cue as opposed to self-paced exercise training. The authors suggested that TNT stimulates the neuromuscular system more effectively than traditional exercise treatments. This was supported by a mean increase of 22.25 points in Victorian Institute of Sport Assessment - Achilles questionnaire (VISA-A) scores after TNT, compared to 16.5 points after eccentric training. Additionally, 75% of participants in the TNT group and 58% in the eccentric training group exceeded the minimal clinically important difference (12). However, a recent meta-analysis showed little clinical evidence to support the use of TNT for reducing tendon pain over standard care (13).

Sensorimotor training (SMT), which was originally designed to restore physiological motor function in people suffering from chronic musculoskeletal clinical conditions, could be an effective alternative (14). SMT employs specific exercises to optimize the coordination and integration between the body's motor and sensory systems, aiming to enhance and optimize the sensorimotor system (15, 16). SMT is thought to affect and influence various parts of the sensorimotor system, with different terms used for specific descriptions (e.g., balance, proprioception, or vibration training). It places emphasis on postural control and introduces progressively challenging exercises to the sensorimotor system without relying on input from other sensory modalities like the TNT (11). SMT also focuses on postural control in a range of circumstances, eliciting automatic and reflexive muscle and active joint stabilization (16). A progressive balance training program that provokes automatic postural stabilization can be defined as an SMT protocol (16). In a similar sense, proprioception integrates sensory information, motor output, and brain processing for postural control as the primary somatosensory system component (17). Thus, proprioception training is defined as a specific training regimen that places emphasis on utilizing somatosensory signals, including proprioceptive or tactile afferents, in order to rehabilitate the sensorimotor system (15). Vibration training can be characterized as another kind of SMT that is specifically designed to stimulate motor neuron activity. The sensorimotor activity was shown to be increased during the vibration training compared to identical exercises without vibration (18), so the training regime is widely used for pathological conditions, especially for those who require sensorimotor alterations, like neurological disorders (19). Presently, such kinds of training are common in rehabilitation (20). Considering the sensorimotor alterations in AT (9), the usage of SMT might be beneficial to restore a normal sensorimotor capacity for muscles as well as for tendons. Nevertheless, no systematic analysis has been conducted to investigate the efficacy of SMT in an AT population.

Based on the current evidence, SMT could produce clinically meaningful improvements in the management of AT. Thus, the purpose of this study was to synthesize current evidence analyzing the outcomes of functional and pain parameters in AT following SMT (balance, stabilization, proprioception, or vibration training).

Methods

Data sources and search criteria

Relevant studies were searched in electronic databases, including PubMed, Web of Science, and the Cochrane Central Register of Controlled Trials, spanning from their inception to December 27, 2023. The search strategy involved combining three primary categories: intervention (exercise, training, therapy, rehabilitation, balance, stabilization, proprioception, vibration), pathology (tendinopathy, tendinitis, tendinosis, Achillodynia, paratenonitis, peritendinitis), and anatomical location (Achilles tendon, triceps surae, calcan*, plantarflex*) using the Boolean operator "OR" between the key terms and "AND" between the categories (Table 1 shows the search terms used in PubMed). Each search term was directed to the "title and abstract" headings, and filters such as article type (clinical trial, RCTs, etc.), language (English), and species (human) were applied if a database had the capability. Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) guidelines were followed for conducting and reporting the research (21). This study was pre-registered (PROSPERO: CRD42023467698).

TABLE 1 Search terms used in PubMed.

Combiners	Terms
Intervention	Exercise OR training OR therapy OR rehabilitation OR balance OR stabilization OR proprioception OR vibration
	AND
Anatomical location	Achilles tendon OR triceps surae OR calcan* OR plantarflex*
	AND
Pathology	Tendinopathy OR tendinitis OR tendinosis OR Achillodynia OR paratenonitis OR peritendinitis

Eligibility criteria

Studies that prospectively investigated the effect of SMT on functional outcomes and clinical outcomes of pain in AT were considered. Eligible studies investigated people (≥18 years) with insertional or mid-portion AT diagnosed by clinical or sonographic evaluation. Studies were excluded if they were performed on healthy participants, children, adolescents, or animals. The functional outcomes included analysis of kinetic parameters such as strength (i.e., dorsiflexion and plantarflex peak torques), endurance (i.e., number of heel raises), and performance (i.e., jump height) or kinematic parameters such as range of motion (ROM) and muscle flexibility (i.e., passive resistive torque). For pain outcomes, the Visual Analogue Scale (VAS), VISA-A, or other measurement tools that are based on simple numerical rating scales (NRS) were considered. Studies were included if they measured a minimum of one of the outcome parameters whilst applying SMT in an AT population.

SMT was defined as a therapeutic training approach aimed at restoring normal motor function by employing specialized exercises, such as balance, stabilization, proprioception, or vibration training that optimize coordination and integration between the body's motor and sensory systems. An SMT protocol should be prescribed with specific guidelines (e.g., type, intensity, progression, and training period) that provoke the sensorimotor system through the Achilles tendon for at least four weeks. In each study, one of the groups had to implement a SMT protocol. Studies investigating the outcome parameters without groups undergoing specific or additional SMT programs were excluded. Co-interventions (e.g., eccentric training) with SMT were allowed. Included study designs were randomized controlled trials (RCTs).

Study selection

After consolidating all the articles that were searched into a data sheet, two authors (MHK and WM) independently reviewed titles and abstracts to assess eligibility according to the inclusion and exclusion criteria. Subsequently, the remaining articles underwent full-text reviews to make the final decision. The corresponding authors were supposed to be contacted to obtain the full text if any articles were inaccessible, but this was not the case. Throughout the process, disagreements between the authors were resolved by discussion and consultation with a third author (MC), which was performed when consensus between the two authors was not reached.

Methodological quality assessment

The Joanna Briggs Institute (JBI) quality assessment tools were used for the included studies. For each question, answers of "yes" or "no" were given. For "unclear" questions, the answer "no" was provided. The RCT scale consists of 13 criteria. Studies with scores of ≥9 were considered "high-quality" (equivalent to 70%), 6–8 were considered "medium-quality" (46%), and scores <6 were considered "low-quality" (39%). This JBI tool is widely used in the literature, establishing its relevance for use in systematic reviews (22, 23). Two reviewers (MHK and WM) assessed the methodological quality independently, and discrepancies were resolved by discussion.

Data extraction

Data were extracted using a standardized data form [Microsoft Excel; version (2,303 build)]. Characteristics including study information (i.e., author, year, design), participant [i.e., sample size, sex, mean age (years), mean duration of symptoms (number of months), site of injury (insertion or mid-portion)], interventions (i.e., duration, type of intervention, sets & repetitions, frequency, progression), outcomes (name), results [mean ± standard deviation (SD), *p*-value], and the adherence of intervention were extracted. If there was no applicable mean and SD, other values such as median and interquartile range (IQR) were extracted. For studies that did not provide any data for pre and post comparisons, at least the description of results was taken. Disagreements between the authors (MHK and WM) were resolved by discussion.

Data synthesis

The differences in the outcome measures before and after the interventions were described qualitatively because of the varied nature of the outcome parameters and the differences in exercise interventions.

Outcomes were categorized by task features (e.g., strength, performance) considering the SMT treatments applied. Where studies covered several relevant outcomes, the results from the single study were categorized accordingly. The levels of evidence (24) were provided to each outcome based on the methodological quality assessment with delta changes from the pre- to post-exercise as a percentage (%) where possible, to show the degree of changes in results.

Results

Study selection

The search found a total of 824 studies, from which 112 duplicates were eliminated through automated title matching, with an additional 11 duplicates confirmed manually using Microsoft Excel [version (2,303 build)]. Subsequently, 691 articles were excluded during the screening process of titles and abstracts. The remaining 10 articles underwent a comprehensive full-text eligibility assessment. Among them, 5 studies were excluded due to inappropriate exercise interventions (e.g., eccentric and concentric combined exercise treatment), while 2 studies lacked relevant outcome parameters (e.g., rectus abdominis thickness), leaving 3 applicable studies (Figure 1). Detailed information on the included and excluded studies is shown in the Supplementary File S2.

Methodological quality assessment

A total of three RCT studies were included (25–27). The average quality score on the RCT scale was 8 points, with individual scores of 7, 8, and 10 points for each study, respectively. One study was ranked as high-quality (25), and two studies were raked as medium-quality (26, 27). Subject allocation was concealed by one study (25). All three studies performed blinding of assessors, but only one study met the criteria of blinding therapists (25). Blinding of subjects was marked "no" in all the studies due to the nature of exercise treatments. Two studies adequately described the complete follow-up (26, 27), and two studies retained the same number of subjects until the post-measurement (25, 26). However, only one study was marked as conducting appropriate statistical analysis (25) because the other studies did not consider a power analysis (26, 27). The detailed contents of the quality assessment are seen in Table 2.

Characteristics of included studies

The total number of subjects was 126, comprised of 35 (27.8%) females and 63 (50%) males, with 28 (22.2%) unknown cases. Among that, the subjects of SMT groups were 56, comprised of 15 (26.8%) females and 30 (53.6%) males, with 11 (19.6%) unknown sexes. While the mean age of all subjects was 42.3 years, ranging from 41 ± 5.9 years (26) to 47 ± 14.7 years (27), the mean age of the SMT subjects was 45 years, ranging from 35 ± 6.7 years (26) to 47 ± 14.7 years (27). The mean duration of symptoms was 20 months, ranging from 7.9 ± 6.8 months (26) to 41 ± 55.9 (27) months, and the mean duration of symptoms in SMT subjects was 18.7 months, ranging from 17.3 ± 18.7 (26) to 20 ± 25.4 (27) months. One study did not state the symptom duration (25). Two studies included mid-portion AT (26, 27), whereas one study included insertion and mid-portion AT (25). The characteristics of the included studies are summarized in Table 3.

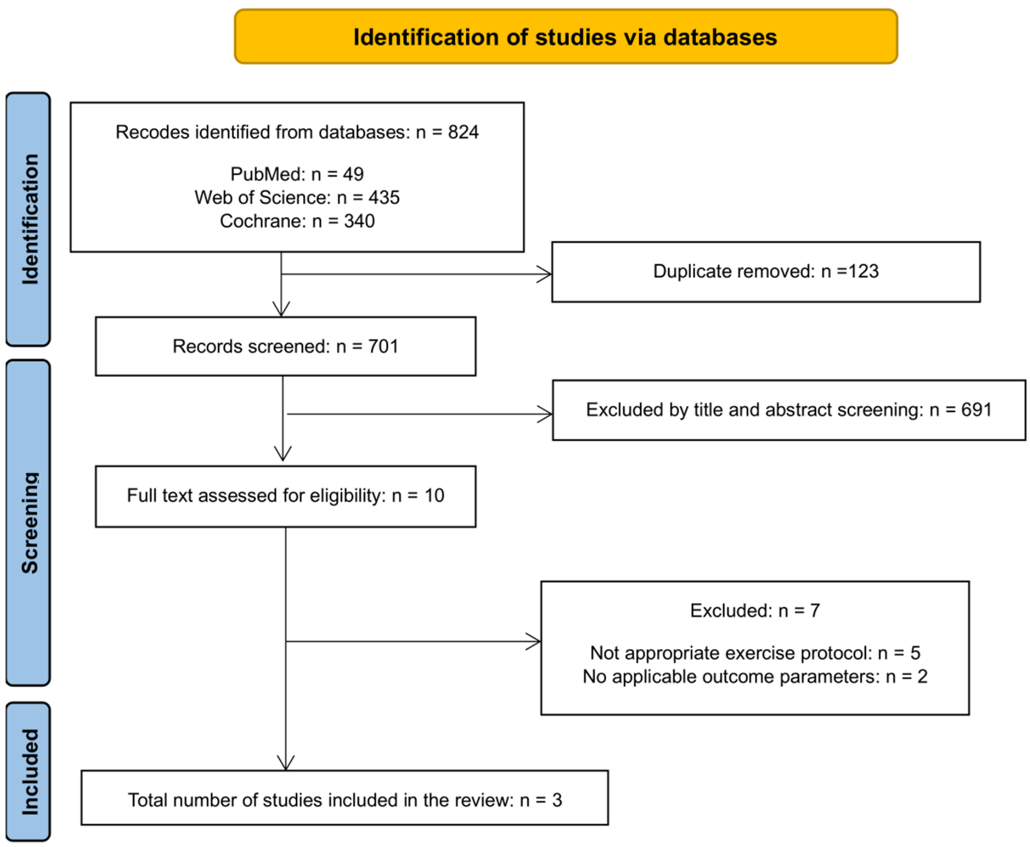

FIGURE 1
Study selection flowchart according to the PRISMA statements.

TABLE 2 Joanna briggs institute (JBI) randomized controlled trial studies tool.

Study (year)	1	2	3	4	5	6	7	8	9	10	11	12	13	Total score	Quality assessment
Silbernagel et al. (27)	Y	N	Y	N	N	Y	Y	Y	N	Y	Y	N	Y	8	Medium
Mayer et al. (26)	N	N	Y	N	N	Y	Y	Y	Y	Y	N	N	Y	7	Medium
Horstmann et al. (25)	Y	Y	Y	N	Y	Y	Y	N	Y	Y	N	Y	Y	10	High

1. Was true randomization used for assignment of participants to treatment groups?
2. Was allocation to treatment groups concealed?
3. Were treatment groups similar at the baseline?
4. Were participants blind to treatment assignment?
5. Were those delivering treatment blind to treatment assignment?
6. Were outcomes assessors blind to treatment assignment?
7. Were treatment groups treated identically other than the intervention of interest?
8. Was follow up complete and if not, were differences between groups in terms of their follow up adequately described and analyzed?
9. Were participants analyzed in the groups to which they were randomized?
10. Were outcomes measured in the same way for treatment groups?
11. Were outcomes measured in a reliable way?
12. Was appropriate statistical analysis used?
13. Was the trial design appropriate, and any deviations from the standard RCT design (individual randomization, parallel groups) accounted for in the conduct and analysis of the trial?

Characteristics of the interventions

The first study by Silbernagel et al. compared a group performed the Silbernagel protocol (balance training with isometric, concentric, and eccentric training) to a group that underwent concentric training for 12 weeks (27). The second study by Mayer et al. compared a group undergoing physiotherapy (balance and stabilization training combined with deep-friction massage, local pulsed ultrasound, and icing) to a second group wearing custom fitted insoles and a third group of untreated controls after a period of 4 weeks (26). The third study by Horstmann et al. compared a group that performed WBVT (with intermittent concentric and eccentric loadings) to a second group performing eccentric training and a third group of controls who maintained their recreational activities for 12 weeks (25). A detailed overview of the interventions is summarized in the Supplementary File S3.

Changes of functional outcomes

Strength, performance, and range of motion

The study by Mayer et al. using balance plus stabilization training showed a significant increase in plantarflexion peak torque at 4 weeks (26), whereas the study by Horstmann et al. examining WBVT reported no significant increases in concentric and eccentric plantarflexion peak torques at 20°/s and 60°/s after 12 weeks (25). Regarding dorsiflexion, both studies showed no significant increases (25, 26). The study by Silbernagel et al. using balance training showed significant increases in countermovement jump (+8%) and in the number of toe-raises (+20%) at 12 weeks (27). Conversely, no significant changes in plantarflexion ROM (+1%) were seen (27).

Changes of pain outcomes

Balance training led to a significant decrease in VAS measured at the most painful site (−29%) at 12 weeks (27). The study with WBVT reported significant decreases at osseous insertion (−57.9%) and 2 cm proximal to insertion (−51%), but not at musculotendinous junction (+2.4%) at 12 weeks (25). Pain relief was found to be present during daily activities by use of balance plus stabilization training and WBVT at 4 and 12 weeks, respectively (25, 26). In addition, Mayer et al. reported no increase in pain index after a treadmill run and a strength test (26).

Discussion

The purpose of this systematic review was to synthesize and appraise the results of SMT (i.e., balance, stabilization, proprioception, or vibration training) on functional and clinical outcomes of pain in people with AT. The search yielded three eligible studies investigating the effects of SMT as a co-intervention to high loading strategies (e.g., in addition to eccentric training). Conflicting results were observed for strength outcomes on plantar flexion torque, with positive short-term effects following balance plus stabilization training and no effects after WBVT. Improvements in performance outcomes (countermovement jump and number of toe-raises) were found in one study, while there were no relevant effects on plantarflexion ROM detectable. All studies reported reduced clinical pain outcomes following different SMT regimens.

Functional outcomes of SMT

This review indicates improvements in performance and strength outcomes following different SMT regimens, while ROM was unaffected. Specifically, in terms of strength outcomes, one study that applied balance plus stabilization training showed a significant increase in plantarflexion peak torque (26), while another study using the WBVT reported insignificant increases in concentric and eccentric plantarflexion peak torques at 20°/s and 60°/s (25).

TABLE 3 Characteristics of the included studies.

Study (year)	Deign	Intervention	Sample size (Female/male)	Mean age in years	Symptom duration in months	Outcomes	Results
Silbernagel et al. (27)	RCT	A:Silbernagel protocol B: CT	22 (5/17) 18 (4/14)	47 ± 14.7 41 ± 10.2	20 ± 25.4 41 ± 55.9	CMJ height (cm)	**A**; 13 ± 7 vs. 14 ± 7.9 ($P < 0.05$, +8%), **B**; 6 vs. 16 ± 2.9 ($P < 0.05$, +6.7%), Significantly increased in both groups
						Plantarflexion ROM (degrees)	**A**; 72 ± 6.9 vs. 73 ± 5 (+1%), **B**; 73 ± 6.6 vs. 72 ± 5.7 (−1.4%)
						Toe-raise test (numbers)	**A**; 20 ± 11 vs. 24 ± 10.8 ($P < 0.05$, +20%), **B**; 22 ± 11 vs. 28 ± 14.9 ($P < 0.05$, +27.3%), Significantly increased in both groups
						VAS (palpation pain)	**A**; 49 ± 26.2^{a} vs. 35 ± 24.8^{a} ($P < 0.05$, −29%), **B**; 27 ± 21.5^{a} vs. 31 ± 26^{a} (+14.8%)
Mayer et al. (26)	RCT	A: Physiotherapy intervention (sensory motor training + deep-friction massage + local pulsed ultrasound + ice) B: Custom insole C: Control	11 (sex not stated) 9 (sex not stated) 8 (sex not stated)	41 ± 5.9 35 ± 6.7 38 ± 4.9	17.3 ± 18.7 13.8 ± 16.5 7.9 ± 6.8	Dorsiflexion peak torque	**A, B, C**; not increased (mean <10%)
						Plantarflex peak torque	**A, B**; Significantly increased (mean >10%), **C**; not increased (mean <10%)
						Pain Disability Index (PDI) during activities of daily life	**A, B**; Significantly Improved ($P < 0.05$), **C**; not improved
						Pain Experience Scale (PES) at pre-test, after treadmill run and strength test	**A, B**; Pain was increased up to a maximum following the tests before the intervention, whereas pain was not increased after the interventions, **C**; Pain was increased up to a maximum following the tests before and after the intervention
Horstmann et al. (25)	RCT	A: Whole-body vibration training B: ET C: Wait-and-see approach	23 (10/13) 19 (9/10) 16 (7/9)	46 ± 6.9 45.7 ± 8.5 44.4 ± 7.7	Not stated Not stated Not stated	Passive resistive torques (Nm) at 0°, 5°, 10°, 15°, 20°, and 25° during ankle dorsiflexion	**A**; Significantly increased at 10°, 15°, 20°, and 25° angles ($P < 0.05$), **B**; Significantly increased at 15°, 20°, and 25° angles ($P < 0.05$), **C**; Not increased
						Concentric plantarflexion and dorsiflexion assessment at 60°/s	**A**; Increased during plantarflexion, **B**; Increased during plantarflexion and dorsiflexion **C**; Not increased
						Concentric/eccentric plantarflexion and dorsiflexion assessment at 20°/s	**A**; Increased during concentric plantarflexion, **B**; Increased during concentric/eccentric plantarflexion and dorsiflexion **C**; Not increased
						VAS (palpation pain at osseous insertion, 2 cm proximal to insertion, and at musculotendinous junction)	**A**; Significantly improved at osseous insertion; 25.4 ± 32.3 vs. 10.7 ± 22.8 ($P < 0.05$, −57.9%), at 2 cm proximal to insertion; 72.9 ± 31.4 vs. 35.4 ± 32.1 ($P < 0.05$, −51%), but not at musculotendinous junction; 33.6 ± 31.4 vs. 34.4 ± 34.2 (+2.4%), **B**; Significantly improved at osseous insertion; 28.4 ± 31.9 vs. 5.6 ± 15.0 ($P < 0.05$, −80.3%), at 2 cm proximal to insertion; 69.0 ± 33.6 vs. 22.6 ± 27.8 ($P < 0.05$, −67.3%), and at musculotendinous junction; 47.0 ± 36.1 vs. 16.4 ± 24.1 ($P < 0.05$, −65.1%), **C**; Significantly improved at osseous insertion; 23.3 ± 30.2 vs. 11.9 ± 20.1 ($P < 0.05$, −48.9%), at 2 cm proximal to insertion; 62.6 ± 28.5 vs. 45.1 ± 31.4 ($P < 0.05$, −28%), but not at musculotendinous junction; 18.7 ± 24.3 vs. 39.3 ± 31.4 ($P < 0.05$, +110.2%)
						VAS (impact of pain on recreation, running training, other physical activities, family responsibility at home, and social activities)	**A**; Significantly improved during recreation; 27.2 ± 27.3 vs. 15.8 ± 21.3 ($P < 0.05$, −41.9%), running training; 60.2 ± 35.0 vs. 35.3 ± 34.7 ($P < 0.05$, −41.4%), other physical activities; 45.3 ± 35.0 vs. 24.4 ± 27.7 ($P < 0.05$, −46%), but not during family responsibility at home; 8.4 ± 19.7 vs. 3.8 ± 6.2 (−54.8%), and social activities; 5.8 ± 12.9 vs. 3.3 ± 7.0 (−43%),

(*Continued*)

TABLE 3 Continued

Study (year)	Deign	Intervention	Sample size (Female/male)	Mean age in years	Symptom duration in months	Outcomes	Results
							B; Significantly improved during recreation; 29.1 ± 23.0 vs. 9.4 ± 16.9 (P < 0.05, −67.7%), running training; 76.3 ± 27.3 vs. 24.7 ± 30.3 (P < 0.05, −64.1%), other physical activities; 54.1 ± 31.4 vs. 14.2 ± 21.5 (P < 0.05, −73.8%), family responsibility at home; 13.0 ± 17.7 vs. 5.5 ± 15.8 (P < 0.05, −57.7%), and social activities; 9.4 ± 15.8 vs. 1.0 ± 2.0 (P < 0.05, −89.4%), **C**; Improved but not significant during recreation; 29.7 ± 30.3 vs. 19.8 ± 26.2 (−33.3%), running training; 63.9 ± 33.6 vs. 51.0 ± 38.1 (−20.2%), other physical activities; 40.0 ± 31.5 vs. 27.2 ± 33.8 (−32%), family responsibility at home; 16.0 ± 28.2 vs. 10.2 ± 26.5 (−36.3%), and not improved during social activities; 7.2 ± 17.1 vs. 9.9 ± 16.9 (+38%)

Data is presented in mean ± standard deviation.
CT, concentric training; ET, eccentric training; RCT, randomized controlled trial; CCT, controlled clinical trial; CMJ, counter movement jump; VAS, visual analogue scale; VISA-A, Victorian Institute of Sports Assessment – Achilles; NRS, numerical rating scale.
[a]Median ± Inter Quartile Range.

In the study of Mayer et al., significant increases in strength measured by plantar flexor peak torque and pain during activities of daily living at four weeks when applying balance and stabilization exercises (physiotherapy group) has been reported (26). The extent to which the SMT is responsible for the positive short-time outcome stays unclear. Interestingly, the insole group showed similar improvements like the physiotherapy group. The authors explained the improvements by changes in neuromuscular control due to modulation of afferent input to influence neuromuscular regulation (i.e., reduced spinal inhibition) leading to an optimized dynamic joint stability and postural control, respectively (26). Clinical effectiveness of insoles is also supposed to develop by an early optimization of muscular-regulated joint stability leading to pain-relief and increased strength outcomes. Exemplarily, a combination of longitudinal arch support with rear foot stabilization led directly (compression of the peroneal tendon) or indirectly (stimulation of the proprioceptors by altering joint position) to modulation of the afferent input (28). The fact that the insole effects can be expected to be almost completed after 4 weeks (28) supports this hypothesis since it is well known that sensorimotor training effects are present within 4 weeks (29). However, the design of the study of Mayer et al. did not allow for clarification of the effect mechanism of insoles (26).

Considering the study of Silbernagel et al. improvements observed after balance plus stabilization training could be attributed to the inclusion of drop jumps and countermovement jumps, which involve plantarflexion movements (27). Conversely, the WBVT study included only a few intermittent heel-raises (25). The insignificant increases in dorsiflexion peak torque in both studies can be explained by the influence of contraction mode on training gains (30).

The outcome parameter of plantarflexion ROM showed an insignificant increase (+1%) (27). However, increased ankle ROM has been reported to be clinically relevant (31). This might be due to increased tendon compliance, potentially indicating reduced loading capacity (31). Interestingly, a previous systematic review did not recommend plantarflexion and dorsiflexion ROM as reliable outcome measures in the AT population (23). Therefore, the ankle ROM has to be interpreted with caution in general.

In the study reporting improvements in performance (countermovement jumps and number of toe- raises), participants engaged in balance training, which included 30 s of one-leg standing for 5 sets, and 5 m of toe or heel walking for 5 sets, performed from three times a day to once a day (27). To maintain balance, the neuromuscular system collaborates with the somatosensory, visual, and vestibular systems to control the body and stabilize the body's center of mass. This cooperative sensory information forms the basis of the ability to control balance (32). Thus, this type of training potentially creates new response strategies (33) by modulation of afferent input (29) that may lead to improvements in functional outcome parameters.

Pain outcomes of SMT

Significant decreases were noted at most measurement sites, including the osseous insertion and 2 cm proximal to the insertion of the Achilles tendon after WBVT (25), and at the most painful site following balance training (27). These findings underscore the potential effectiveness of additional SMT regimens in reducing pain associated with AT. However, pain at the musculotendinous junction (MTJ) did not decrease following the vibration training (25). The authors speculated that this discrepancy may be due to the unique characteristics of the MTJ (25). This region may possess histological or biomechanical properties that make it more susceptible to the effects of vibration (25). The MTJ is a critical transition zone between muscle and tendon, experiencing high mechanical stress and strain during intense activities (34). Its unique morphology, characterized by a highly folded muscle membrane infiltrated with collagen fibrils from the tendon (35), further complicates its response to vibration training. Consequently, it may be less responsive to the mechanical stimuli provided by the vibration training (36). These multifaceted factors warrant further investigation to elucidate the underlying reasons.

Pain relief is largely influenced by rapid neural changes, while functional improvements potentially require long-term physiological adaptations lasting up to six months, influenced not only by neural systems but also by various factors such as muscular systems and tendon structure (33). SMT enhances proprioception, the body's ability to sense position and movement, leading to improved coordination and postural control, which may reduce strain on the Achilles tendon during activities, thereby alleviating pain (37). In contrast, functional improvements result from long-term physiological adaptations (38) involving complex biological processes like tendon structure reorganization, and adjustments in neural and muscular systems (33).

Future studies ought to extend the duration of interventions to capture long-term physiological adaptations and include comprehensive assessments of both neural and physiological changes. Additionally, researchers should ensure larger sample sizes and standardized protocols to enhance the generalizability and consistency of findings across various treatments.

Limitations

There are limitations that should be considered. Two out of three studies did not conceal allocation to treatment groups, which might cause selection bias (26, 27). Also, the same two studies failed to blind therapists who administered the training intervention, so performance bias might have been present (26, 27). Moreover, data for training adherence was reported only in one study that implemented vibration training (25), so it was impossible to draw a concrete conclusion regarding the effect of balance and balance plus stabilization exercises that were conducted in the other two studies (26, 27). Furthermore, due to the lack of available studies that implemented SMT, its effectiveness to eccentric training was not investigated. In addition, the small number of studies included in this review and the variability among the interventions can hinder consistent conclusions and limit the generalizability of the findings.

Although eccentric exercise is recognized in literature as the gold standard conservative management for the population of AT, it is questionable whether the effect is different from other exercise therapies (39). According to the "time-under-tension" hypothesis (40), beneficial adaptations can occur with any sort of loading as long as the mechanical strain is applied adequately within the optimal range of 4.5%–6.5% (41, 42). Recently, the personalized isometric training interventions that addressed muscle-tendon imbalances showed improvements in the outcomes of the triceps surae and knee extensor muscle strength (41–43). The authors argued that muscle and tendon tissues exhibit distinct sensitivities to mechano-metabolic stimuli, with muscle adaptation being responsive to various metabolic stress, while tendon adaptation is predominantly driven by the experienced strain (43). This differential adaptation can lead to significant imbalances, causing high level tendon strain that is closely associated with tendinopathies (34, 44). The results of the studies emphasize individualized training regimens considering the balance between muscle and tendon development to mitigate injury risk (41–43). However, its effectiveness in single treatment or combined treatment modalities has to be explored.

Clinical implication

The main finding of the present systematic review is that SMT might be considered as an optional exercise treatment in addition to other co-interventions, such as eccentric training, for AT population. Clinicians may consider adding balance with stabilization components while performing eccentric training, e.g., by use of a stability pad. One-leg balance for 5 sets of 30 s and toe or heel walking for 5 sets of 5 m could be the examples as seen in the Silbernagel et al. study. Those exercises are applicable to a wide range of patients with co-morbidities and do not require much time, with a duration of approximately 10 min, including rest in between sessions. Therein implementation as co-interventions might bring meaningful results as early as 4 weeks when additionally performed to standard care two or three times a week.

Conclusion

This is the first study systematically investigating the efficacy of SMT on functional and clinical outcomes of pain in people with AT, identifying an area that needs further exploration. The search yielded three eligible studies investigating the effects of SMT as a co-intervention to high loading strategies (e.g., in addition to eccentric training). SMT, in addition to other interventions (e.g., eccentric training, physiotherapy), showed potential effects on strength outcomes in short-term and improvements in performance outcomes (i.e., countermovement jump and number of toe-raises). In addition, all the included studies with different SMT regimens reported reduced clinical pain outcomes. SMT can therefore be recommended as part of a multimodal treatment strategy protocol in patients suffering from Achilles tendinopathy. However, the current evidence is weak; its additional effectiveness to golden standard therapy (high loading protocols) should be evaluated. The small number of studies included in this review and the variability between the SMT protocols impedes the ability to draw consistent conclusions across study protocols including SMT loads and modalities used. Future studies ought to extend the duration of interventions to capture long-term physiological adaptations and include comprehensive assessments of both neural and physiological changes. Additionally, researchers should ensure larger sample sizes and standardized protocols to enhance the generalizability and consistency of findings across various treatments.

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.

Author contributions

MHK: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Resources, Validation, Visualization, Writing – original draft, Writing – review & editing. WM: Data curation, Investigation, Methodology, Resources, Validation, Writing – review & editing. AQ: Methodology, Writing – review & editing. JS: Conceptualization, Methodology, Writing – review & editing. TE: Conceptualization, Methodology, Writing – review & editing. MC: Conceptualization, Methodology, Project administration, Supervision, Writing – review & editing.

Funding

The author(s) declare financial support was received for the research, authorship, and/or publication of this article. Deutsche Forschungsgemeinschaft (DFG, German Research Foundation) — Projektnummer 491466077.

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.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2024.1414633/full#supplementary-material

References

1. Cook JL, Rio E, Purdam CR, Docking SI. Revisiting the continuum model of tendon pathology: what is its merit in clinical practice and research? *Br J Sports Med.* (2016) 50:1187–91. doi: 10.1136/bjsports-2015-095422

2. Hirschmüller A, Frey V, Deibert P, Konstantinidis L, Mayer F, Südkamp N, et al. Achilles tendon power Doppler sonography in 953 long distance runners—a cross sectional study. *Ultraschall Med.* (2010) 31(04):387–93. doi: 10.1055/s-0029-1245189

3. Rio E, Moseley L, Purdam C, Samiric T, Kidgell D, Pearce AJ, et al. The pain of tendinopathy: physiological or pathophysiological? *Sports Med.* (2014) 44(1):9–23. doi: 10.1007/s40279-013-0096-z

4. Chang YJ, Kulig K. The neuromechanical adaptations to achilles tendinosis. *J Physiol.* (2015) 593(15):3373–87. doi: 10.1113/JP270220

5. Alfredson H, Pietilä T, Jonsson P, Lorentzon R. Heavy-load eccentric calf muscle training for the treatment of chronic achilles tendinosis. *Am J Sports Med.* (1998) 26 (3):360–6. doi: 10.1177/03635465980260030301

6. Langberg H, Ellingsgaard H, Madsen T, Jansson J, Magnusson SP, Aagaard P, et al. Eccentric rehabilitation exercise increases peritendinous type I collagen synthesis in humans with achilles tendinosis. *Scand J Med Sci Sports.* (2007) 17 (1):61–6. doi: 10.1111/j.1600-0838.2006.00522.x

7. De Vos RJ, Weir A, Tol JL, Verhaar JAN, Weinans H, Van Schie HTM. No effects of PRP on ultrasonographic tendon structure and neovascularisation in chronic midportion achilles tendinopathy. *Br J Sports Med.* (2011) 45(5):387–92. doi: 10.1136/bjsm.2010.076398

8. Murphy MC, Travers MJ, Chivers P, Debenham JR, Docking SI, Rio EK, et al. Efficacy of heavy eccentric calf training for treating mid-portion achilles tendinopathy: a systematic review and meta-analysis. *Br J Sports Med.* (2019) 53 (17):1070–7. doi: 10.1136/bjsports-2018-099934

9. Wyndow N, Cowan SM, Wrigley TV, Crossley KM. Triceps surae activation is altered in male runners with achilles tendinopathy. *J Electromyogr Kinesiol.* (2013) 23(1):166–72. doi: 10.1016/j.jelekin.2012.08.010

10. Sayana MK, Maffulli N. Eccentric calf muscle training in non-athletic patients with achilles tendinopathy. *J Sci Med Sport.* (2007) 10(1):52–8. doi: 10.1016/j.jsams.2006.05.008

11. Rio E, Kidgell D, Lorimer Moseley G, Gaida J, Docking S, Purdam C, et al. Tendon neuroplastic training: changing the way we think about tendon rehabilitation: a narrative review. *Br J Sports Med.* (2016) 50(4):209–15. doi: 10.1136/bjsports-2015-095215

12. Ryan D, O'Donoghue G, Rio E, Segurado R, O'Sullivan C. The effect of combined action observation therapy with eccentric exercises in the treatment of mid-portion achilles-tendinopathy: a feasibility pilot randomised controlled trial. *BMC Sports Sci Med Rehabil.* (2022) 14(1):201. doi: 10.1186/s13102-022-00594-z

13. Carmody D, Conanan A, Moeller D, Khoblall S, Keating C. Efficacy of externally paced training on pain in tendinopathy: a systematic review and meta analysis. *Cureus.* (2023) 15(6):e39994. doi: 10.7759/cureus.39994

14. Page P. Sensorimotor training: a "global" approach for balance training. *J Bodyw Mov Ther.* (2006) 10(1):77–84. doi: 10.1016/j.jbmt.2005.04.006

15. Aman JE, Elangovan N, Yeh IL, Konczak J. The effectiveness of proprioceptive training for improving motor function: a systematic review. *Front Hum Neurosci.* (2015) 8:1075. doi: 10.3389/fnhum.2014.01075

16. Di Corrado D, Francavilla VC, La Paglia R, Parisi MC, Buscemi A, Coco M. Short-term effects of specific sensorimotor training on postural assessment in healthy individuals: a pilot study with a randomized placebo-controlled trial. *J Funct Morphol Kinesiol.* (2023) 8(2):46. doi: 10.3390/jfmk8020046

17. Han J, Waddington G, Adams R, Anson J. A proprioceptive ability factor underlying all proprioception tests? Response to tremblay (2013). *Percept Mot Skills*. (2014) 119(1):301–4. doi: 10.2466/10.23.24.PMS.119c16z2

18. Abercromby AFJ, Amonette WE, Layne CS, Mcfarlin BK, Hinman MR, Paloski WH. Variation in neuromuscular responses during acute whole-body vibration exercise. *Med Sci Sports Exerc*. (2007) 39(9):1642–50. doi: 10.1249/mss.0b013e318093f551

19. Alashram AR, Padua E, Annino G. Effects of whole-body vibration on motor impairments in patients with neurological disorders: a systematic review. *Am J Phys Med Rehabil*. (2019) 98(12):1084–98. doi: 10.1097/PHM.0000000000001252

20. Zemková E. Instability resistance training for health and performance. *J Tradit Complement Med*. (2016) 7(2):245–50. doi: 10.1016/j.jtcme.2016.05.007

21. Page MJ, McKenzie JE, Bossuyt PM, Boutron I, Hoffmann TC, Mulrow CD, et al. The PRISMA 2020 statement: an updated guideline for reporting systematic reviews. *Int J Surg*. (2021) 88:105906. doi: 10.1016/j.ijsu.2021.105906

22. Ma LL, Wang YY, Yang ZH, Huang D, Weng H, Zeng XT. Methodological quality (risk of bias) assessment tools for primary and secondary medical studies: what are they and which is better? *Mil Med Res*. (2020) 7(1):7. doi: 10.1186/s40779-020-00238-8

23. Kim M, Lin CI, Henschke J, Quarmby A, Engel T, Cassel M. Effects of exercise treatment on functional outcome parameters in mid-portion achilles tendinopathy: a systematic review. *Front Sports Act Living*. (2023) 5:1144484. doi: 10.3389/fspor.2023.1144484

24. van Tulder M, Furlan A, Bombardier C, Bouter L. Updated method guidelines for systematic reviews in the cochrane collaboration back review group. *Spine (Phila Pa 1976)*. (2003) 28(12):1290–9. doi: 10.1097/01.BRS.0000065484.95996.AF

25. Horstmann T, Jud HM, Fröhlich V, Mündermann A, Grau S. Whole-body vibration versus eccentric training or a wait-and-see approach for chronic achilles tendinopathy: a randomized clinical trial. *J Orthop Sports Phys Ther*. (2013) 43(11):794–803. doi: 10.2519/jospt.2013.4762

26. Mayer F, Hirschmüller A, Müller S, Schuberth M, Baur H. Effects of short-term treatment strategies over 4 weeks in achilles tendinopathy. *Br J Sports Med*. (2007) 41(7):e6. doi: 10.1136/bjsm.2006.031732

27. Silbernagel KG, Thomeé R, Thomeé P, Karlsson J. Eccentric overload training for patients with chronic achilles tendon pain—a randomised controlled study with reliability testing of the evaluation methods. *Scand J Med Sci Sports*. (2001) 11(4):197–206. doi: 10.1034/j.1600-0838.2001.110402.x

28. Hirschmüller A, Baur H, Müller S, Helwig P, Dickhuth HH, Mayer F. Clinical effectiveness of customised sport shoe orthoses for overuse injuries in runners: a randomised controlled study. *Br J Sports Med*. (2011) 45(12):959–65. doi: 10.1136/bjsm.2008.055830

29. Bruhn S, Kullmann N, Gollhofer A. The effects of a sensorimotor training and a strength training on postural stabilisation, maximum isometric contraction and jump performance. *Int J Sports Med*. (2004) 25(1):56–60. doi: 10.1055/s-2003-45228

30. Morrissey MC, Harman EA, Johnson MJ. Resistance training modes: specificity and effectiveness. *Med Sci Sports Exerc*. (1995) 27(5):648–60. doi: 10.1249/00005768-199505000-00006

31. Yeh CH, Calder JD, Antflick J, Bull AMJ, Kedgley AE. Maximum dorsiflexion increases achilles tendon force during exercise for midportion achilles tendinopathy. *Scand J Med Sci Sports*. (2021) 31:1674–82. doi: 10.1111/sms.13974

32. Dunsky A, Zeev A, Netz Y. Balance performance is task specific in older adults. *Biomed Res Int*. (2017) 2017:6987017. doi: 10.1155/2017/6987017

33. Rio E, Docking SI. Adaptation of the pathological tendon: you cannot trade in for a new one, but perhaps you don't need to? *Br J Sports Med*. (2018) 52(10):622–3. doi: 10.1136/bjsports-2016-097325

34. Karamanidis K, Epro G. Monitoring muscle-tendon adaptation over several years of athletic training and competition in elite track and field jumpers. *Front Physiol*. (2020) 11:607544. doi: 10.3389/fphys.2020.607544

35. Jakobsen JR, Krogsgaard MR. The myotendinous junction-A vulnerable companion in sports. A narrative review. *Front Physiol*. (2021) 12:635561. doi: 10.3389/fphys.2021.635561

36. Contreras-Hernandez I, Falla D, Schneebeli A, Martinez-Valdes E. Neuromechanical changes in achilles tendinopathy and the effects of exercise-induced mechanical tendon loading: a protocol for a systematic review. *BMJ Open*. (2022) 12(2):e050186. doi: 10.1136/bmjopen-2021-050186

37. Aicale R, Oliviero A, Maffulli N. Management of achilles and patellar tendinopathy: what we know, what we can do. *J Foot Ankle Res*. (2020) 13(1):59. doi: 10.1186/s13047-020-00418-8

38. Ahtiainen JP. Physiological and molecular adaptations to strength training. In: *Concurrent Aerobic and Strength Training* (2019). p. 51–73. doi: 10.1007/978-3-319-75547-2_5

39. Malliaras P, Barton CJ, Reeves ND, Langberg H. Achilles and patellar tendinopathy loading programmes: a systematic review comparing clinical outcomes and identifying potential mechanisms for effectiveness. *Sports Med*. (2013) 43:267–86. doi: 10.1007/s40279-013-0019-z

40. Hasani F, Haines T, Munteanu SE, Schoch P, Vicenzino B, Malliaras P. LOAD-intensity and time-under-tension of exercises for men who have achilles tendinopathy (the LOADIT trial): a randomised feasibility trial. *BMC Sports Sci Med Rehabil*. (2021) 13(1):57. doi: 10.1186/s13102-021-00279-z

41. Domroes T, Weidlich K, Bohm S, Mersmann F, Arampatzis A. Personalized tendon loading reduces muscle-tendon imbalances in male adolescent elite athletes. *Scand J Med Sci Sports*. (2024) 34(1):e14555. doi: 10.1111/sms.14555

42. Weidlich K, Domroes T, Bohm S, Arampatzis A, Mersmann F. Addressing muscle-tendon imbalances in adult male athletes with personalized exercise prescription based on tendon strain. *Eur J Appl Physiol*. (2024). Available online at: https://pubmed.ncbi.nlm.nih.gov/38842575/

43. Lambrianides Y, Epro G, Arampatzis A, Karamanidis K. Evidence of different sensitivity of muscle and tendon to mechano-metabolic stimuli. *Scand J Med Sci Sports*. (2024) 34(5):e14638. doi: 10.1111/sms.14638

44. Pentidis N, Mersmann F, Bohm S, Schroll A, Giannakou E, Aggelousis N, et al. Development of muscle-tendon adaptation in preadolescent gymnasts and untrained peers: a 12-month longitudinal study. *Med Sci Sports Exerc*. (2021) 53(12):2565–76. doi: 10.1249/MSS.0000000000002742

frontiers | Frontiers in Sports and Active Living

TYPE Original Research
PUBLISHED 14 August 2024
DOI 10.3389/fspor.2024.1411037

OPEN ACCESS

EDITED BY
Kiros Karamanidis,
London South Bank University,
United Kingdom

REVIEWED BY
Daniel Friemert,
Koblenz University of Applied Sciences,
Germany
Julian Werth,
London South Bank University,
United Kingdom

*CORRESPONDENCE
Takahiro Higuchi
higuchit@tmu.ac.jp

RECEIVED 03 April 2024
ACCEPTED 31 July 2024
PUBLISHED 14 August 2024

CITATION
Hakamata T, Sakazaki J and Higuchi T (2024) Placing the leading limb closer to an obstacle reduces collision of the trailing limb: an investigation in a virtual environment.
Front. Sports Act. Living 6:1411037.
doi: 10.3389/fspor.2024.1411037

Placing the leading limb closer to an obstacle reduces collision of the trailing limb: an investigation in a virtual environment

Tomoki Hakamata[1,2], Juntaro Sakazaki[1,3] and Takahiro Higuchi[1]*

[1]Department of Health Promotion Science, Tokyo Metropolitan University, Tokyo, Japan, [2]Department of Rehabilitation, Kasai Central Hospital, Tokyo, Japan, [3]Division of Physical Therapy, Department of Rehabilitation Sciences, Faculty of Health Care Sciences, Chiba Prefectural University of Health Sciences, Chiba, Japan

Introduction: When walking and stepping over an obstacle of a certain height, tripping occurs more frequently with the trailing limb than the leading limb. The present study was designed to address whether collisions involving the trailing limb can be improved with experimental manipulation of the placement of the leading limb after stepping over an obstacle. We used an immersive, virtual obstacle-crossing task to ensure that the collision was not improved simply due to the experience of physical collision with an obstacle.
Methods: Fourteen young participants (12 males and 2 females, 28.7 ± 3.5 years) were required to walk and step over a virtual horizontal pole under one of four conditions. In three conditions, participants were required to place their leading foot on a square target located along their walking path after crossing the obstacle. The target was positioned so that it was relatively close to the obstacle (10 cm from the obstacle, referred to hereafter as the closer condition), at a position that would naturally be stepped on in successful trials without a collision (20 cm from the obstacle, the middle condition), or relatively far from the obstacle (40 cm from the obstacle, the farther condition). For the fourth condition, participants were free to select where they would step after stepping over the obstacle (the control condition).
Results and discussion: The results showed that the collision rate of the trailing limb was significantly lower under the closer condition than under the other three conditions. Compared to the control condition, under the closer condition the movement of the trailing limb was modified so that obstacle crossing was performed at approximately the moment when the height of the toe of the trailing limb was higher, and the walking speed was slower. These findings suggest that placing the foot of the leading limb closer to the obstacle after crossing the obstacle may ensure safe obstacle avoidance by the trailing limb.

KEYWORDS
walking, obstacle crossing, collision avoidance, virtual reality, motion analysis

Introduction

Obstacle crossing during walking is essential for preventing trip-induced falls and injuries. Previous studies have shown that tripping occurs more frequently with the trailing limb than with the leading limb (1–4). Several reasons exist for collisions being more frequent with the trailing limb; compared to the leading limb, the foot of the trailing limb is often placed closer to an obstacle before stepping over it (1, 2, 5, 6), it has a lower clearance height (7), and it moves faster at the moment of stepping over the obstacle (7). Moreover, collisions involving the trailing limb are considered to be more likely given that it is out of sight at the moment of stepping over an obstacle (2). Previous studies have suggested that the trailing limb may have a lower priority for cognitive information processing when planning movement (8), possibly because the leading limb has a higher risk of causing falling when it collides with an obstacle compared to the trailing limb (1). Therefore, it is helpful to consider how to control the trailing limb in order to ensure that collisions are avoided and how to lead people who have difficulties in avoiding collisions, such as older adults (9), in order to improve their behavior.

Even if the movement of the trailing limb could potentially be altered by direct intervention in the movement, such an approach may not be ideal if it is dependent upon prioritizing the cognitive information processing required for planning the movement of the trailing limb over that required for the leading limb. In fact such a change could increase the risk of falling or tripping with the leading limb (1). We therefore considered that intervention in the landing position of the leading limb after stepping over an obstacle could improve control of the trailing limb. In particular, we considered that placing the foot of the leading limb relatively close to an obstacle after stepping over an obstacle might be effective for avoiding tripping with the trailing limb. This is because it could lead to (a) slower walking speed due to more careful control of the leading limb during obstacle crossing in order to avoid collision with a closer obstacle, (b) the foot placement of the trailing limb being not too close to an obstacle, which could help avoid tripping during the initial phase of obstacle crossing, and (c) higher foot clearance as a result of obstacle crossing during the latter phase of crossing. Moreover, recent studies suggest that there is interaction between the left and right, or leading and trailing, limbs in the motor control system (10, 11). If so, precise control of the leading limb could facilitate more precise control of the trailing limb while avoiding the need to prioritize planning the movement of the trailing limb during cognitive information processing.

The aim of the present study was to address whether collisions involving the trailing limb could be minimized with experimental manipulation of the placement of the leading limb after stepping over an obstacle. We hypothesized that manipulating foot placement of the leading limb closer to the obstacle after obstacle crossing would be effective for avoiding collisions of the trailing limb. We tested this hypothesis with younger adults to determine whether is worth implementing in the future with older adults. We also tested this hypothesis in a virtual reality (VR) environment in an effort to create a situation in which no physical collision with an obstacle occurred. This was necessary because collisions act as powerful feedback, indicating that the behavior was unsuccessful, and increase motivation to change behavior, even if experimental manipulation under each condition is not effective.

Materials and methods

Participants

Fourteen young individuals (12 males and 2 females, mean age = 28.7 years, SD = 3.5 years) participated in the experiment. The sample sizes were determined based on similar studies (12, 13) and an *a priori* power analysis assuming repeated measures analysis of variance. We calculated the sample size based on the power analysis performed with the G* Power software package (14) using the following parameters: effect size = 0.5, signifiance threshold (α) = 0.05 and power levels (1-β) = 0.8. The effect size of 0.5 was employed based on a previous report (12). All participants had normal or corrected-to-normal vision. Their mean standing height was 171.8 cm (SD = 9.2 cm) and their limb length was 85.5 cm (SD = 4.9). The right limb was dominant in all participants. Testing was approved by the Ethics Committee of Tokyo Metropolitan University, Japan (H5-25). Written informed consent was obtained from all participants in accordance with the Ethics Committee of Tokyo Metropolitan University and the Declaration of Helsinki. Participants received a bookstore gift card as a reward for their participation.

Apparatus

The experiment was conducted in a room measuring 6.7 m × 4.9 m (Figure 1A). Participants were asked to walk for 4 m from a starting line on a walking path measuring 5.5 m long × 1.25 m wide. A desktop computer (OMEN by HP Obelisk Desktop 875-1xxx, HP, USA) was used for data collection and stimulus presentation. Participants wore a head-mounted display (HMD) (Oculus Rift S, USA) with a resolution of 1,280 pixels × 1,440 pixels per eye and a diagonal viewing angle of 111 degrees. The HMD was wired (the length of the cable was 5 m) to ensure stable communication. To reduce the feeling of being pulled by the cable while walking, the cable was suspended from a circular sling attached to the ceiling. The spatial positions of the HMD, the entire body, and obstacles were captured by 18 cameras (OQUS and MIQUS, Qualisys, Sweden) for three-dimensional motion analysis at a sampling frequency of 60 Hz. The cameras tracked a total of 50 markers to measure the location of the HMD, the wooden box used to capture the location of a virtual obstacle, poles, start and stop positions, and the participant's whole body (see Appendix A for details). Three-dimensional marker positions were streamed from the software for the motion analysis (Qualisys Track Manager) to the Unity game engine (Unity Technologies, USA) with a delay of approximately 40 ms. Visual 3D version 6 (C-Motion) was used for data processing.

A wooden box (Figure 1A) was used to represent the location of a virtual object (a long horizontal bar) that was presented in the virtual environment and to facilitate capture of the location of the virtual object for three-dimensional motion analysis (i.e.,

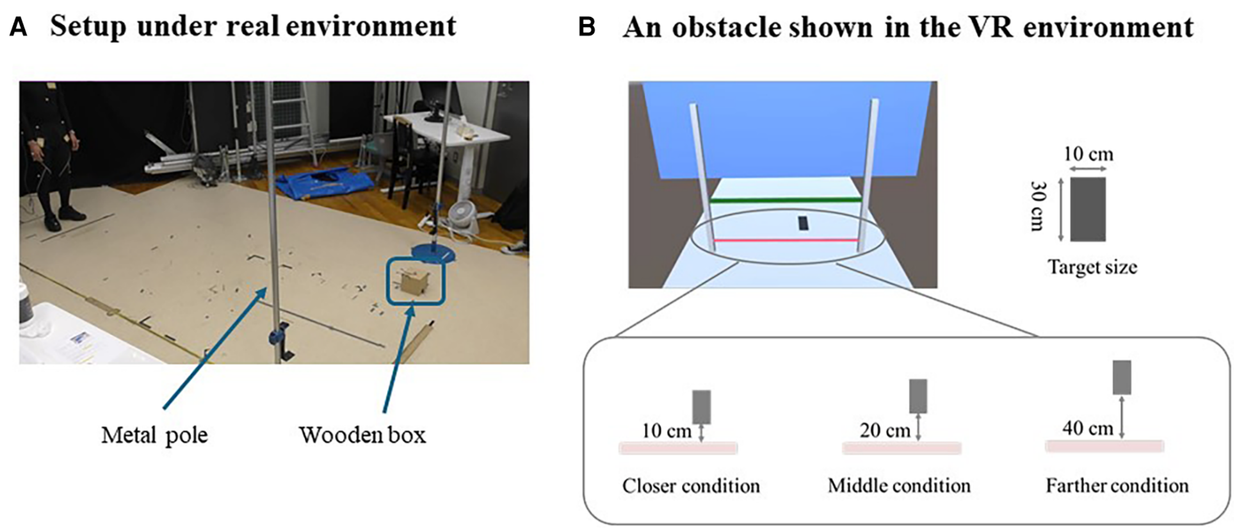

FIGURE 1
(A) experimental setup in a real-world environment. A wooden box was placed outside the walking path and was used to determine the position where the virtual object would be displayed in a virtual environment. Two vertical metal Poles were also used to indicate the obstacle height using the location of the reflective markers on the pole. (B) (Top left) an obstacle (horizontal pole) shown in a virtual environment; (Top right) A black square presented as a target on the ground; (Bottom) Three out of the four experimental conditions in which a target was presented.

since the virtual objects cannot be captured, we captured the wooden box which represented the location of the virtual bar). A total of five reflective markers were placed on the box in a noncollinear arrangement to represent the box as a rigid body. Its three-dimensional position was streamed to the Unity game engine to display the virtual bar and was used for conducting the subsequent motion analysis to test our hypothesis. The wooden box was placed outside the walking path so that physical collisions with the obstacle did not occur. Two vertical metal poles, one on each side of the walking path, were used to visually confirm the location of the obstacle, as well as to indicate the obstacle height using the location of a marker on the pole as a reference point during motion analyses.

The VR environment was set up as shown in Figure 1B. The walking path was 5.5 m long × 1.25 m wide and the two vertical metal poles indicated the location of the obstacle in the real environment. A virtual obstacle, a long horizontal pole to step over, was presented in pink and was positioned 3 m from the starting line. The height of the pole was 20% of the participant's lower-limb length, represented as the length from the greater trochanter to the plantar surface. The diameter of the horizontal pole was 2 cm. The height of the obstacle used in this study has been employed in previous studies (15–17). In a preliminary study, we confirmed that trailing limb collisions typically occurred at a height equivalent to 20% that of the lower-limb length (see Appendix B for details). The preliminary study was necessary to examine the effect of the position of the stepping mark on the collision rate. A target, consisting of a black square (30 cm long × 10 cm wide), was placed ahead of the obstacle. A green line positioned 1 m beyond the obstacle, marked the position to stop, while a red tape, located 3 m in front of the obstacle, marked the starting position for walking. Two virtual blue walls were located at each end of the walking path to ensure that participants stopped walking and did not collide with the physical walls of the room. Other than the walking path, which was colored light blue, the floor was colored gray and the surrounding area was colored light blue. These are the default colors in the Unity program. No other objects were presented in the virtual environment.

Task and procedure

The experimental task was to walk in the virtual environment and step over a virtual obstacle (a long horizontal bar) located 3 m from the walking starting position. For each trial, participants wearing an HMD stood in front of the starting position, indicated by a red line in the VR environment. After receiving a verbal instruction to start walking walk by the experimenter, participants started walking at a comfortable speed. Initially, they were asked to step over the virtual horizontal pole with their right limb while trying to avoid colliding with it. They selected which limb to use when initiating walking so as to comfortably step over the pole with their right limb. They stopped walking when they reached the green line on the walking path. We asked them to close their eyes while waiting on the green line. This was necessary to prevent VR sickness, caused by the image distortion that occurs at the end of the motion analysis in each trial. Participants opened their eyes and returned to the starting position after being requested to do so by the experimenter. No visual information about body movement was collected.

The task was performed under four experimental conditions: closer, middle, farther, and control (see Figure 1B). For three of these conditions, participants were requested to place the foot of their leading limb within a designated area (a black target) on the walkway after stepping over the obstacle. The target was located at a position that would naturally be stepped upon for successful trials without a collision (referred to as the middle condition), at a point 10 cm closer to the obstacle than the position used for the middle condition (the closer condition), or at a point 20 cm farther from the obstacle as compared to the position of the middle condition (the farther condition). For the fourth condition, no mark was placed after the obstacle, and participants were asked to step freely over the obstacle (control condition). A previous study showed that to successfully cross an obstacle in a real environment, the horizontal distance between the obstacle and the leading limb just after stepping over an obstacle averaged 30 cm (18). However, this distance may not be the same in a VR environment, given that the perception of distance differs between VR and real environments (19, 20). Therefore, we conducted a preliminary experiment to measure the average horizontal distance between the obstacle and the leading limb for successfully crossing the obstacle in our VR environment. The results showed that the average distance was 20 cm (see Appendix B for details). Based on these findings, we set the distance between the target and the obstacle.

Participants performed 10 trials for each experimental condition to give a total of 40 main trials. Prior to performing the main trials, three practice trials were conducted for each condition to allow participants to familiarize themselves with the experimental procedure. This was necessary because none of the participants were familiar with walking in a VR environment. For all conditions, except the control condition, the participants were also requested to step onto the target with their leading limb. Because this additional request also required familiarization, we decided to start the experiment with the control condition for all participants. The order of the other three conditions was randomized.

Data analysis

The following seven variables were measured to assess how participants stepped over the obstacle (Figure 2): (A) foot placement of the leading limb after stepping over the obstacle, (B) foot placement of the trailing limb before the obstacle, (C) trailing limb collision rate, (D) clearance height of the trailing limb, (E) time difference between the moment of obstacle crossing and the maximum toe height of the trailing limb, (F) walking speed at the time of obstacle crossing with the trailing limb, and (G) step length when crossing an obstacle. Since no collisions were observed involving the leading limb, we did not include the data in our statistical analysis.

The placement of the foot of the leading limb after the obstacle was defined as the horizontal distance between the heel marker of the leading limb and the obstacle at the moment of the initial step after stepping over the obstacle. This parameter was measured to address whether the experimental manipulation under the closer,

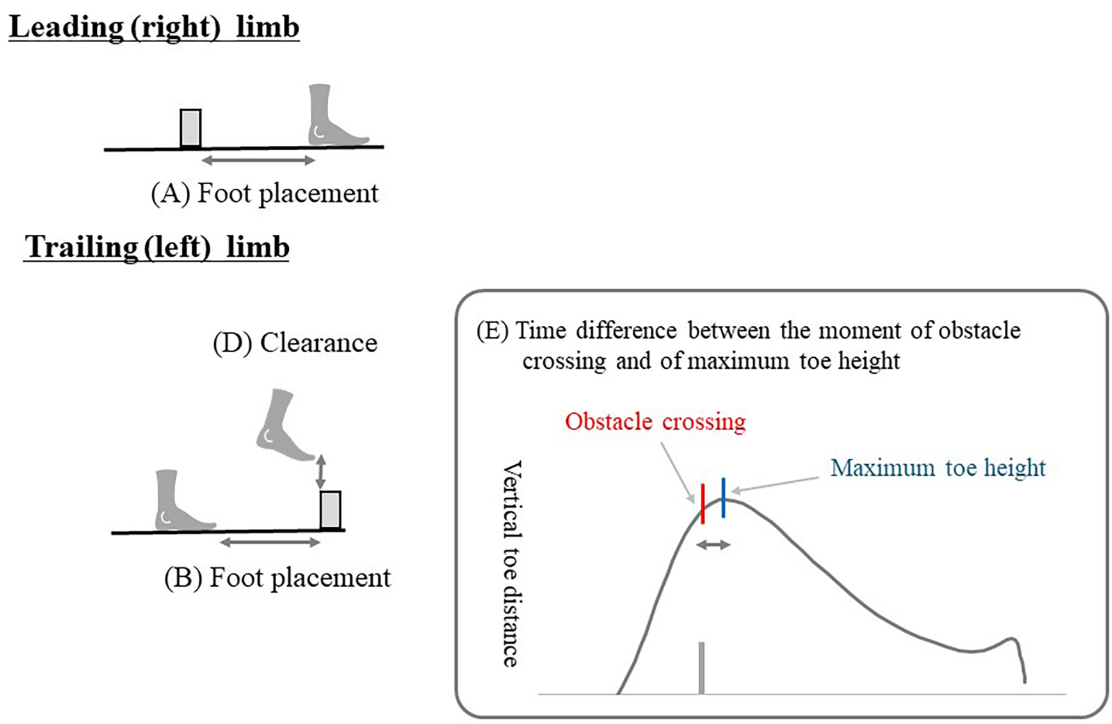

FIGURE 2
Dependent variables used to show the spatial or temporal characteristics of stepping over an obstacle. (A) foot placement of the leading limb after stepping over the obstacle, (B) foot placement of the trailing limb before the obstacle, (D) clearance height of the trailing limb, and (E) time difference between the moment of obstacle crossing and the maximum toe height of the trailing limb.

middle, and farther conditions was appropriate (i.e., 10 cm, 20 cm, and 40 cm, respectively). The foot placement of the trailing limb before the obstacle at the moment of obstacle crossing with the leading limb was defined as the horizontal distance between the second metatarsal marker of the trailing limb and the obstacle before stepping over the obstacle. The collision rate of the trailing limb was determined based on the vertical distance between the second metatarsal marker of the trailing limb and the marker on the vertical pole, which represented the height of the vertical obstacle. A collision was determined to have occurred when the distance value was below zero. The clearance height of the trailing limb was defined as the vertical distance between the second metatarsal marker of the trailing limb and the obstacle at the moment of stepping over the obstacle with the trailing limb. The time difference between the moment of obstacle crossing and of maximum toe height of the trailing limb was the difference between the time when stepping over with the trailing limb and when the toe of the trailing limb reached the maximum height. If the difference was zero, then it follows that the obstacle was stepped over at the moment that the trailing limb was raised to the highest point. Positive values of difference indicate that the trailing limb was raised to its highest point after the moment of obstacle crossing. Walking speed at the time of obstacle crossing with the trailing limb was calculated as the AP direction COM velocity at the moment the trailing limb crossed the obstacle marker. Step length was defined as the horizontal distance between the heel markers of the leading and trailing limbs at the time the leading limb landed. Prior to calculations, all three- dimensional data were processed with a Butterworth filter using a cutoff frequency of 4 Hz.

For statistical analyses, we initially conducted a one-way MANOVA (foot placement) with repeated measures for the following set of dependent variables: collision rate, walking speed, trail foot placement, clearance, and time difference. A MANOVA was conducted to explore potential interactions among variables. We then conducted a one-way repeated analysis of variance (ANOVA within-factor) to perform detailed analyses on the individual dependent variables. Since the distribution for the collision rate of the trailing limb was not normal, the data were adjusted using an arcsine transformation for statistical analysis by one-way repeated ANOVA. When a significant main effect was identified, Scheffe's post-hoc tests were carried out to estimate the significant differences. The level of significance for all analyses was set at $p < 0.05$. Notably, all of the participants started the task from the control condition. To confirm that this did not produce severe order effects, particularly on the collision rate, we sorted the data for the collision rate obtained from the four experimental conditions in the order in which they were measured, regardless of the condition of foot placement, and statistically analyzed the data using a one-way (order) ANOVA.

Results

The results of the MANOVA showed a main effect of foot placement [Wilks' $\lambda = 0.51$, F $(15,132.9) = 2.41$, $p = 0.004$, $\eta^2_P = 0.19$], suggesting that experimental manipulation of the leading foot placement significantly impacts the outcomes, despite the interdependence of the dependent variables.

The ANOVA results are as follows: the mean foot placement of the leading limb after stepping over the virtual obstacle under each experimental condition is shown in Table 1. A main effect was significant [F $(3, 39) = 101.29$, $p < 0.001$, $\eta^2_G = 0.72$]. The foot placement of the leading limb after stepping over the virtual obstacle was significantly shorter under the closer condition compared to the middle ($p < 0.001$), control ($p < 0.001$), and farther conditions ($p < 0.001$). Conversely, the foot placement was significantly longer under the farther condition than under the closer ($p < 0.001$), control ($p < 0.001$), and middle conditions ($p < 0.001$). In each condition where a target was presented, it was confirmed that the leading limb successfully landed on the target (e.g., 20.67 cm in the middle condition where the target was placed at 20 cm). Similar results were obtained as in the preliminary study (see Appendix B for details).

The mean trail foot placement before the obstacle under each experimental condition is shown in Table 1. A main effect was significant [F $(3, 39) = 11.47$, $p < 0.001$, $\eta^2_G = 0.12$]. The foot placement was significantly closer under the middle condition than that under the control condition ($p = 0.049$). The foot placement was also significantly closer under the farther

TABLE 1 Mean and standard deviation of five dependent variables under four experimental conditions.

Conditions	Closer[b]	Middle[b]	Farther[b]	Control[b]	Statistics
Foot placement of the leading limb after stepping over the obstacle (A)[a]	12.88 (4.88)	20.67 (4.58)	40.56 (8.91)	23.15 (6.46)	Cl < M, F, Co F > Cl, M, Co
Foot placement of the trailing limb before the obstacle (B)[a]	23.78 (7.98)	22.92 (7.66)	18.84 (6.24)	26.18 (7.18)	F < Cl, M, Co M < Co
Clearance height of the trailing limb (D)[a]	13.33 (9.38)	9.75 (11.18)	6.41 (9.87)	11.97 (10.83)	Cl > M, F Co > F
Walking speed at the time of obstacle crossing with the trailing limb (F)[a]	0.79 (0.15)	0.86 (0.15)	0.98 (0.12)	0.89 (0.13)	Cl < M, F, Co F > Cl, M, Co
Step length when crossing an obstacle (G)[a]	52.42 (8.66)	59.7 (9.31)	76.18 (6.9)	66.57 (8.3)	Cl < M, F, Co F > Cl, M, Co M < Co

Cl, closer; M, middle; F, farther; Co, control.

[a]Uppercase letters in parentheses correspond to the letter for dependent variables.

[b]Standard deviation in parentheses.

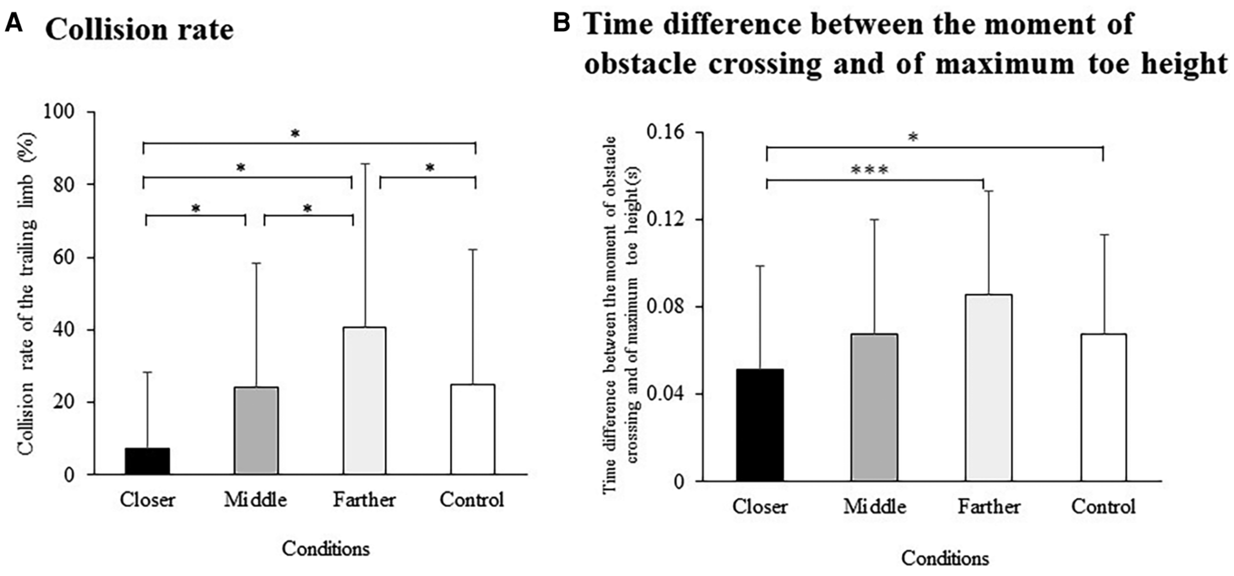

FIGURE 3
(A) mean and standard deviation of the collision rate involving the trailing limb under four experimental conditions; and (B) time difference between the moment of obstacle crossing and of maximum toe height under four experimental conditions. Significance levels are indicated by $^*p < 0.05$ and $^{***}p < 0.001$, as determined by Scheffe's *post-hoc* test.

condition than under the closer ($p = 0.003$), control ($p < 0.001$), and middle conditions ($p = 0.014$). Regarding the additional analysis to determine if an order effect influenced the collision rate results, the one-way ANOVA revealed no significant main effect of order [F (1.9, 24.72) = 0.77, $p = 0.466$, $\eta^2_G = 0.01$], suggesting that order effects were negligible.

The mean collision rate of the trailing limb under each experimental condition is shown in Figure 3A. An ANOVA applied to the data adjusted using the arcsine transformation revealed a significant main effect [F (1.68, 21.88) = 9.06, $p = 0.002$, $\eta^2_G = 0.11$]. Specifically, the collision rate was significantly lower under the closer condition compared to the middle ($p = 0.023$), control ($p = 0.039$), and farther conditions ($p = 0.023$). Conversely, the collision rate was significantly higher under the farther condition than under the closer ($p = 0.023$), control ($p = 0.039$), and middle conditions ($p = 0.039$).

The mean clearance height of the trailing limb under each experimental condition is shown in Table 1. A significant main effect was observed [F (1.93, 25.03) = 8.98, $p = 0.001$, $\eta^2_G = 0.06$], with the clearance height significantly higher under the closer condition compared to the middle ($p = 0.002$) and farther conditions ($p < 0.001$). Additionally, the clearance height was lower under the farther condition than under the control condition ($p = 0.028$).

The mean time difference between the moment the obstacle is crossed and the maximum toe height of the trailing limb is shown in Figure 3B. The ANOVA showed a significant main effect [F (2.16, 28.06) = 5.79, $p = 0.006$, $\eta^2_G = 0.06$]. This difference was significantly smaller under the closer condition compared to the control ($p = 0.047$) and farther conditions ($p < 0.001$).

The mean walking speed at the moment of crossing the obstacle with the trailing limb under each experimental condition is shown in Table 1. A significant main effect was observed [F (2.08, 27.02) = 27.59, $p < 0.001$, $\eta^2_G = 0.2$]. Walking speed was significantly slower under the closer condition compared to the middle ($p < 0.001$), control ($p < 0.001$), and farther conditions ($p < 0.001$). Walking speed was significantly faster under the farther condition compared to the closer ($p < 0.001$), control ($p = 0.003$), and middle conditions ($p < 0.001$).

The mean step length when crossing an obstacle under each experimental condition is shown in Table 1. A significant main effect was observed [F (3, 39) = 55.21, $p < 0.001$, $\eta^2_G = 0.52$]. Step length was significantly shorter under the closer condition compared to the middle ($p < 0.001$), control ($p < 0.001$), and farther conditions ($p < 0.001$). Step length was significantly longer under the farther condition compared to the closer ($p < 0.001$), control ($p < 0.001$), and middle conditions ($p < 0.001$). Step length was also shorter under the middle condition than under the control condition ($p = 0.009$).

Discussion

In this study, we examined whether experimental manipulation involving shortening of the horizontal distance between the obstacle and leading limb immediately after stepping over an obstacle would be effective for avoiding collisions of the trailing limb. The results supported the hypothesis in that the collision rate of the trailing limb was significantly lower under the closer condition than other conditions (Figure 3A). We hypothesized that experimental manipulation of the closer condition would reduce the collision rate of the trailing limb due to slower walking speed and more careful control of the leading limb, foot placement of the trailing limb not being too close to an obstacle, and higher foot clearance. Among these expectations, the present findings showed that a slower walking speed and foot placement of the trailing limb not being too close to an obstacle supported our hypothesis. Although the clearance height of the trailing limb tended to be larger in the closer condition (Table 1), this increase was not significant. Placing the leading limb closer to the obstacle effectively maintains a reasonable distance between the trailing limb and the obstacle. However, this placement also increased the risk of collisions with the trailing limb. Consequently, the leading limb may have been more carefully controlled, which resulted in a slower walking speed. Importantly, the step length was significantly shorter under the closer conditions compared to the other conditions (Table 1). This reduction in stride length contributed to the slower walking speed observed in the closer condition.

Trade-offs between speed and accuracy is a robust phenomenon in human motor performance (21, 22). Previous studies have shown that trade-offs occur in obstacle avoidance (23), as well as walking and stepping tasks (24). Previous studies have shown that decreasing walking speed is a strategy employed to increase stability in passing (25, 26). Thus, experimental manipulations that shorten the horizontal distance between the obstacle and the leading limb immediately after stepping over an obstacle could improve the accuracy of trailing limb movement. This improvement may be facilitated by strategies such as shortening the stride length or decreasing the walking speed, highlighting a trade-off between speed and accuracy.

An unexpected but interesting finding was that the time difference between the moment of obstacle crossing and the maximum toe height of the trailing limb decreased significantly under the closer condition (Figure 3B; see also Figure 2 for the meaning of the results). This suggests that, although there was no increase in the maximum toe height, the risk of collision with the trailing limb was reduced because participants crossed the obstacle when the trailing limb was at its maximum height. Based on these findings, we propose that the experimental manipulation of the leading limb placement after stepping over an obstacle resulted in temporal rather than spatial adjustments of the trailing limb trajectories.

Placing the leading limb closer to an obstacle has generally been considered to increase collision risk (27). To the best of our knowledge, no studies have specifically examined the effects of placing the leading limb closer to an obstacle. However, considering that vision is crucial for controlling the placement of the leading limb (4, 28), we considered that avoiding a collision with an obstacle is possible through careful vision monitoring, even when the leading limb is positioned close to the obstacle. Additionally, if there is an interaction between the left and right, or leading and trailing, limbs in the motor control system (10, 11), then precise control of the leading limb would influence the subsequent control of the trailing limb. Some researchers have also suggested that the trailing limb may be deprioritized in cognitive information processing during movement planning (8). For this reason, there is concern that direct manipulation of the trailing limb may not preserve its priority and collisions involving the leading limb may increase. Therefore, our aim was to safely avoid collisions involving the trailing limb while preserving the prioritization of cognitive processing by specifically manipulating the placement of the leading limb. The results of the present study show that placing the leading limb closer to the obstacle, such as approximately 10 cm away, effectively reduced the number of collisions involving the trailing limb.

This study has several limitations. First, only the immediate effect of manipulating the placement of the foot on the leading limb was examined. It may be necessary to investigate retention effects, as in a previous study (26), and also to examine the effects of long-term duration. Second, the extent to which the task, involving manipulation of the placement of the leading limb closer to the obstacle, can be generalized to real-world environments remains unknown. Collision generally occurs more frequently in the VR environment—25% under the control conditions in this study and 26% in a previous study (18)—as opposed to 0.6% in the real environment (1). Future studies are needed to test whether the experimental manipulation conducted in the VR environment of the present study would also be effective in the real-world environment. Third, collision was determined based on the vertical distance between the second metatarsal and the obstacle. Practically, even though there is a space between the two, collision would occur if individuals were wearing thick-soled shoes. This suggests that collision avoidance in practical settings involves adaptation to the constraints imposed by footwear. Future studies need to consider how to support such adaptations in order to generalize the findings of the present study. Fourth, although we set the target location at 10 cm from an obstacle, the optimal target placement could differ among individuals, particularly in older adults. Previous studies have shown that the placement of the leading limb after crossing an obstacle is not only closer to the obstacle in older participants than in younger participants (27), but it also more variable (29). Such findings suggest that individual differences in older adults, influenced by factors such as walking speed and step length, contribute to variations in the optimal placement of the target for each person. This suggests that the target location is not necessarily the same for all participants. Finally, our use of a relatively short distance between the location of the obstacle and the location of the stop affected the COM velocity because participants needed to slow down after crossing. The data representing the movement patterns of leading and trailing limbs, such as foot placement of the trailing limb before an obstacle (e.g., 18.84 cm on average under the farther condition), clearance height (6.41 cm), and foot placement of the leading limb after stepping over the obstacle (40.56 cm), were generally comparable with those reported in previous studies (30, 31). Based on the findings, we concluded the movement patterns of the leading and trailing limbs, but not movement speed, are generally likely to be preserved even in our setting, which employed a short distance to the stopping point after crossing the obstacle. Further testing is necessary to ascertain whether similar results would be observed if individuals were requested to continue walking over relatively long distances after crossing the object.

In conclusion, this study demonstrated that experimentally shortening the horizontal distance between the leading limb and the obstacle can enhance collision avoidance by the trailing limb. While placing the leading limb closer to the obstacle did increase the risk of a collision involving the leading limb, this risk did not translate into a higher collision rate, likely due to more careful control. Instead, such placement appeared to improve the collision rates of the trailing limb. Future studies are necessary to examine whether such experimental manipulation could effectively improve collision avoidance behaviors, particularly in populations prone to collisions, such as older adults (9).

Data availability statement

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

Ethics statement

The studies involving humans were approved by the Ethics Committee of Tokyo Metropolitan University. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

THa: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Resources, Software, Validation, Visualization, Writing – original draft, Writing – review & editing. JS: Formal Analysis, Writing – review & editing. THi: Conceptualization, Funding acquisition, Methodology, Project administration, Supervision, Visualization, Writing – review & editing.

Funding

The author(s) declare financial support was received for the research, authorship, and/or publication of this article.

THi was funded by the Japan Society for the Promotion of Science (Grant No. 19K11466).

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.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2024.1411037/full#supplementary-material

References

1. Heijnen MJ, Muir BC, Rietdyk S. Factors leading to obstacle contact during adaptive locomotion. *Exp Brain Res*. (2012) 223(2):219–31. doi: 10.1007/s00221-012-3253-y

2. Heijnen MJ, Romine NL, Stumpf DM, Rietdyk S. Memory-guided obstacle crossing: more failures were observed for the trail limb versus lead limb. *Exp Brain Res*. (2014) 232(7):2131–42. doi: 10.1007/s00221-014-3903-3

3. Berard JR, Vallis LA. Characteristics of single and double obstacle avoidance strategies: a comparison between adults and children. *Exp Brain Res*. (2006) 175(1):21–31. doi: 10.1007/s00221-006-0529-0

4. Rietdyk S, Rhea CK. Control of adaptive locomotion: effect of visual obstruction and visual cues in the environment. *Exp Brain Res*. (2006) 169(2):272–8. doi: 10.1007/s00221-005-0345-y

5. Chou LS, Draganich LF. Placing the trailing foot closer to an obstacle reduces flexion of the hip, knee, and ankle to increase the risk of tripping. *J Biomech*. (1998) 31(8):685–91. doi: 10.1016/s0021-9290(98)00081-5

6. Patla AE, Greig M. Any way you look at it, successful obstacle negotiation needs visually guided on-line foot placement regulation during the approach phase. *Neurosci Lett*. (2006) 397(1-2):110–4. doi: 10.1016/j.neulet.2005.12.016

7. Patla AE, Rietdyk S, Martin C, Prentice S. Locomotor patterns of the leading and the trailing limbs as solid and fragile obstacles are stepped over: some insights into the role of vision during locomotion. *J Mot Behav*. (1996) 28(1):35–47. doi: 10.1080/00222895.1996.9941731

8. Rinaldi NM, Lim J, Hamill J, Van Emmerik R, Moraes R. Walking combined with reach-to-grasp while crossing obstacles at different distances. *Gait Posture*. (2018) 65:1–7. doi: 10.1016/j.gaitpost.2018.06.167

9. Sakurai R, Kodama K, Ozawa Y, Pieruccini-Faria F, Kobayashi-Cuya KE, Ogawa S. Association of age-related cognitive and obstacle avoidance performances. *Sci Rep*. (2021) 11(1):12552. doi: 10.1038/s41598-021-91841-9

10. Miura Y, Shinya M. Foot clearance when crossing obstacles of different heights with the lead and trail limbs. *Gait Posture*. (2021) 88:155–60. doi: 10.1016/j.gaitpost.2021.05.020

11. Hagio S, Kouzaki M. Visuomotor transformation for the lead leg affects trail leg trajectories during visually guided crossing over a virtual obstacle in humans. *Front Neurosci*. (2020) 14:357. doi: 10.3389/fnins.2020.00357

12. Weber A, Friemert D, Hartmann U, Epro G, Seeley J, Werth J, et al. Obstacle avoidance training in virtual environments leads to limb-specific locomotor adaptations but not to interlimb transfer in healthy young adults. *J Biomech*. (2021) 120:110357. doi: 10.1016/j.jbiomech.2021.110357

13. Ida H, Mohapatra S, Aruin AS. Perceptual distortion in virtual reality and its impact on dynamic postural control. *Gait Posture*. (2022) 92:123–8. doi: 10.1016/j.gaitpost.2021.11.017

14. Faul F, Erdfelder E, Lang AG, Buchner A. G*power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behav Res Methods*. (2007) 39(2):175–91. doi: 10.3758/bf03193146

15. Lamoureux E, Sparrow WA, Murphy A, Newton RU. The effects of improved strength on obstacle negotiation in community-living older adults. *Gait Posture*. (2003) 17(3):273–83. doi: 10.1016/s0966-6362(02)00101-7

16. Lu TW, Chen HL, Chen SC. Comparisons of the lower limb kinematics between young and older adults when crossing obstacles of different heights. *Gait Posture*. (2006) 23(4):471–9. doi: 10.1016/j.gaitpost.2005.06.005

17. Hsu WC, Liu MW, Lu TW. Biomechanical risk factors for tripping during obstacle–crossing with the trailing limb in patients with type II diabetes mellitus. *Gait Posture*. (2016) 45:103–9. doi: 10.1016/j.gaitpost.2016.01.010

18. LoJacono CT, MacPherson RP, Kuznetsov NA, Raisbeck LD, Ross SE, Rhea CK. Obstacle crossing in a virtual environment transfers to a real environment. *J Motor Learn Dev*. (2018) 6(2):234–49. doi: 10.1123/jmld.2017-0019

19. Kelly JW, Doty TA, Ambourn M, Cherep LA. Distance perception in the oculus quest and oculus quest 2. *Front Virtual Real*. (2022) 3:850471. doi: 10.3389/frvir.2022.850471

20. Kelly JW, Donaldson LS, Sjolund LA, Freiberg JB. More than just perception-action recalibration: walking through a virtual environment causes rescaling of perceived space. *Atten Percept Psychophys*. (2013) 75(7):1473–85. doi: 10.3758/s13414-013-0503-4

21. Hsieh TY, Liu YT, Mayer-Kress G, Newell KM. The movement speed-accuracy relation in space-time. *Hum Mov Sci*. (2013) 32(1):257–69. doi: 10.1016/j.humov.2012.12.010

22. Hsieh TY, Liu YT, Newell KM. The reciprocal nature of fitts' law in space and time. *J Mot Behav*. (2023) 55(2):162–73. doi: 10.1080/00222895.2022.2137452

23. Austin GP, Garrett GE, Bohannon RW. Kinematic analysis of obstacle clearance during locomotion. *Gait Posture*. (1999) 10(2):109–20. doi: 10.1016/s0966-6362(99)00022-3

24. Viramontes C, Wu M, Acasio J, Kim J, Gordon KE. Speed impacts frontal-plane maneuver stability of individuals with incomplete spinal cord injury. *Clin Biomech (Bristol, Avon)*. (2020) 71:107–14. doi: 10.1016/j.clinbiomech.2019.09.009

25. Chou L, Kaufman S, R K, Brey R, Draganich H, F L. Motion of the whole body's center of mass when stepping over obstacles of different heights. *Gait Posture*. (2001) 13(1):17–26. doi: 10.1016/s0966-6362(00)00087-4

26. Hahn ME, Chou LS. Age-related reduction in sagittal plane center of mass motion during obstacle crossing. *J Biomech*. (2004) 37(6):837–44. doi: 10.1016/j.jbiomech.2003.11.010

27. Lowrey CR, Watson A, Vallis LA. Age-related changes in avoidance strategies when negotiating single and multiple obstacles. *Exp Brain Res*. (2007) 182(3):289–99. doi: 10.1007/s00221-007-0986-0

28. Mohagheghi AA, Moraes R, Patla AE. The effects of distant and on-line visual information on the control of approach phase and step over an obstacle during locomotion. *Exp Brain Res*. (2004) 155(4):459–68. doi: 10.1007/s00221-003-1751-7

29. Harley C, Wilkie RM, Wann JP. Stepping over obstacles: attention demands and aging. *Gait Posture*. (2009) 29(3):428–32. doi: 10.1016/j.gaitpost.2008.10.063

30. Kim A, Kretch KS, Zhou Z, Finley JM. The quality of visual information about the lower extremities influences visuomotor coordination during virtual obstacle negotiation. *J Neurophysiol*. (2018) 120(2):839–47. doi: 10.1152/jn.00931.2017

31. Kim S, Nussbaum MA, Ulman S. Impacts of using a head-worn display on gait performance during level walking and obstacle crossing. *J Electromyogr Kinesiol*. (2018) 39:142–8. doi: 10.1016/j.jelekin.2018.02.007

Appendix A

A total of 50 retro-reflective markers were used. Six markers were attached to the HMD, five markers to the wooden box, a single marker on each of the left and right vertical poles, and a single marker each at the start and stop positions along the walking path. On each participant's body, seven markers were attached to the trunk (the superior end of the sternum, the xiphoid process, the seventh cervical vertebra, the 10th thoracic vertebra, the right and left acromion, and the right scapula), 12 markers were attached at six locations to the left and right upper extremities (humerus, lateral epicondyle of the humerus, dorsal forearm, medial and lateral wrist joints, and dorsal third finger), four markers to the left and right superior anterior iliac spines and the left and right superior posterior iliac spines, and 12 markers were attached at six locations to the left and right lower extremities (lateral femur, lateral femoral epicondyle, lateral lower limb, external ankle joint, second metatarsal bone, and upper calcaneus bone).

Appendix B

We conducted a preliminary study to determine the optimal foot placement of the leading limb for crossing a virtual pole in a VR environment. Participants comprised 12 young adults (29.0 ± 4.1 years old). The task was identical to that of the control condition of the main task, i.e., to walk for 3 m at a comfortable speed and step over a virtual obstacle (a long horizontal bar with a height of 20% of the lower limb length) with the right limb. We conducted five successful trials without collision and calculated the mean foot placement of the leading limb after stepping over an obstacle for each participant from the five trials. The results showed that the average value of the foot placement of the leading limb in 12 participants was 20.3 ± 7.2 cm. The mean collision rate of the trailing limb was $5 \pm 9.0\%$, suggesting that collisions could occur under the experimental conditions. This was necessary because the purpose of our present study was to examine the effect on the collision rate of experimentally manipulating the placement of a mark to step onto. Based on the findings, we set the position of the target under the middle condition to 20 cm.

frontiers | Frontiers in Sports and Active Living

TYPE Original Research
PUBLISHED 27 September 2024
DOI 10.3389/fspor.2024.1445455

Check for updates

OPEN ACCESS

EDITED BY
Maria-Elissavet Nikolaidou,
National and Kapodistrian University of Athens, Greece

REVIEWED BY
Adamantios Arampatzis,
Humboldt University of Berlin, Germany
Sebastian Bohm,
Humboldt University of Berlin, Germany

*CORRESPONDENCE
Hans-Peter Köhler
✉ hans-peter.koehler@uni-leipzig.de

RECEIVED 07 June 2024
ACCEPTED 13 September 2024
PUBLISHED 27 September 2024

CITATION
Köhler H-P, Schödlbauer M and Witt M (2024) How the acceleration phase influences energy flow and the resulting joint moments of the throwing shoulder in the deceleration phase of the javelin throw.
Front. Sports Act. Living 6:1445455.
doi: 10.3389/fspor.2024.1445455

How the acceleration phase influences energy flow and the resulting joint moments of the throwing shoulder in the deceleration phase of the javelin throw

Hans-Peter Köhler*, Maximilian Schödlbauer and Maren Witt

Department of Biomechanics in Sports, Leipzig University, Leipzig, Germany

Introduction: The throwing motion in the javelin throw applies high loads to the musculoskeletal system of the shoulder, both in the acceleration and deceleration phases. While the loads occurring during the acceleration phase and their relationship to kinematics and energy flow have been relatively well investigated, there is a lack of studies focusing the deceleration phase. Therefore, the aim of this study is to investigate how the throwing arm is brought to rest, which resultant joint torques are placed on the shoulder and how they are influenced by the kinematics of the acceleration phase.

Methods: The throwing movement of 10 javelin throwers were recorded using a 12-infrared camera system recording at 300 Hz and 16 markers placed on the body. Joint kinematics, kinetics and energy flow were calculated between the touchdown of the rear leg and the timepoint of maximum internal rotation after release +0.1 s. Elastic net regularization regression was used to predict the joint loads in the deceleration phase using the kinematics and energy flow of the acceleration phase.

Results: The results show that a significant amount of energy is transferred back to the proximal segments, while a smaller amount of energy is absorbed. Furthermore, relationships between the kinematics and the energy flow in the acceleration phase and the loads placed on the shoulder joint in the deceleration phase, based on the elastic net regularized regression, could be established.

Discussion: The results indicate that the loads of the deceleration phase placed on the shoulder can be influenced by the kinematics of the acceleration phase. For example, an additional upper body forward tilt can help to increase the braking distance of the arm and thus contribute to a reduced joint load. Furthermore, the energy flow of the acceleration phase can be linked to joint stress. However, as previously demonstrated the generation of mechanical energy at the shoulder seems to have a negative effect on shoulder loading while the transfer can help optimize the stress. The results therefore show initial potential for optimizing movement, to reduce strain and improve injury prevention in the deceleration phase.

KEYWORDS

optimization, injury prevention, athletics, modelling, inverse dynamics

1 Introduction

Javelin throwing is a demanding activity for the throwing arm and shoulder. In Javelin throwing release velocities over 30 ms^{-1} are achieved in competition (1). To accelerate the implement, mechanical energy must be transferred through the shoulder and throwing arm, while the energy is generated by the larger proximal segments by the acceleration of the thrower and the implement in the run-up phase. The resulting loads placed on the joints of the throwing arm, which are necessary to transfer the energy through the kinetic chain exceed the requirements in baseball due to the much heavier implement, although significantly higher throwing speeds are achieved when pitching (2). After the release, the body of the athlete must be brought to rest to avoid crossing the foul line. Therefore, also the throwing arm, which is accelerated to high velocities in the acceleration phase, must be decelerated. The remaining kinetic energy must be dissipated, for which different options are possible. In general, energy can be produced or absorbed by the muscles, energy can be stored and recoiled by elastic elements of the muscle-tendon unit and the ligaments, and energy can be transferred from proximal to distal joints and vice versa via the biarticular muscles and/or gravitational and inertial forces (3). However, the deceleration motions put the muscles of the rotator-cuff, its tendons, and the capsule of the humerus of the throwing shoulder under high stress, as not only must the motion be stopped, but the humeral head must also be prevented from distraction. This phase was attributed to tensile failure and resulting rotator cuff tears due to the high loads necessary to decelerate the arm (4–6).

While the resultant joint torque placed on the shoulder have been investigated frequently during the acceleration phase, especially in baseball throwing, only little is known about the demands of the deceleration/follow-through phase. The kinematic variables that the athlete must deal with have been particularly well researched. For instance, shoulder internal rotation velocities up to 8,000°/s, reached shortly after release and linear velocities of the hand near the release speed of the implement must be decelerated in baseball pitching (7, 8). In javelin throwing, the (angular) velocities that have to be dealt with are lower due to the higher mass of the implement and the associated lower release speeds (9). Therefore, the remaining energy of the segments, which must be dissipated, should also be lower. In baseball, Fleisig et al. (10) summarized the resultant joint torques placed on the shoulder in the deceleration phase, which the muscles of a joint have to balance and are a necessary basis for energy transfer and absorption, with up to 83 ± 26 Nm for shoulder adduction, 97 ± 25 Nm for shoulder horizontal abduction (extension) and 7 ± 5 Nm for external shoulder rotation. To the best of our knowledge, this overview is complete as the focus in recent years has been largely on the requirements of the acceleration phase. Which factors influences the resultant joint torques at the shoulder in decelerating motions is unknown. Only Solomito et al. (11) have calculated the influence of kinematics, i.e., the elbow angle on the resultant joint torques of the elbow valgus torque in the deceleration phase. They concluded that greater flexion angles of the elbow during pitching raises the resultant joint torques placed on the elbow in the deceleration phase. It must therefore be assumed for javelin throwing that the kinematics of the acceleration phase influence the resultant joint torques placed on the shoulder in the deceleration phase, as different joint angles lead to different lever arms and moments of inertia.

However, not only the kinematics and kinetics are of interest, so is also the energy flow (EF) between the segments of the throwing arm caused by gravitational and inertial forces. Due to the requirement of high rates of energy transfer and large amount of mechanical energy transferred to the distal segment in the acceleration phase, higher release speeds also mean that segments must be accelerated to higher velocities (2, 9). Therefore, higher speeds must also be decelerated, and the remaining energy dissipated. But how this is done has not yet been thoroughly examined. Only Wasserberger et al. (12) investigated the EF of the deceleration phase in baseball pitching by calculating the EF due to resultant moments and torques or in other words due to gravitational and inertial forces. They showed, that large amounts of energy are transferred proximally through the elbow (168 ± 72 J) and shoulder (129 ± 61 J) joint, when decelerating the throwing arm. They were also able to show that the shoulder absorbs a significant amount of mechanical energy (79 ± 36 J). The technique of analyzing the EF due to gravitational and inertial forces and torques used by Wasserberger et al. (12) has recently become very popular. This technique can be used to study the EF between adjacent segments and enables the calculation of the transfer, generation, and absorption of mechanical energy at the connecting joint. Furthermore, it is also possible to determine the contributions of rotational and linear kinetics which arise from resultant joint torques and forces, respectively (13). EF analysis has been used in different sports like tennis (14), baseball (15, 16), table tennis (17) and javelin throwing (2) for different regions of the body. The various sports have shown that EF analysis enables the investigation of mechanical patterns and thus expands the understanding of the movements in the kinematic chain.

While the focus of studies has mostly been on the acceleration phase, the deceleration phase has been much less frequently studied. Furthermore, the influence of kinematics and energy flow in the acceleration phase on the resultant joint torque and energy flow in the deceleration phase has not been investigated to date in javelin throwing. Therefore, the aim of the study was to investigate (i) how the remaining mechanical energy of the throwing arm is dissipated through gravitational and inertial forces via the shoulder, (ii) which resultant joint torques are placed on the shoulder and (iii) how the resultant joint loads at the shoulder are influenced by the kinematics and energy flow of the acceleration phase.

2 Methods

2.1 Participants

Ten right-handed javelin throwers (body height: 189.2 ± 7.2 cm; body mass: 92.4 ± 9.3 kg; age: 21.8 ± 3.6 years; personal best: 78.23 ± 11.38 m) participated in the study. At the timepoint

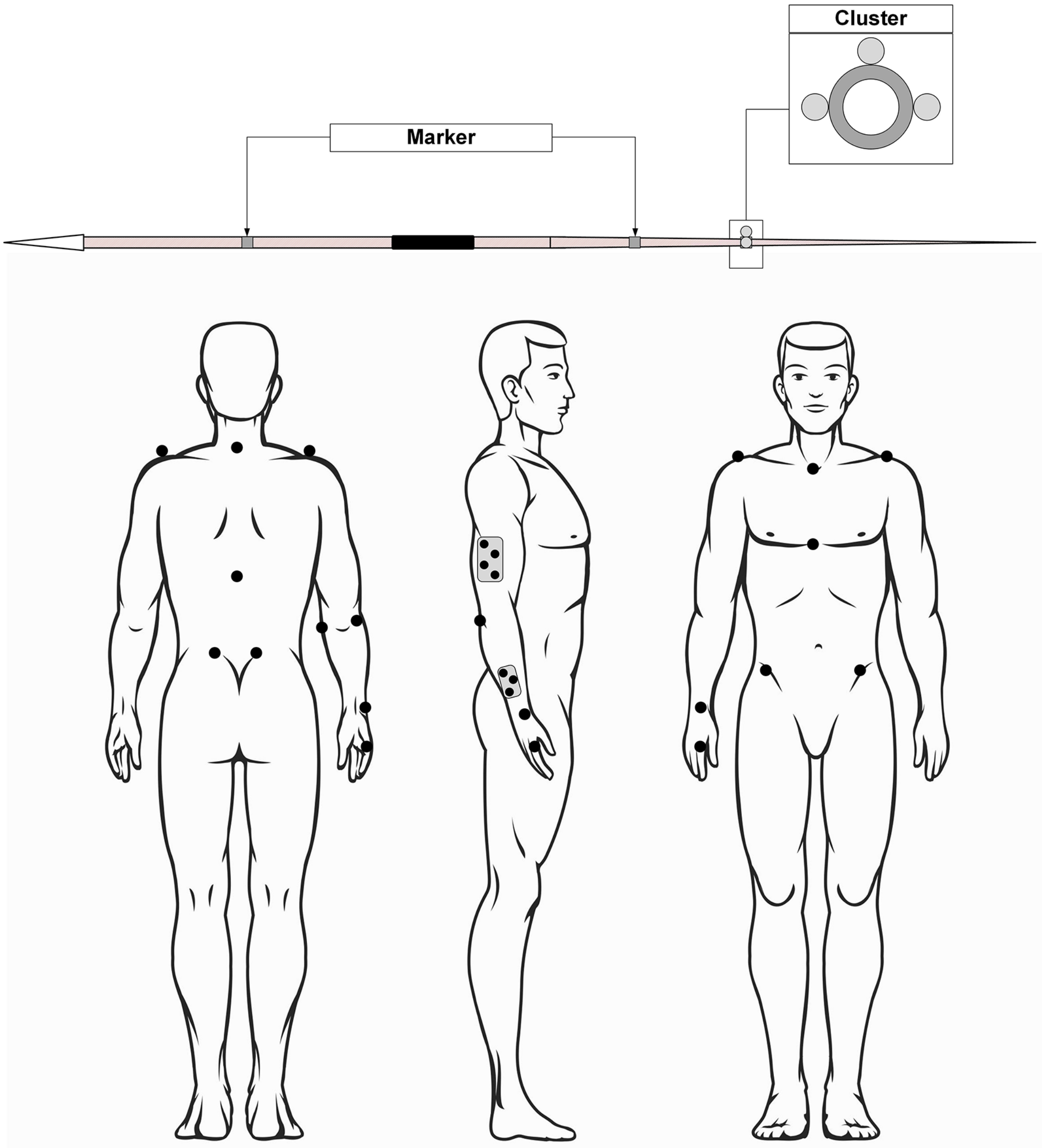

FIGURE 1
Marker placement.

of the investigation, all athletes were free from injury. Leipzig University Ethics Committee approved the investigation (ethical approval nr: 462/18-EK). Prior to the investigation, all participants gave written informed consent to participate in the study. The study was conducted according to the declaration of Helsinki.

2.2 Material and experimental protocol

Sixteen markers (metacarpophalangeal joint of the 2nd and 5th finger; ulnar and radial styloid; lateral and medial epicondyle of the humerus; left and right acromion; 7th cervical vertebrae and 12th thoracic vertebrae; processus xiphoideus; incisura jugularis; left and right spina iliaca posterior superior; left and right spina iliaca anterior superior) and two clusters (upper arm, forearm) were placed on anatomical landmarks off each subject in order to record the movements of the thrower's torso and upper extremities (Figure 1). The javelin (GETRA Kinetic, 800 g, 70 m) had five markers attached to it. It was modified for indoor use by replacing the sharp metal tip with a dull carbon one. As part of the indoor investigation, the athletes threw the javelin into a safety net.

Twelve infrared cameras (Qualisys AB, Gothenburg, Sweden) were used to record the three-dimensional location data of the markers at 300 Hz. Furthermore, the throws were recorded at 150 Hz by two perpendicular video cameras (Qualisys AB, Gothenburg, Sweden). Half of the infrared cameras were positioned on either side of the approach, thus forming an oval with the camera system positioned around 10 m in front and 2 m behind the foul line. One video camera was placed orthogonally to the approach, approximately two meters in front of the foul line. The second video camera recorded the athletes from behind, approximately 10 m behind the foul line. The calibrations average residual was 0.75 mm.

Following the warm-up routine of each athlete (approximately 30–45 min), every participant executed a minimum of three trials from their favored approach (average approach speed: 5.05 ± 0.62 ms^{-1}). The javelin's release speed (v_0) was used to choose the best three throws of each athlete for further analysis.

2.3 Data processing

Three crucial events were identified from the recorded video data prior to additional data analysis: (1) the touchdown of the rear leg, (2) the touchdown of the bracing leg, and (3) the javelin's release (Figure 2). In order to also consider the deceleration and follow-through phase, the time period of analysis was first set from the touchdown of the rear leg to 100 frames after release of the javelin A fourth order, zero-lag Butterworth filter was then used to filter the marker trajectories. Residual analysis was used to identify the cut-off frequencies (8–11 Hz) for each marker (18).

The kinematics and kinetics were calculated in Visual3D (Ver. 2024.03.1; C- motion, Germantown, USA) using a five-segment model of the javelin, right hand, right forearm, right upper arm, and thorax. While the wrist and elbow joint centers were determined as midpoints between the ulnar and radial styloid, and the medial and lateral humeral epicondyles respectively, the shoulder joint center was determined using the functional methods proposed by Schwartz and Rozumalski (19) and implemented into Visual 3D. Joint angles of the shoulder and elbow joint were calculated using Euler-/Cardan-sequences proposed by the International Society of Biomechanics (20). The position of the thorax in space was calculated via Cardan-sequence (ZYX) with respect to the laboratory coordinate system. Angular velocities and joint angular velocities were calculated as time derivatives of the respective rotation matrices.

The resultant joint forces (RJF) and torques (RJT) were calculated as external torques and forces by the top-down approach using Newton-Euler equations of motion and inverted to express them as internal torques and forces. This analysis is based on the assumption that the joint torque is generated by the muscles alone and that no translation is possible within the joint (21). De Leva's (22) body segment inertia parameters were used for invers-dynamic calculation, the center of mass (CoM) and the moments of inertia of the javelin were estimated with a reaction board and torsion pendulum (23). In order to calculate the kinetics before and after the release (REL), the computation was done twice. While the first included the javelin, the implement was removed for the second pass. All further data processing was done using custom written MATLAB (Ver. 23.2.0.25; The Mathworks Inc., Natick, MA, USA) script. The data of both modeling runs were connected at REL. Therefore, REL was determined more accurately by first calculating the center of mass of the javelin in each frame from its relative position to the attached markers. In the second step, the acceleration of the javelin's CoM was calculated as the second derivative of its position. REL was then determined as the last frame after the peak acceleration, where the Euclidean norm of the acceleration was greater than zero (24). Afterwards, the timepoint of the maximum internal rotation (tMIR) after REL was identified and the time range of analysis was set to tMIR + 0.1 s (12). As the joint angular velocities and RJT and RJF had been calculated in the global coordinate system, they were afterwards rotated into orthogonal coordinate systems, as proposed by Fleisig et al. (10).

From the calculated kinematics and kinetics, the EF between segments due to gravitational and inertial forces was computed by a segmental power analysis for all segments (hand, forearm, upper arm) at the connecting joints (wrist, elbow, shoulder). For the proximal and distal segments of the joints, the segment torque power (STP) and joint force power (JFP) were calculated as:

$$\mathrm{STP} = \mathbf{T}_{ij}\dot{\boldsymbol{\theta}}_{ij}$$

$$\mathrm{JFP} = \boldsymbol{F}_{ij}\boldsymbol{v}_{j},$$

where $\mathbf{T}_{ij}$ and $\mathbf{F}_{ij}$ denote the RJT and RJF vector of the i^{th} segment of the j^{th} joint, respectively. $\dot{\boldsymbol{\theta}}_{ij}$ denotes the angular velocity of the i^{th} segment of the j^{th} joint and $\boldsymbol{v}_j$ denotes the linear velocity of the j^{th} joint. While $\dot{\boldsymbol{\theta}}$ is not necessarily the same for both segments

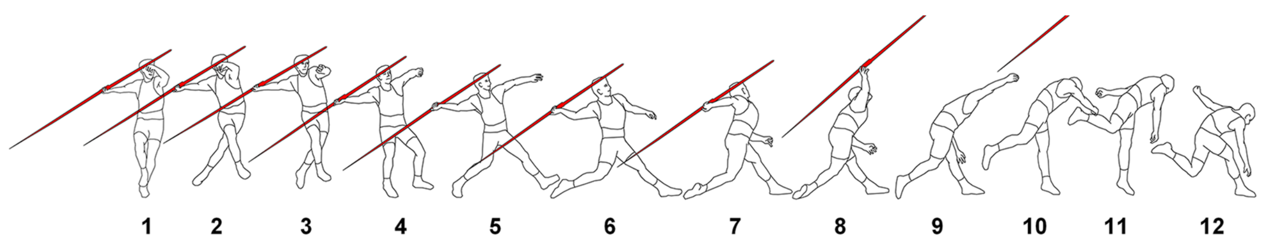

FIGURE 2
Time interval of a javelin throw from the push-off to the impulse step to the end of the deceleration movement. Other key points are the touchdown of the rear leg (4), touchdown of the bracing leg (6) and the release of the javelin (8).

TABLE 1 Calculation of the transfer, generation and absorption of mechanical energy depending on the magnitude and sign of the STP of both segments of a joint (2, 12, 21).

		Generation	Absorption	Transfer
Same sign				
Both positive		To proximal segment at $T_p\dot{\theta}_p$ To distal segment at $T_d\dot{\theta}_d$	0	0
Both negative		0	From proximal segment at $T_p\dot{\theta}_p$ From distal segment at $T_d\dot{\theta}_d$	0
Opposit sign				
$\lvert STP_p \rvert$ >	$\lvert STP_d \rvert$			
+	-	To proximal segment at $T_p(\dot{\theta}_p$-$\dot{\theta}_d)$		To proximal segment at $T_d\dot{\theta}_d$
-	+	0	From proximal segment at $T_p(\dot{\theta}_p$-$\dot{\theta}_d)$	To distal segment at $T_d\dot{\theta}_d$
$\lvert STP_p \rvert$ <	$\lvert STP_d \rvert$			
+	-		From distal segment at $T_d(\dot{\theta}_d$-$\dot{\theta}_p)$	To proximal segment at $T_p\dot{\theta}_p$
-	+	To distal segment at $T_d(\dot{\theta}_d$-$\dot{\theta}_p)$	0	To distal segment at $T_p\dot{\theta}_p$

STP_p, segment torque power of the proximal segment; STP_d, segment torque power of the distal segment; T_p, proximal joint torque vector; T_d, distal joint torque vector; $\dot{\theta}_p$, angular velocity vector of the proximal segment; $\dot{\theta}_d$, angular velocity vector of the distal segment.

connected by a joint, $\boldsymbol{v}$ is equal for both segments of a joint. Therefore, the JFP represents the rate of energy loss of one segment whose magnitude is equal to the rate of energy gain of the second segment at the same joint. In contrast, the STP reflects more than the rate of transfer of mechanical energy. Due to the different angular velocities of the segments, STP also contains mechanical energy generation and absorption (21). Table 1 shows how energy is transferred, absorbed, or generated at both segments depending on the STP. From the segmental power analysis, the net rate of energy transfer, rate of energy generation and energy absorption were calculated as outlined bin Table 1. The net rate of energy transfer was calculated as the sum of JFP and the part of STP, which reflects the rate of energy transfer. The resulting power-time curves were then integrated over time for the acceleration phase (until REL) and deceleration phase (after REL) in order to calculate the mechanical energy that was transferred, generated, or absorbed.

To quantify the demands placed on the shoulder by the resultant joint torques in the deceleration phase, the peak shoulder external rotation torque, the peak shoulder horizontal extension torque and the peak shoulder adduction torque were identified after REL.

Variables of interest, which presumably could influence resultant joint torques were identified in the acceleration phase along with the release velocity. The variables analyzed are segments and joints that are in direct close range to the shoulder joint and thus can have a direct influence on it. Therefore, the forward lean of the thorax (the angle between the frontal plane of the thorax and vertical axis of the global coordinate system), the shoulder external rotation, the shoulder horizontal extension and elbow angle were calculated at REL. Furthermore, the maximum thorax angular velocity about its longitudinal and sagittal axis, maximum shoulder internal rotation and horizontal flexion velocities, and the maximum elbow extension velocity were calculated. Besides the kinematics, the energy flow in the acceleration was quantified. Therefore, the kinetic energy of the javelin at release, the peak rate of energy transfer from proximal to distal, the peak rate of energy generation at the shoulder, the amount of energy transferred from proximal to distal and the amount of energy generated at the shoulder were computed.

2.4 Statistics

To investigate the influence of the kinematics and energy flow during the acceleration phase, specifically at the timepoint of release, on the resultant joint torques of the shoulder during the deceleration phase, regularized regression models were fitted. As the normalization of the kinetic variables by body mass or body mass*body height could lead to distortions in prediction, we decided to not normalize our kinetic data and instead to include mass and height as predictor variables (25). To eliminate magnitude influences on regressor shrinkage due to different measurement scales, predictor variables were standardized. To combine the advantages of ridge regression and the least absolute shrinkage and selection operator (LASSO), the Elastic Net adjustment for regularized regression was used (26). As the focus lied on variable selection, the alpha-coefficient was set to 0.75. Each regression was cross-validated 10 times over 100 λ-values, which controlled severity of the penalty in regularized regression (26, 27). For each peak resultant joint torque in the deceleration phase two regularized regression models were calculated, whereby each dependent variable was attempted to be explained by the kinematic or EF variables. The model that best predicts the dependent variable was chosen using the smallest mean squared error (MSE) which was calculated for the model of each λ-value. This ensured the best accuracy for each model (26). The unstandardized regression coefficients were reported for the standardized predictors.

3 Results

The mean release velocity of the investigated athletes reached $v_0 = 21.48 \pm 1.23\ \mathrm{ms}^{-1}$. The movement of the throwing arm reached a mean of $8.94 \pm 10.32°$ external rotation and a mean of

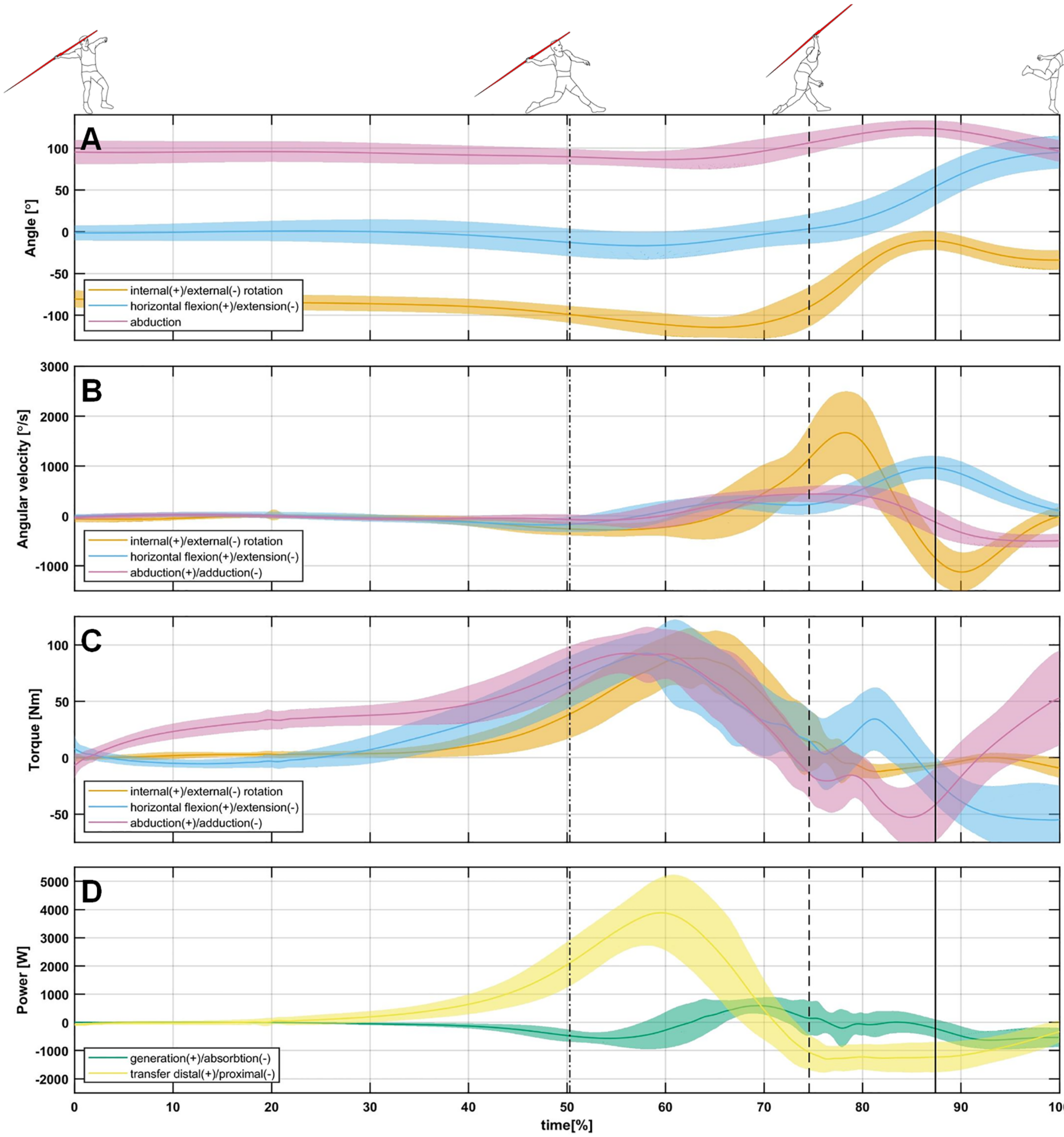

FIGURE 3
Time series of the shoulder joint angles (A), shoulder joint angular velocities (B), shoulder joint torques (C), and rates of energy transfer and generation and absorption at the shoulder (D) from the touchdown of the rear leg to the timepoint of maximal internal rotation + 0.1 s, as relative time. The dash-dotted line indicates the touchdown of the bracing leg, the dashed line indicates the release, and the continuous line represents the timepoint of maximum internal rotation.

82.97 ± 6.39° horizontal flexion in the deceleration (Figure 3A). In this phase the external joint loads peaked at a mean shoulder horizontal extension torque of 66.56 ± 32.54 Nm, while the mean peak shoulder external rotation torque reached 23.40 ± 8.48 Nm (Figure 3C). The mean peak adduction torque was calculated with a value of 63.95 ± 19.11 Nm. The mean peak rate of energy absorption at the shoulder reached 924 ± 383 W, while energy was transferred with a mean peak rate of 1,621 ± 413 W from distal to proximal (Figure 3D). Thereby, a mean energy of 49.6 ± 17.4 J was absorbed at the shoulder, while a mean energy of 166.5 ± 53.1 J was transferred from distal to proximal.

The regularized regression models were able to predict the resultant joint torques placed on the shoulder based on kinematics (Table 2) and EF (Table 3). While the shoulder

TABLE 2 Mean and standard deviation (Std) of the kinematic predictors of the acceleration phase, determined with elastic Net regularized regression, as well as the unstandardized regression coefficients for the models predicting peak shoulder external rotation torque (T^{ER}), peak shoulder horizontal extension torque (T^{hExt}), and the peak shoulder adduction torque (T^{ADD}).

			Mean Std	T^{ER}	T^{hExt}	T^{ADD}
	Intercept			−18.82	28.67	−101.16
	Mass	[kg]	93.34 ± 10.3	0.078		0.316
	Height	[m]	1.89 ± 0.1		3.121	13.609
	Release speed	[ms^{-1}]	21.48 ± 1.2		1.319	3.599
Angle at release	Thorax forward tilt	[°]	1.54 ± 15.1	−0.033		0.370
	Shoulder external rotation	[°]	92.98 ± 12.1	−0.175	−0.011	
	Shoulder hor. Flexion	[°]	1.99 ± 14.4	0.116	0.283	−0.024
	Elbow flexion	[°]	29.05 ± 6.0			
Maximum angular velocity	Shoulder internal rotation	[°/s]	2,183 ± 504.6	0.004		
	Shoulder horizontal flexion	[°/s]	1,055 ± 162.6	0.007		0.009
	Elbow extension	[°/s]	308 ± 144.3	0.006		0.011
	Thorax rotational	[°/s]	823 ± 139.0		−0.011	−0.007
	Thorax forward tilt	[°/s]	275 ± 84.4		0.042	0.088

external rotation and horizontal extension torque could be predicted by kinematics and energy flow, the shoulder adduction could only be predicted by kinematics. The number of predictors varied between 6 and 9 (kinematics) and 4–6 (energy flow).

4 Discussion

The aim of the study was to investigate (i) how the remaining mechanical energy of the throwing arm is dissipated through gravitational and inertial forces via the shoulder, (ii) which resultant joint torques are placed on the shoulder and (iii) how resultant joint torques at the shoulder are influenced by the kinematics and energy flow of the acceleration phase. This study is therefore the first study trying to establish a relation between the parameters of the release phase and the resultant joint torques on the shoulder in the deceleration phase.

The results show that a large part of the energy that must be dissipated from the throwing arm is transferred back to the upper body through the shoulder. Interestingly, the return of energy begins even before the javelin has left the hand (Figure 3D). The amount of energy that was returned is more than the values reported by Wasserberger et al. (12), even though the athletes had release velocities 10 m/s^{-1} higher than the throwers in this study. However, it must be taken into account that the athletes investigated by Wasserberger et al. (12) had significantly less mass (74.1 ± 4.2 kg) than the athletes examined here. Relative to body mass, the energy transferred backwards is comparable between both studies. This results in 1.74 ± 0.82 J/kg and 1.8 ± 0.57 J/kg between pitching and javelin throwing, respectively. While these values are comparable, it must be considered that the javelin throwers return an equal amount of energy mainly due to a higher mass of the arm, while the baseball players achieve these amounts of energy mainly due to higher speeds. When one considers the amount of energy absorbed, this ratio changes, the baseballers absorb significantly higher amounts of energy in total (79 ± 36 J vs. 49.6 ± 17.4 J) and therefore also relative to body mass (1.01 ± 0.45 J/kg vs. 0.53 ± 0.19 J/kg). Due to the same relative amount of energy that is returned proximally, it can be hypothesized that the amount of energy returned in a given time interval is limited and thus the athletes in baseball have to compensate by absorbing energy (28). However, one could also assume that the javelin throwers have better muscular stabilization of the shoulder due to their higher training age and are therefore better able to redirect the remaining energy. In this context, Barfield et al. (29) have shown that athletes with a more muscularly secured shoulder exhibit higher rates of energy transfer in the deceleration phase. Since the absorption of mechanical energy requires eccentric contraction, but the transfer of mechanical energy does not, the transfer of energy backwards can be considered less stressful. It has also been shown for the acceleration phase that the transfer of mechanical energy is less stressful than its generation (2, 13). However, this is yet to be examined in more detail. When comparing energy transfer of the deceleration to the acceleration phase (2), a significant lower transfer of energy can be noted. But it must be remembered, that (a) a large amount is transferred to the implement and therefore does not have to be transferred backwards and (b) at the end of the analyzed time period the arm is still not at rest and thus contains kinetic energy that must be dissipated.

TABLE 3 Mean and standard deviation of the energy flow predictors from the acceleration phase determined with elastic Net regularized regression, as well as the unstandardized regression coefficients for the models predicting peak shoulder external rotation torque (T^{ER}), peak shoulder horizontal extension torque (T^{hExt}) and the peak shoulder adduction torque (T^{ADD}).

		Mean Std	T^{ER}	T^{hExt}	T^{ADD}
Intercept			−283.07	17.01	61.70
Mass	[kg]	93.3 ± 10	0.40		
Height	[m]	1.9 ± 0	121.03	17.41	
$P^{P\to D}$	[W]	4,466 ± 1,071	−0.01		
P^{gen}	[W]	785 ± 168	0.02	−0.01	
$E^{P\to D}$	[J]	392 ± 73			
E^{gen}	[J]	32 ± 9	0.23	−0.89	
E^{Kin}	[J]	193.4 ± 22	0.25	0.27	

$P^{P\to D}$ peak rate of energy transfer from proximal to distal; P^{gen} peak rate of energy generation; $E^{P\to D}$ energy transferred from proximal to distal; E^{gen} energy generated; E^{Kin} kinetic energy of the javelin at release.

Compared to the resultant joint moments in baseball, the calculated RJT of this study are clearly lower (5, 10). However, the distribution of joint torques is similar, while the horizontal extension and adduction show the highest torques, the external rotation torque is well below them. While Köhler & Witt (2) calculated even higher joint torques for the acceleration phase which they primarily attributed to the higher mass of the javelin, no implement was present in the deceleration phase. The increased muscular demands could therefore be attributed to the higher speeds in baseball, which must be slowed down after release. The regression models that use the kinematics as predictors also show that the release speed is an influencing factor for the resultant joint torques in the deceleration phase. The higher the release velocity, the higher the following resultant joint torques on the shoulder. This has already been proven for the acceleration phase in baseball and javelin throwing (2, 30). However, it can also be demonstrated that the resultant joint torques on the shoulder are influenced by other kinematic variables of the acceleration phase. The external rotation and horizontal flexion of the shoulder at the time of release influence the external rotation and horizontal extension torque. The greater the external rotation, the lower the resultant joint torque, and the bigger the horizontal flexion the higher the resultant joint torque during deceleration. The greater external rotation gives athletes a longer braking path, which means they can apply less torque over a longer period and still stop the arm, wheras a higher horizontal flexion would decrease the stopping distance. Furthermore, it can be seen, at least for the external rotation torque, that a more forward tilted upper body at the time of release reduces the RJT. As the athletes must accelerate the javelin along its longitudinal axis and at the same time achieve an optimal release angle, a further forward movement of the upper body means that the athletes must remain in external rotation in order to do so. Therefore, the extended forward tilt of the upper body could influence the external rotation and thus work towards reducing the resultant joint torque during deceleration. This may imply that an increased forward tilt of the thorax would not only be a prerequisite for achieving high release speeds (2) but could also be useful in preventing injuries. However, it must also be further examined as to whether the reduction of one resultant joint torque does not result in an increase in another torque. For example, the regression models show that increasing the upper body forward tilt reduces the external rotation torque but increases the adduction torque at the same time. For the horizontal extension angle at release, an opposite behavior of both torques can be shown, as the angle increases, the external rotation torque increases and the adduction torque decreases.

Furthermore, the resultant joint torques are influenced by the peak angular velocities of the acceleration phase. Thus, an increase in angular velocities leads to an increase in the resultant joint torques in the deceleration phase as these higher velocities must be stopped. However, the high angular velocities at the shoulder are also a prerequisite for high release speed (9). On the other hand, the peak angular velocity of the upper body about its longitudinal axis can reduce resultant joint torques. As Aguinaldo & Escamilla (31, 32) reported, the rotational motion of the thorax is an important contributor to energy transfer across the shoulder. A faster rotating thorax may lead to smaller horizontal flexion angles (or higher horizontal extension angles) at release and therefore reduce the resultant joint torques, as stated for the horizontal flexion angle before. This is partly confirmed by the regression models of kinetics. A higher transfer of mechanical energy via the shoulder to the distal segment leads to a reduction of the resultant joint torques, but not for the horizontal extension torque. As it has already been proven for the acceleration phase, the generation of energy has a negative effect on the resultant joint torques (13, 14), at least in the case of external rotation, while the horizontal extension shows a reversed relationship. This correlation should be examined more closely, as it cannot be explained by the authors at this point. The kinetic energy of the javelin at the time of its release is, like the release speed of the implement, associated with an increase in the resultant joint torques in the deceleration phase. Performing more work on the implement therefore also requires more energy from the segments, which can be achieved by increasing their velocity. However, the authors would also have expected the amount of transferred energy to be a predictor of the resultant joint torques in the deceleration phase, as this is one of the most important factors for increased release speed and could therefore also influence the demands on the joint (2, 12). When comparing the resultant joint torques between the acceleration and deceleration phase one can note differences in the magnitudes of the resultant joint torques. While in the acceleration phase a broad variety of muscles is active to stop external rotation by eccentric contraction and accelerating the arm due to concentric contraction, the rotator cuff is mainly active to control humeral head positioning. In the deceleration phase the muscles of the rotator cuff have to contract to resist distraction, horizontal adduction, and internal rotation of the shoulder (33). These additional tasks and the eccentric contraction to stop the internal rotation, could possibly explain why the rotator cuff is exposed to a greater risk of injury during the deceleration phase, even though the joint torques are lower. The reduction of the resultant joint torques could by adjusting the body positioning due to technical improvement is therefore crucial to minimize the damends placed on the joint and therefore reduce the risk of injury either due to repeated stress or singular events.

5 Limitations

The following limitations should be considered when evaluating the current study's findings. First and foremost, the sample size needs to be considered. From a statistical point of view, the group size is relatively small. However, if you look at the athletes' personal bests and the fact that this is not a competition investigation, there are no comparable investigations to date. Furthermore, the study shows results that are consistent with other findings. We therefore assume that the results have a practical relevance despite the small sample size.

Second, in contrast to competition results, the release velocities were comparatively low. There could be several causes for this. (a), it is important to note that the investigation was completed several months prior to the competitive season. (b), the investigation was conducted indoors, which contrasts with the requirements of a competition.

Third, although great importance was placed on the exclusion of predictors in the regularized regression estimation with a value of $\alpha = 0.75$, the regression models still contain a relatively large number of variables. However, several relevant (practical and clinical) results could be found. Despite this, more research is needed to better understand the deceleration movement, the loads occurring in this phase and how they are linked to the acceleration phase.

Ultimately, there are several restrictions associated with motion capture and multi-body modeling. Errors can occur when calculating joint centers, due to marker motion, and the estimation of body segment inertia parameters. Nevertheless, every effort was made to reduce their impact as much as possible within the selected approaches.

6 Conclusion and perspectives

Our study is the first to investigate the resultant joint torques and energy flow in the deceleration phase of the javelin throw and how they are linked to the energy flow and kinematics of the acceleration phase. We were able to show that energy flow in the acceleration phase and the resultant joint torques in the deceleration phase are linked, but that at the same time, the demands can be altered by changed joint angles at the point of release. Therefore, it is possible to optimize the movement regarding load minimizing and performance maximizing at the same time. However, the results only represent a first approach. More studies are needed to understand the mechanisms of the kinematic chain and the underlying mechanical patterns in the acceleration and the deceleration phase and their linkage. This improved understanding could lead to better technical preparation for athletes and thus contribute to injury prevention.

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 humans were approved by Leipzig University ethics committee (ethical approval nr: 462/18-EK). The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

H-PK: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Software, Visualization, Writing – original draft, Writing – review & editing. MS: Data curation, Software, Writing – review & editing, Visualization. MW: Conceptualization, Funding acquisition, Project administration, Resources, Supervision, Writing – review & editing.

Funding

The author(s) declare financial support was received for the research, authorship, and/or publication of this article. This work was funded by the Federal Institute of Sports Science on behalf of the German Bundestag (No. ZMVI4-070801/19-20). The funders had no role in study design, data collection and analysis, decision to publish, or preparation of the manuscript.

Acknowledgments

We would like to thank Eva Böker, Simon Kiem and Florian Hallmann for their help with data recording. We would also like to express our thanks to our colleagues at the Institute for Applied Training Science for their support. Finally, we would like to thank all athletes and coaches who made this investigation possible. This publication was funded by the Open Access Publishing Fund of Leipzig University supported by the German Research Foundation within the program Open Access Publication Funding.

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.

Publisher's note

References

1. Bartonietz K. Javelin throwing: an approach to performance development. In: Zatsiorsky VM, editor. *Biomechanics in Sport. Performance Enhancement and Injury Prevention*. Oxford: Blackwell Science (2000). p. 401–34.

2. Köhler H-P, Witt M. Energy flow in men's javelin throw and its relationship to joint load and performance. *PeerJ*. (2023) 11:e16081. doi: 10.7717/peerj.16081

3. Zajac FE, Neptune RR, Kautz SA. Biomechanics and muscle coordination of human walking: part I: introduction to concepts, power transfer, dynamics and simulations. *Gait and Posture*. (2002) 16(3):215–32. doi: 10.1016/S0966-6362(02)00068-1

4. Escamilla RF, Andrews JR. Shoulder muscle recruitment patterns and related biomechanics during upper extremity sports. *Sports Med*. (2009) 39(7):569–90. doi: 10.2165/00007256-200939070-00004

5. Fleisig GS, Barrentine SW, Escamilla RF, Andrews JR. Biomechanics of overhand throwing with implications for injuries. *Sports Med*. (1996) 21(6):421–37. doi: 10.2165/00007256-199621060-00004

6. Schmitt H, Hansmann HJ, Brocai DRC, Loew M. Long term changes of the throwing arm of former elite javelin throwers. *Int J Sports Med*. (2001) 22(4):275–9. doi: 10.1055/s-2001-13814

7. Hirashima M, Kudo K, Watarai K, Ohtsuki T. Control of 3D limb dynamics in unconstrained overarm throws of different speeds performed by skilled baseball players. *J Neurophysiol*. (2007) 97(1):680–91. doi: 10.1152/jn.00348.2006

8. Oi T, Slowik JS, Diffendaffer AZ, Takagi Y, Tanaka H, Nobuhara K, et al. Biomechanical differences between Japanese and American professional baseball pitchers. *Orthop J Sports Med*. (2019) 7:1–8. doi: 10.1177/2325967119825625

9. Köhler H-P, Hepp P, Witt M. Quantifizierung der belastungen des muskel-band-apparats im schultergelenk von speerwerfern. *Obere Extremität*. (2022) 17(1):27–33. doi: 10.1007/s11678-021-00674-5

10. Fleisig GS, Andrews JR, Dillman CJ, Escamilla RF. Kinetics of baseball pitching with implications about injury mechanisms. *Am J Sports Med*. (1995) 23(2):233–9. doi: 10.1177/036354659502300218

11. Solomito MJ, Garibay EJ, Golan E, Nissen CW. Elbow flexion post ball release is associated with the elbow varus deceleration moments in baseball pitching. *Sports Biomechanics*. (2021) 20(3):370–9. doi: 10.1080/14763141.2018.1560495

12. Wasserberger KW, Giordano KA, de Swart A, Barfield JW, Oliver GD. Energy generation, absorption, and transfer at the shoulder and elbow in youth baseball pitchers. *Sports Biomech*. (2021) 23:1–16. doi: 10.1080/14763141.2021.1933158

13. Howenstein J, Kipp K, Sabick MB. Energy flow analysis to investigate youth pitching velocity and efficiency. *Med Sci Sports Exerc*. (2019) 51(3):523–31. doi: 10.1249/MSS.0000000000001813

14. Martin C, Bideau B, Bideau N, Nicolas G, Delamarche P, Kulpa R. Energy flow analysis during the tennis serve: comparison between injured and noninjured tennis players. *Am J Sports Med*. (2014) 42(11):2751–60. doi: 10.1177/0363546514547173

15. de Swart AFMJ, van Trigt B, Wasserberger K, Hoozemans MJM, Veeger DHEJ, Oliver GD. Energy flow through the lower extremities in high school baseball pitching. *Sports Biomech*. (2022):1–15. doi: 10.1080/14763141.2022.2129430

16. Pryhoda MK, Sabick MB. Lower body energy generation, absorption, and transfer in youth baseball pitchers. *Front Sports Act Living*. (2022) 4(4):1–15. doi: 10.3389/fspor.2022.975107

17. Iino Y, Kojima T. Mechanical energy generation and transfer in the racket arm during table tennis topspin backhands. *Sports Biomech*. (2016) 15(2):180–97. doi: 10.1080/14763141.2016.1159722

18. Winter DA. *Biomechanics and Motor Control of Human Movement*. 4th ed. Hoboken, NJ: John Wiley & Sons (2009).

19. Schwartz MH, Rozumalski A. A new method for estimating joint parameters from motion data. *J Biomech*. (2005) 38(1):107–16. doi: 10.1016/j.jbiomech.2004.03.009

20. Wu G, Van Der Helm FCT, Veeger HEJ, Makhsous M, Van Roy P, Anglin C, et al. ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion - part II: shoulder, elbow, wrist and hand. *J Biomech*. (2005) 38(5):981–92. doi: 10.1016/j.jbiomech.2004.05.042

21. Robertson DGE, Winter DA. Mechanical energy generation, absorption and transfer amongst segments during walking. *J Biomech*. (1980) 13(10):845–54. doi: 10.1016/0021-9290(80)90172-4

22. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech*. (1996) 29(9):1223–30. doi: 10.1016/0021-9290(95)00178-6

23. Sommerfeld A. *Mechanics of Deformable Bodies. Lectures on Theoretical Physics Vol. II*. New York, NY: Academic Press, Inc (1950).

24. Köhler H.-P. (2020). *Untersuchung zur Quantifizierung des Energieeintrags und -transfers innerhalb der oberen Extremitäten und deren Bedeutung für die Wettkampfleistung im Speerwurf der Männer*. Available online at: https://nbn-resolving.org/urn:nbn:de:bsz:15-qucosa2-708368 (Accessed May 6, 2020).

25. Giordano KA, Barrack AJ, Murrah WM, Karduna AR, Michener LA, Wasserberger KW. Usage and acceptability of data normalization in baseball pitching. *Sports Biomech*. (2024):1–12. doi: 10.1080/14763141.2024.2302830

26. Kipp K, Warmenhoven J. Applications of regularized regression models in sports biomechanics research. *Sports Biomech*. (2022):1–19. doi: 10.1080/14763141.2022.2151932

27. Krzyszkowski J, Kipp K. Prediction of throwing distance in the men's and women's javelin final at the 2017 IAAF world championships. *J Sports Sci*. (2021) 39(9):1055–9. doi: 10.1080/02640414.2020.1856462

28. Bartonietz K. Movement control and behaviour in upper extremity movements - the technique of javelin throwing. In: Šarabon N, editor. *Shoulder & Sports: Injuries, Prevention, Movement Control, Rehabilitation: 16th to 17th June 2006*. Šmarješke Toplice, Slovenia: Združenje rokometnih trenerjev Slovenije (2006). p. 21–9.

29. Barfield JW, Wasserberger KW, Oliver GD. Relationship between humeral energy flow during the baseball pitch and glenohumeral stability. *Int J Sports Med*. (2020) 2(3):159–63. doi: 10.1055/a-1303-4369

30. Slowik JS, Aune KT, Diffendaffer AZ, Lyle Cain E, Dugas JR, Fleisig GS. Fastball velocity and elbow-varus torque in professional baseball pitchers. *J Athl Train*. (2019) 54(3):296–301. doi: 10.4085/1062-6050-558-17

31. Aguinaldo A, Escamilla R. Segmental power analysis of sequential body motion and elbow Valgus loading during baseball pitching: comparison between professional and high school baseball players. *Orthop J Sports Med*. (2019) 7(2):232596711982792. doi: 10.1177/2325967119827924

32. Aguinaldo AL, Escamilla RF. Induced power analysis of sequential body motion and elbow valgus load during baseball pitching. *Sports Biomech*. (2022) 21(7):824–36. doi: 10.1080/14763141.2019.1696881

33. Fleisig GS, Barrentine SW, Zheng N, Escamilla RF, Andrews JR. Kinematic and kinetic comparison of baseball pitching among various levels of development. *J Biomech*. (1999) 32(12):1371–5. doi: 10.1016/S0021-9290(99)00127-X

frontiers | Frontiers in Sports and Active Living
TYPE Original Research
PUBLISHED 07 January 2025
DOI 10.3389/fspor.2024.1498333

OPEN ACCESS

EDITED BY
Adamantios Arampatzis,
Humboldt University of Berlin, Germany

REVIEWED BY
Hans-Peter Wiesinger,
Paracelsus Medical University, Austria
Carlos Cruz-Montecinos,
University of Chile, Chile

*CORRESPONDENCE
Kai-Yu Ho
✉ kaiyu.ho@unlv.edu

RECEIVED 18 September 2024
ACCEPTED 25 November 2024
PUBLISHED 07 January 2025

CITATION
Liang JN, Bashford G, Kulig K and Ho K-Y (2025) Achilles tendon morphology adaptations in chronic post-stroke hemiparesis: a comparative analysis with neurologically intact controls.
Front. Sports Act. Living 6:1498333.
doi: 10.3389/fspor.2024.1498333

Achilles tendon morphology adaptations in chronic post-stroke hemiparesis: a comparative analysis with neurologically intact controls

Jing Nong Liang[1], Greg Bashford[2], Kornelia Kulig[3] and Kai-Yu Ho[1]*

[1]Department of Physical Therapy, University of Nevada, Las Vegas, Las Vegas, NV, United States, [2]Department of Biological Systems Engineering, University of Nebraska-Lincoln, Lincoln, NE, United States, [3]Division of Biokinesiology & Physical Therapy, University of Southern California, Los Angeles, CA, United States

Introduction: In individuals with chronic post-stroke hemiparesis, slow walking speed is a significant concern related to inadequate propulsion of the paretic limb. However, an overlooked factor is this population's altered morphology of the Achilles tendon, which may compromise the propulsive forces by the paretic limb. This study aimed to explore changes in Achilles tendon morphology, including gross thickness and intra-tendinous collagen fiber bundle organization, following stroke-induced brain lesions.

Methods: Fifteen individuals with chronic post-stroke hemiparesis (at least 6 months post-stroke) and 19 neurologically intact controls participated. Ultrasound imaging was used to evaluate Achilles tendon thickness and collagen organization in the paretic and non-paretic limbs of post-stroke participants, as well as in the right limb (control limb) of the neurologically intact control group.

Results and discussion: Compared to control individuals, the paretic limb in individuals post-stroke showed increased tendon thickness at the Achilles tendon insertion and 2 cm above it. The collagen fiber bundle at the Achilles tendon insertion of the paretic limb showed reduced organization compared to that in the control limb. Individuals post-stroke also exhibited slower walking speed, and increased plantarflexor muscle tone in the paretic limb compared to controls. In conclusion, individuals with chronic post-stroke hemiparesis demonstrated tendon thickening and collagen disorganization in the paretic limb, particularly at the insertion site of the Achilles tendon, likely due to an abnormal loading environment influenced by increased plantarflexor muscle tone, muscle co-activation, and muscle disuse and atrophy. These changes may increase tendon compliance, impair force transmission and propulsion, and contribute to slower walking speed. Addressing Achilles tendon integrity should be incorporated as a component of strategies to improve neuromuscular control in this population.

KEYWORDS

Achilles tendon, post-stroke hemiparesis, walking speed, muscle tone, morphology

1 Introduction

Stroke is a leading cause of long-term disability, significantly impacting mobility and quality of life for millions of survivors. Restoring walking ability is a primary goal of post-stroke rehabilitation, as it is crucial for independence and participation in daily activities (1–3). However, despite regaining the ability to walk, many individuals continue to experience gait impairments, especially slow walking speed, even years after stroke (4, 5). This persistent slow walking speed is a major concern because it limits community reintegration and increases the risk of secondary health complications, falls, and dependence on caregivers (1, 6, 7). A critical factor contributing to slow walking speed after stroke is insufficient propulsion generated by the paretic leg (8, 9). Propulsion, the force that drives the body forward, is primarily produced by the ankle plantarflexor muscles (gastrocnemius and soleus) during the push-off phase of gait. Weakness in these muscles, altered muscle activation patterns, and increased muscle tone, are key factors underlying insufficient propulsion (10, 11). To execute effective walking, locomotor control requires the integration and coordination of descending central drive, spinal circuits, and the force transmission capabilities of muscle-tendon units. Studies suggest that the Achilles tendon may undergo significant adaptations following stroke (12–14), which may impact its ability to effectively transmit forces, thus negatively affecting propulsion and step-to-step transitions during gait.

In individuals post-stroke, the Achilles tendon of the paretic limb is exposed to a complex loading environment. Increased plantarflexor muscle tone (12, 14) and disrupted timing of agonist and antagonist muscle activity, along with inappropriate co-activation of lower limb muscles (15), contribute to sustained low-grade forces within the Achilles tendon. Despite this, force transmission from the plantarflexor muscles is compromised due to muscle atrophy and disuse in persons post-stroke (13). These abnormal forces may influence tendon morphology and mechanical properties, as tendons are highly responsive and adaptive to mechanical loading (12, 13, 16). However, existing literature presents inconsistent findings on Achilles tendon morphology, particularly concerning tendon length, size [e.g., cross-sectional area (CSA) and thickness], and mechanical properties in individuals post-stroke. Regarding tendon length, some studies reported increased tendon length in the paretic limb, potentially due to muscle contracture and higher muscle fascicle stiffness, which may cause dislocation of the muscle-tendon junction. For instance, Zhao et al. found that the paretic limb's Achilles tendon was 6% longer than the non-paretic limb (12). They suggested that as the calf muscles shorten and become stiffer post-stroke, the muscle-tendon junction shifts proximally, resulting in a lengthened tendon. In their next study, Zhao and colleagues corroborated this finding, reporting a 5% increase in tendon length in the paretic limb (13). They suggested that muscle atrophy, impaired neural control, and reduced movement post-stroke might contribute to shorter and stiffer muscle fascicles, which, in turn, exert a prolonged stretch on the tendon, leading to tendon elongation. However, other studies found no significant difference in tendon length between the paretic and non-paretic limbs. For example, Freire et al. compared Achilles tendon length in stroke survivors' paretic and non-paretic limbs with those of neurologically intact controls and found no significant differences between groups (14). Similarly, Dias et al. reported no difference in tendon length between stroke survivors' paretic and non-paretic limbs (16). These differences may stem from variations in age, activity level and/or the time since the stroke incidence among participants of these studies.

Regarding tendon size, the available literature also presents conflicting evidence. Some studies observed a reduced Achilles tendon CSA in the paretic limb compared to the non-paretic limb and neurologically intact controls. Dias et al. (16) reported an 18% decrease in Achilles tendon CSA in the paretic limb of individuals post-stroke compared to controls, while Zhao et al. (13) observed a 5% reduction in CSA in the paretic limb compared to the non-paretic limb in individuals post-stroke. The authors posit that muscle atrophy and reduced force transmission from muscle contractions to the tendon, possibly due to spasticity, might contribute to the reduced CSA. However, Zhao et al. (12) did not find a statistically significant difference in CSA between the paretic and non-paretic limbs. In our recent preliminary study (17), we observed thicker Achilles tendon at the insertion site in the paretic limbs of stroke survivors compared to control limbs of neurologically intact controls.

Several studies have investigated the mechanical properties of the Achilles tendon in the paretic limb of individuals after stroke with conflicting results. While Freire et al. (14) found no differences in tendon stiffness between paretic, non-paretic and control limbs, most studies reported decreased stiffness and Young's modulus in the paretic limb compared to both the non-paretic limb in individuals post-stroke (12, 13, 16) and the control limb of neurologically intact individuals (16). The decreased stiffness may be attributed to a reduction in collagen content, likely resulting from diminished force transmission to the tendon due to reduced muscle contraction (16). Limitations of previous studies on tendon morphology and mechanical properties include insufficient examination of intra-tendinous structure (e.g., collagen organization) and tendon thickness (13, 16), small sample sizes (17), and/or a lack of comparisons between paretic and non-paretic limbs in individuals post-stroke and those of neurologically intact controls (12, 13).

Ultrasound imaging commonly assesses gross morphological changes in diseased tendons, such as increased thickness (18–21) and CSA (22). It can also assess intra-tendinous morphology (21, 23). Bashford et al. (24) introduced spatial frequency analysis (SFA) to evaluate tendon collagen structure, represented by the "speckle pattern" on ultrasound images (20, 25). SFA can infer the structure of a group of collagen fiber bundles through frequency analysis of the speckle pattern. Briefly, the presence of higher spatial frequency content in the speckle pattern is indicative of increased organization. SFA operates in two dimensions, making it less dependent on probe angle than axial-only spectrum analysis. Among SFA parameters, peak spatial frequency radius (PSFR) is the most reported; lower PSFR indicates collagen disarray, while higher values suggest better collagen organization (24, 25). Kulig et al. (25) found healthy tendons have a mean PSFR of 2.07 mm^{-1}, while degenerated tendons average 1.55 mm^{-1}.

For these reasons, a larger-scale study is needed to examine both the intra-tendinous and gross morphology of the Achilles tendon, comparing the paretic and non-paretic limbs of individuals with stroke to those of neurologically intact controls.

A systematic review by van der Vlist et al. (26) identified nine clinical risk factors for Achilles tendinopathy, indicating that decreased plantarflexor strength and abnormal gait patterns with reduced forward propulsion are two key biomechanical contributors to the development of this condition, which are also particularly prevalent in individuals post-stroke (8). Moreover, it is important to consider that individuals post-stroke may have an increased vulnerability to insertional Achilles tendinopathy (27) due to theoretical associations with metabolic syndrome—factors hypothesized to contribute to the condition's onset (28–30). Our preliminary study involving two individuals chronically post-stroke and two neurologically intact controls revealed potential associations among increased plantarflexor muscle tone, Achilles tendon thickening, and reduced propulsive forces during gait in the paretic limb of individuals post-stroke (17). An abnormal loading environment on the Achilles tendon is thought to alter its morphology, leading to increased tendon compliance, and subsequently causing delayed and reduced force transmission during gait. These changes may lead to poorly timed and insufficient propulsion (12, 17, 31). By investigating the morphological changes of the Achilles tendon in individuals post-stroke in the chronic phase of recovery, this study aimed to contribute to a more comprehensive understanding of the factors related to gait impairments after stroke. This knowledge is essential for developing effective rehabilitation strategies to improve walking speed, mobility, and quality of life for stroke survivors.

Therefore, the primary objective of this cross-sectional study was to examine differences in the gross morphology of the Achilles tendon, explicitly focusing on thickness, as well as intra-tendinous morphology related to collagen fiber bundle organization, using ultrasound imaging between individuals with chronic post-stroke hemiparesis and neurologically intact controls. We hypothesized that compared to neurologically intact controls, individuals with chronic post-stroke hemiparesis would show alterations in Achilles tendon morphology in the paretic limb. Additionally, the secondary aim was to evaluate plantarflexor muscle tone, lower extremity function, and walking speed in stroke participants to enhance the clinical relevance of our findings. We anticipated that individuals post-stroke would exhibit increased plantarflexor muscle tone, reduced lower extremity function, and decreased walking speed.

2 Methods

2.1 Participants

Before the study, we conducted a sample size calculation, projecting an effect size (Cohen's d) of 1.0 for differences in intra-tendinous morphology (collagen fiber bundle organization) between groups with and without tendon pathology, based on findings from previous research (20). Using this estimated effect size, an alpha level of 0.05, and a power of 0.80, our analysis determined that that 28 participants (14 per group) would be required. For tendon thickness differences between stroke and control groups, we projected an effect size of 2.1 based on our previous work (17). The analysis for tendon thickness indicated that 10 participants (5 per group) would be sufficient, using the same alpha level and power. Ultimately, we selected a sample size of 28 participants, deemed adequate to detect differences in Achilles tendon morphology, specifically regarding both thickness and collagen fiber bundle organization.

Inclusion criteria for individuals with chronic post-stroke hemiparesis comprised the following: age over 18 years old, a minimum of six months after a single, unilateral, cortical, or subcortical stroke with residual hemiparesis, the ability to walk 10 meters overground independently without assistive devices, and the ability to follow cues and adhere to instructions. Inclusion criteria for neurologically intact controls were age over 18 years old, able to complete the 10-Meter Walk Test, and ability to follow cues and adhere to instructions.

Individuals post-stroke were excluded if they had a history of cerebellar stroke(s), lower extremity surgery, Achilles tendon tendinopathy (past or present), pregnancy or suspected pregnancy, had other neurological diagnoses, or had used medications to manage spasticity or received Botox injection in the lower extremity within the past six months. Control participants were excluded if they were pregnant or suspecting pregnancy, had a history of lower extremity surgery, had a history and/or current Achilles tendon tendinopathy, or had a diagnosis of any neurological disorder. Control participants were matched for sex and age, with age differences kept within 10% compared to their counterparts in the stroke group.

Participant recruitment and data collection took place between 2019 and 2022 in the Las Vegas area. Before their involvement in the study, we obtained signed informed consent from each participant. The protocol was approved by the Institutional Review Board at the University of Nevada, Las Vegas (Protocol #: 1250168).

2.2 Procedures

2.2.1 Clinical outcome measurements

Prior to tendon morphology measurements, we assessed overground walking speed using the 10-Meter Walk Test (32), lower extremity motor function using Fugl-Meyer assessment of the lower extremity (FMA-LE) (33), and plantarflexor muscle tone using Modified Ashworth Scale (MAS) (34).

Trained investigators assessed the FMA-LE for each participant within the stroke group, with 34 being the maximum possible score for lower extremity motor function (33). The 10-Meter Walk Test was administered to both the stroke-impaired and neurologically intact control groups, involving two trials of overground walking trials at self-selected speed for each participant. The average of two trials was documented to show ambulatory capacity at self-selected walking speed (32). To assess plantarflexor muscle tone, individuals post-stroke underwent MAS assessments involving passive dorsiflexion of the ankle, where zero indicated normal

10.3389/fspor.2024.1498333

muscle tone, and higher scores represented elevated spasticity and increased resistance to passive movement. A score of 4 on the MAS indicated rigidity during the assessment (34).

2.2.2 Tendon morphology measurements

Bilateral Achilles tendon morphology, including thickness and collagen fiber bundle organization, was assessed using ultrasound imaging (GE LOGIQ-e, GE Healthcare, Milwaukee, WI, USA). Participants were positioned prone on a treatment table with the knee extended and the ankle positioned in neutral (0 degree between shank and foot segments). Using a preset for ankle musculoskeletal examination at a depth of 2 cm, we employed a linear array ultrasound transducer (GE 12L-RS, bandwidth 5–13 MHz, width 38.4 mm, GE Healthcare, Milwaukee, WI, USA) to capture longitudinal images at three specific locations: the Achilles tendon insertion on the calcaneus, 2 cm above the insertion, and 4 cm above the insertion (23).

2.3 Image analysis

The thickness of the Achilles tendon at 2 cm and 4 cm above the insertion were determined as the perpendicular distance between the borders outlining the Achilles tendon at the longitudinal image's center (17, 23). For the thickness at Achilles tendon insertion, the measurement was taken as the perpendicular distance between the borders at the Achilles tendon-calcaneus intersection (Figures 1A,C) (17).

To quantify collagen fiber bundle organization (PSFR) on each longitudinal image (Figures 1B,D), SFA was performed using a custom MATLAB program, following the methodology described in previous studies (20, 21, 23). In this process, a region of interest (ROI) was selected to encompass the maximum visible tendon and peritenon area while excluding the distal and proximal image boundaries. This selection aimed to minimize potential errors arising from the curvature of the distal and proximal tendon. For each ROI, SFA was performed on all possible 32×32 pixel (~2.0 mm) sub-images ('kernels'). Briefly, 2D Fast Fourier Transforms (FFTs) were applied to each kernel after zero-padding to 128×128 samples to increase frequency sampling. A 2D high pass filter (−3 dB cut-off about 1.0 mm^{-1}) was applied to attenuate low spatial frequency artifacts. Spatial frequency parameters were extracted and averaged over all kernels of the ROI, thereby analyzing spatial frequencies in both the axial and lateral directions.

2.3.1 Image acquisition and analysis reliability

Before the study commenced, senior sonographer K.H. conducted comprehensive training for the investigators, focusing

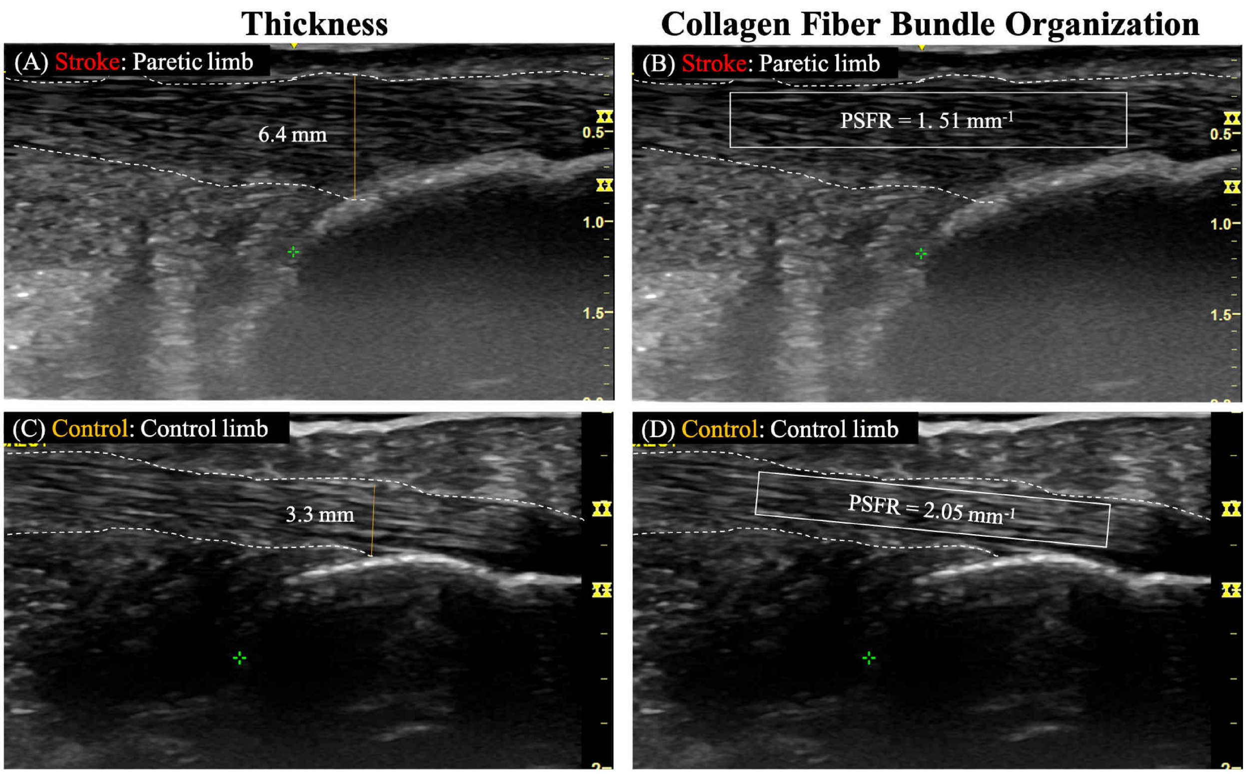

FIGURE 1
Representative ultrasound images analyzing achilles tendon thickness and collagen fiber bundle organization. Panels **(A,B)** show the paretic limb of a participant with chronic post-stroke hemiparesis, and panels **(C,D)** illustrate the control limb of a neurologically intact control participant. White dashed lines outline the tendon borders and the yellow solid line represents the tendon thickness. The solid-lined box indicates the region of interest (ROI) for spatial frequency analysis (SFA).

on image acquisition and analysis techniques for Achilles tendon thickness and PSFR. The investigators were assigned specific tasks only after demonstrating repeatability in image acquisition and analysis through a reliability study.

To assess intra-rater reliability in image acquisition, researchers A.C. and M.S. conducted two data collection sessions with 5 participants, each session separated by a one-week interval. For image analysis reliability, undertaken by investigators A.C., M.S., and T.D., measurements of images from five individuals were taken on two separate days, spaced a week apart. Intra-class correlation coefficient (ICC) and standard error of measurement (SEM) were utilized to measure intra-rater reliability for image acquisition and analysis, with ICC values categorized as poor (<0.4), fair (0.4–0.7), good (0.7–0.9), and excellent (>0.9) (35). The SEM was calculated by multiplying the standard deviation by the square root of 1 minus the reliability coefficient (23).

The investigators responsible for image acquisition demonstrated good to excellent intra-rater reliability in measuring tendon thickness (ICC = 0.852–0.997; SEM = 0.039–0.092 mm) and PSFR (ICC = 0.731–0.828; SEM = 0.056–0.062 mm^{-1}). Similarly, those involved in image analysis exhibited good to excellent intra-rater reliability for measuring tendon thickness (A.C. and M.S.; ICC = 0.808–0.999; SEM = 0.011–0.105 mm) and PSFR (A.C. and T.D.; ICC = 0.993–0.977; SEM = 0.024–0.031 mm^{-1}).

2.4 Statistical analysis

The outcome measures of this study included thickness and PSFR of the Achilles tendon. The paretic and non-paretic limbs were included for participants with post-stroke hemiparesis. The right limb was used for neurologically intact participants to represent the control limb. We conducted a Shapiro-Wilk test and the results indicated that the outcome variable data followed a normal distribution. Therefore, one-way ANOVAs and post-hoc pairwise comparisons with Bonferroni corrections were used to compare thickness and PSFR of the Achilles tendon among the three limbs (paretic, non-paretic, and control limbs). All statistical analyses were performed using SPSS software (ver. 27, International Business Machines Corp. New York, USA). A significant difference was defined as $p \leq 0.05$.

3 Results

3.1 Participant characteristics

Table 1 shows the characteristics of participants with chronic post-stroke hemiparesis and neurologically intact controls. Both groups had similar sex distribution (47% females and 53% males): the chronic post-stroke hemiparesis group comprised 7 females, while the control group had 9 females. Both groups had similar age, height, and weight. Participants with chronic post-stroke hemiparesis had a slower overground self-selected walking speed compared to control participants. On average, participants with chronic post-stroke hemiparesis had a MAS score of 0.8, suggesting slight increase in ankle plantarflexor muscle tone.

3.2 Tendon morphology

3.2.1 Thickness

ANOVA results revealed a statistically significant difference in tendon thickness both 2 cm above the insertion [$F(2, 46) = 3.64$, $p = 0.03$] and at the insertion of the Achilles tendon [$F(2, 46) = 5.68$, $p < 0.01$]. *Post hoc* analyses, when compared to the control limb showed that the paretic limb exhibited significantly greater tendon thickness at both 2 cm above insertion ($p = 0.03$) and at the insertion ($p < 0.01$). No statistically significant difference was observed in tendon thickness between the non-paretic and paretic limbs among individuals with post-stroke hemiparesis at either 2 cm above the insertion ($p = 1.00$) or at the insertion of the Achilles tendon ($p = 0.11$). Moreover, no statistically significant difference in tendon thickness was found among the three limbs at 4 cm above the insertion of the Achilles tendon [$F(2, 46) = 1.38$, $p = 0.26$] (Table 2).

3.2.2 Collagen fiber bundle organization (peak spatial frequency radius)

ANOVA results revealed a statistically significant difference in the PSFR at the insertion of the Achilles tendon [$F(2, 46) = 3.32$, $p = 0.04$]. Subsequent *post hoc* analyses showed that the paretic limb exhibited a lower PSFR at the insertion of the Achilles tendon compared to the control limb ($p = 0.04$). There was no statistically significant difference in PSFR at the insertion of the

TABLE 1 Participant characteristics.

	Participants with chronic post-stroke hemiparesis (n = 15)	Neurologically intact controls (n = 19)	p
Sex	7 females; 8 males	9 females; 10 males	NA
Age, y	59.29 (10.38), 53.70–64.80	55.32 (14.52), 47.40–62.10	0.31
Height, cm	168.70 (6.00), 153.70–183.60	169.60 (8.90), 163.60–175.60	0.84
Weight, kg	79.30 (13.20), 46.50–112.10	71.30 (15.00), 61.20–81.40	0.42
Years post-stroke	6.70 (5.40), 3.80–9.70	NA	NA
Self-selected overground walking speed, m/s	0.88 (0.18), 0.45–1.32	1.44 (0.47), 1.29–1.71	0.01*
Fugl-Meyer assessment of the lower extremity (FMA-LE)	22.57 (5.30), 18.70–25.10	NA	NA
Modified Ashworth Scale (MAS)	0.80 (0.45), 0.20–1.20	NA	NA

The data were displayed as mean (standard deviation), and 95% confidence interval.

TABLE 2 Comparisons of tendon thickness and peak spatial frequency radius (PSFR) at the achilles tendon insertion, 2 cm above the insertion, and 4 cm above the insertion between the paretic and non-paretic limbs of participants with post-stroke hemiparesis and the control limb of neurologically intact control participants.

	Control limb	Non-paretic limb	Paretic limb	Effect size	p
Thickness (mm)					
4 cm above insertion	4.58 (0.07), 4.25–4.92	5.00 (0.13), 4.29–5.70	5.10 (0.07), 4.66–5.46	0.06, 0.00–0.19	0.26
2 cm above insertion	4.37 (0.06), 4.09–4.64	4.73 (0.07), 4.37–5.09	4.93 (0.07), 4.57–5.30[a]	0.14, 0.00–0.30	0.03*
At insertion	4.08 (0.07), 3.76–4.41	4.36 (0.07), 3.96–4.75	4.92 (0.08), 4.47–5.38[a]	0.20, 0.02–0.37	<.01*
PSFR (mm^{-1})					
4 cm above insertion	1.95 (0.24), 1.83–2.06	1.81 (0.19), 1.70–1.92	1.89 (0.20), 1.78–2.00	0.07, 0.00–0.21	0.19
2 cm above insertion	1.91 (0.27), 1.78–2.04	1.87 (0.37), 1.67–2.07	1.85 (0.20), 1.74–1.96	0.01, 0.00–0.08	0.82
At insertion	1.94 (0.32), 1.78–2.09	1.78 (0.26), 1.64–1.93	1.70 (0.21), 1.59–1.82[a]	0.13, 0.00–0.29	0.04*

The data of thickness and PSFR were displayed as mean (standard deviation), and 95% confidence interval. Effect sizes, along with their 95% confidence intervals, are also provided.
[a]Denotes a significant difference from the control limb.
*Denotes a significant difference using one-way ANOVA with repeated measures.

Achilles tendon between the non-paretic and paretic limbs in individuals with post-stroke hemiparesis ($p = 1.00$). Furthermore, no statistically significant difference in PSFR was identified among the three limbs at both 2 cm [$F(2, 46) = 0.20$, $p = 0.82$] and 4 cm above the insertion of the Achilles tendon [$F(2, 46) = 1.75$, $p = 0.19$] (Table 2).

4 Discussion

This study sought to investigate the alterations in Achilles tendon morphology, with a specific focus on thickness and collagen organization, following stroke-induced brain lesions in the chronic phase of recovery. As hypothesized, individuals with chronic post-stroke hemiparesis exhibited changes in tendon morphology, including increased thickness both 2 cm above the insertion and at the insertion site of the Achilles tendon in comparison to the control limb of neurologically intact individuals. Moreover, individuals with chronic post-stroke hemiparesis demonstrated a decreased PSFR at the insertion of the Achilles tendon, suggesting lesser collagen fiber bundle organization, when compared to the control limb of neurologically intact individuals. Additionally, individuals post-stroke walked slower than neurologically intact controls. They also exhibited increased plantarflexor muscle tone and diminished lower extremity function, which aligned with our expectations.

A key observation in this study is that the primary area of tendon changes was concentrated at the insertion point of the Achilles tendon in the paretic limb of individuals with post-stroke hemiparesis. This agrees with our preliminary case study (17), where we observed increased thickness at the calcaneal insertion of the paretic Achilles tendon, compared to similar thickness between limbs in neurologically intact controls. Dias et al. examined tendon length and CSA, reporting that the paretic limb showed reduced CSA compared to the neurologically intact control limb and the non-paretic limb, which also had reduced CSA compared to the neurologically intact control limb (16). Although the exact location of these CSA measurements was not specified, the study's focus on overall mechanical properties suggested these values likely represent averages across the tendon rather than measurements at specific regions (16). Zhao et al. focused on tendon length, CSA, stiffness, Young's modulus, and hysteresis (12, 13), but like Dias et al., they primarily reported mean differences between limbs without specific regional data. Zhao et al. did, however, note that the soleus muscle-tendon junction might shift proximally in the paretic limb (12). In summary, although existing literature offers valuable insights into alterations of the Achilles tendon post-stroke, it lacks a consistent focus on specific regions, particularly the insertion point. This study is the first to examine and report findings across three distinct regional locations within the Achilles tendon in the post-stroke population. Prior studies predominantly report mean values from multiple tendon sites, limiting the ability to make direct comparisons with our findings. These observations underscore the need for future research to investigate regional variations in tendon morphology after stroke.

The time after stroke, including factors such as chronicity and variability, and spasticity levels are potentially critical factors influencing Achilles tendon properties in individuals post-stroke. Our study, with participants averaging 6.4 years post-stroke and presenting with mild plantarflexor hypertonia (MAS score of 0.8), offers a distinct perspective compared to the available research (Table 1). Dias et al. (16) included individuals who were at least one year post-stroke, but did not specify an average duration. Zhao et al. (12) included participants with a mean of 12 years after stroke, while their later work (13) examined individuals with a mean of 9.5 years after stroke (range 1.8–20.8 years). These studies, along with ours, represent a spectrum of time after stroke, highlighting the importance of considering this factor when comparing findings. The observed changes in tendon properties are also likely influenced by levels of spasticity, which can vary significantly across studies. Zhao et al. (12) reported a mean MAS score of 1.6, indicating moderate spasticity, while their later study (13) noted an even higher mean MAS score of 2.5. Dias et al. (16) included participants with an average MAS score of 1.5, similar to Zhao et al. (12). In contrast, our participants exhibited lower plantarflexor muscle tone, with an average MAS score of 0.8. This difference in spasticity levels may contribute to the observed discrepancies in findings. The observations from these studies and our work highlight the complex interplay between factors such as stroke chronicity and muscle hypertonia, with tendon morphology adaptations. Longitudinal analyses are essential to capture tendon morphology changes at multiple sites over time. Understanding these relationships is crucial for developing targeted rehabilitation interventions aimed at optimizing tendon function and improving functional outcomes in individuals post-stroke. Further research utilizing longitudinal designs and controlled spasticity levels is necessary to fully elucidate these dynamics.

While we did not directly assess the mechanical properties of the Achilles tendon in our study, the increased tendon compliance frequently reported in individuals post-stroke (12, 13, 16) may be linked to the tendon morphological changes observed in our findings. In addition, we observed reduced PSFR in the paretic Achilles tendon of individuals post-stroke. Previous research has shown that lower PSFR is associated with decreased tendon stiffness and Young's modulus in degenerated Achilles tendons (25). This adaption at the Achilles tendon insertion in the paretic limb may stem from collagen fiber disorganization and reduced collagen content (16), which result from a complex, abnormal loading environment characterized by factors such as increased muscle tone (12, 14), muscle co-activation (15), and muscle atrophy and disuse (13). It is possible that the morphological changes in the Achilles tendon among individuals with post-stroke hemiparesis may impair the propulsive force generated by the muscle-tendon unit of the paretic ankle plantarflexors. This deficiency, attributed to excessive tendon compliance, likely contributes to reduced walking speed in this cohort. Additionally, it is important to note that decreased plantarflexor strength and abnormal gait pattern with diminished forward propulsion during gait are recognized risk factors for Achilles tendinopathy (26). This insufficient propulsion may further compromise tendon morphology and health in individuals post-stroke.

Furthermore, the alterations observed at the tendon insertion in individuals post-stroke may be associated with metabolic syndrome. Metabolic syndrome is a cluster of conditions that includes central obesity, hypertension, hyperglycemia and dyslipidemia, contributes to systemic inflammation and altered tissue metabolism (30, 36). This multifaceted condition has been identified as a major contributor to cardiovascular diseases, including stroke (30, 37). Metabolic syndrome affects 62% of ischemic stroke patients (38). While each component of metabolic syndrome is associated with an increased risk of stroke, research suggests that truncal obesity and hypertension are independent risk factors (37, 38). Metabolic syndrome also has significant implications for tendon health. The chronic inflammation associated with obesity can impair tendon healing and increase the risk of tendinopathy. Specifically, obesity elevates the risk of Achilles tendinopathy, likely due to the influence of adipokines and the systematic inflammatory response triggered by adipose tissue (30). The Achilles tendon may be particularly susceptible to the effects of metabolic syndrome due to its role in weight-bearing activities and the high mechanical demands placed upon it during locomotor activities, such as walking and running (39–42). Anatomically, the Achilles tendon has a relatively higher blood supply at its insertion point (43, 44), making this area more reliant on adequate circulation. In conditions associated with metabolic syndrome, reduced blood circulation to the Achilles tendon, can compromise the tendon's ability to recover from mechanical fatigue and increase susceptibility to injury (30). Consequently, the Achilles tendon, particularly at its insertion site, is likely at a heightened risk for ischemia and impaired healing in the presence of metabolic dysfunction. In line with existing literature, the recorded PSFR values for both the paretic and non-paretic limbs within the stroke group, measuring 1.70 mm^{-1} and 1.89 mm^{-1} (Table 2), were notably lower compared to the healthy Achilles tendon (2.07 mm^{-1}) documented in previous studies (25). These findings suggest that, beyond adapting to abnormal Achilles tendon loading, individuals post-stroke may have an increased susceptibility to Achilles tendon pathology, potentially due to the influence of metabolic syndrome. The presence of tendinopathy in our participants with post-stroke hemiparesis remained uncertain. Although none of our participants reported pain in either leg, attributing any potential pain specifically to tendinopathy would be challenging in persons post-stroke. This ambiguity stems from the likelihood of potential impaired pain perception related to stroke lesion (45, 46), which could confound the identification of tendinopathy as the underlying cause of pain.

Our research has potential to impact rehabilitation strategies for individuals with post-stroke hemiparesis. By identifying tendon morphological changes as potential contributors to altered walking mechanics, our findings advocate for a more comprehensive treatment approach that includes both muscle and tendon function, especially in addressing insufficient paretic propulsion during gait. In addition, given our focus on higher-functioning stroke participants—demonstrated by the ability to complete a 10-Meter Walk Test, FMA-LE scores (22.6 out of 34), lower MAS (0.8 out of 4), and community ambulation speed (47)—we recognize that individuals with lower functional levels may exhibit more pronounced or different Achilles tendon changes. In reviewing the limited literature, previous studies examining Achilles tendon morphology and mechanical properties did not offer specific details about the functional levels of their post-stroke group beyond their ability to walk independently or with a cane (12, 13, 16). Other functional measures, such as FMA, overground walking speed (as used in our study), balance, or the ability to perform activities of daily living, were not reported. It is important to note that the functional levels of stroke survivors can vary widely depending on the severity and location of the stroke, as well as individual factors such as age, pre-stroke health status, and access to rehabilitation services. The lack of detailed information about the functional levels of the participants in these studies limits the ability to make comparisons to other stroke populations or to draw conclusions about the relationship between functional level and Achilles tendon morphology or mechanical properties. Expanding research to include participants across a broader range of functional abilities and assessing both muscle co-activation patterns during gait and tendon integrity (e.g., mechanical properties and morphology) could help establish clinically meaningful thresholds or minimal important differences in muscle-tendon function post-stroke. Tracking these adaptations over time, particularly in the acute or subacute phases of stroke recovery, would enhance understanding of rehabilitation needs at various recovery stages.

A limitation of our study is the higher body weight observed in the stroke group, although the difference between groups was not statistically significant. Higher body weight is a common outcome post-stroke due to pre-existing obesity and/or reduced physical activity. Increased body weight may elevate Achilles tendon loading, while decreased activity levels could lower it. Additionally, as our previous work showed pronounced tendon thickening at the insertion site (17), this study focused on the distal portion of the Achilles tendon; therefore, examining the entire tendon, including the proximal region, is warranted. Moreover, while stroke participants exhibited lower PSFR and increased tendon thickness in the paretic limb, the specific thresholds for defining abnormal morphology remain unclear. Although tendon mechanical properties have been previously studied in individuals post-stroke, this study did not evaluate them in conjunction with Achilles tendon morphology. Key biomechanical factors during walking—such as propulsive forces, joint kinematics and kinetics, and muscle co-activation—were also not assessed. To address these gaps, future research should prioritize longitudinal analyses to monitor tendon morphology changes across multiple sites, establish diagnostic thresholds for PSFR and tendon thickness, and consider the impact of factors such as metabolic syndrome in individuals post-stroke. Controlling for body weight and activity levels between stroke and non-stroke participants will also be critical. Building on our findings, it is essential to adopt a more comprehensive research design that integrates tendon morphology, tendon mechanical properties, and biomechanical factors during walking—including kinetics, kinematics, and muscle activation patterns. This holistic approach will provide deeper insights into the mechanisms underlying tendon health and gait performance in individuals post-stroke.

5 Conclusions

Individuals with post-stroke hemiparesis exhibited increased thickness both 2 cm above the insertion and at the insertion site of the Achilles tendon within the paretic limb, along with marked disorganization of collagen fiber bundles at the insertion point. These morphological changes are likely the result of a complex, abnormal loading environment, influenced by factors such as increased plantarflexor muscle tone, muscle co-activation, and muscle atrophy and disuse in the paretic limb. Such changes in tendon structure may lead to increased compliance of the Achilles tendon during gait, impairing force transmission and propulsion in the paretic limb, and ultimately contributing to slower walking speed post-stroke. These findings highlight the importance of a comprehensive treatment approach focused on improving neuromuscular control in this population. To enhance walking speed in individuals post-stroke, it is important to assess and address Achilles tendon integrity, particularly in those with increased plantarflexor muscle tone, inappropriate muscle co-activation patterns, insufficient paretic propulsive forces, and/or metabolic syndrome. Targeting these interconnected impairments may facilitate better recovery of functional gait and overall mobility in this population. The clinical implications of our research findings warrant further investigation to optimize rehabilitation strategies.

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 humans were approved by University of Nevada, Las Vegas Institutional Review Board. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

JL: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Project administration, Resources, Supervision, Validation, Writing – original draft, Writing – review & editing. GB: Methodology, Writing – original draft, Writing – review & editing, Software. KK: Methodology, Writing – original draft, Writing – review & editing. K-YH: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Project administration, Resources, Supervision, Validation, Visualization, Writing – original draft, Writing – review & editing.

Funding

The author(s) declare financial support was received for the research, authorship, and/or publication of this article. The study was supported by the Faculty Opportunity Award, University of Nevada, Las Vegas (JL and K-YH). The publication fees for this article was supported by the UNLV University Libraries Open Article Fund.

Acknowledgments

The authors would like to thank Allina Cummins, Amanda Ferraro, Daniel Krist, Mira Seekins, and Trever Diez for their assistance in data collection and analyses. The authors utilized OpenAI's ChatGPT (version GPT-4) to correct grammatical errors.

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.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

References

1. Eng JJ, Tang P-F. Gait training strategies to optimize walking ability in people with stroke: a synthesis of the evidence. *Expert Rev Neurother*. (2007) 7:1417–36. doi: 10.1586/14737175.7.10.1417

2. Awad LN, Binder-Macleod SA, Pohlig RT, Reisman DS. Paretic propulsion and trailing limb angle are key determinants of long-distance walking function after stroke. *Neurorehabil Neural Repair*. (2015) 29:499–508. doi: 10.1177/1545968314554625

3. Awad LN, Reisman DS, Pohlig RT, Binder-Macleod SA. Reducing the cost of transport and increasing walking distance after stroke: a randomized controlled trial on fast locomotor training combined with functional electrical stimulation. *Neurorehabil Neural Repair*. (2016) 30:661–70. doi: 10.1177/1545968315619696

4. Olney SJ, Richards C. Hemiparetic gait following stroke. Part I: characteristics. *Gait Posture*. (1996) 4:136–48. doi: 10.1016/0966-6362(96)01063-6

5. Bowden MG, Behrman AL, Neptune RR, Gregory CM, Kautz SA. Locomotor rehabilitation of individuals with chronic stroke: difference between responders and nonresponders. *Arch Phys Med Rehabil*. (2013) 94:856–62. doi: 10.1016/j.apmr.2012.11.032

6. Forster A, Young J. Incidence and consequences of falls due to stroke - a systematic inquiry. *Br Med J*. (1995) 311:83–6. doi: 10.1136/bmj.311.6997.83

7. Harris JE, Eng JJ. Goal priorities identified through client-centred measurement in individuals with chronic stroke. *Physiother Can*. (2004) 56:171–6. doi: 10.2310/6640.2004.00017

8. Bowden MG, Balasubramanian CK, Neptune RR, Kautz SA. Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. *Stroke*. (2006) 37:872–6. doi: 10.1161/01.STR.0000204063.75779.8d

9. Awad LN, Reisman DS, Kesar TM, Binder-Macleod SA. Targeting paretic propulsion to improve poststroke walking function: a preliminary study. *Arch Phys Med Rehabil*. (2014) 95:840–8. doi: 10.1016/j.apmr.2013.12.012

10. Parvataneni K, Olney SJ, Brouwer B. Changes in muscle group work associated with changes in gait speed of persons with stroke. *Clin Biomech*. (2007) 22:813–20. doi: 10.1016/j.clinbiomech.2007.03.006

11. Awad LN, Hsiao H, Binder-Macleod SA. Central drive to the paretic ankle plantarflexors affects the relationship between propulsion and walking speed after stroke. *J Neurol Phys Ther*. (2020) 44:42–8. doi: 10.1097/NPT.0000000000000299

12. Zhao H, Ren Y, Wu YN, Liu SQ, Zhang LQ. Ultrasonic evaluations of achilles tendon mechanical properties poststroke. *J Appl Physiol (1985)*. (2009) 106:843–9. doi: 10.1152/japplphysiol.91212.2008

13. Zhao H, Ren Y, Roth EJ, Harvey RL, Zhang LQ. Concurrent deficits of soleus and gastrocnemius muscle fascicles and achilles tendon post stroke. *J Appl Physiol (1985)*. (2015) 118:863–71. doi: 10.1152/japplphysiol.00226.2014

14. Freire B, Dias CP, Goulart NB, De Castro CD, Becker J, Gomes I, et al. Achilles tendon morphology, plantar flexors torque and passive ankle stiffness in spastic hemiparetic stroke survivors. *Clin Biomech (Bristol, Avon)*. (2017) 41:72–6. doi: 10.1016/j.clinbiomech.2016.12.004

15. Dyer JO, Maupas E, Melo Sde A, Bourbonnais D, Forget R. Abnormal coactivation of knee and ankle extensors is related to changes in heteronymous spinal pathways after stroke. *J Neuroeng Rehabil*. (2011) 8:41. doi: 10.1186/1743-0003-8-41

16. Dias CP, Freire B, Goulart NBA, Dias De Castro C, Lemos FA, Becker J, et al. Impaired mechanical properties of achilles tendon in spastic stroke survivors: an observational study. *Top Stroke Rehabil*. (2019) 26:261–6. doi: 10.1080/10749357.2019.1591688

17. Liang JN, Ho KY. Altered achilles tendon morphology in individuals with chronic post-stroke hemiparesis: a case report. *BMC Med Imaging*. (2020) 20:34. doi: 10.1186/s12880-020-00431-0

18. Ozcakar L, Komurcu E, Safaz I, Goktepe AS, Yazicioglu K. Evaluation of the patellar tendon in transtibial amputees: a preliminary sonographic study. *Prosthet Orthot Int*. (2009) 33:324–8. doi: 10.3109/03093640903171010

19. Arya S, Kulig K. Tendinopathy alters mechanical and material properties of the achilles tendon. *J Appl Physiol (1985)*. (2010) 108:670–5. doi: 10.1152/japplphysiol.00259.2009

20. Kulig K, Landel R, Chang YJ, Hannanvash N, Reischl SF, Song P, et al. Patellar tendon morphology in volleyball athletes with and without patellar tendinopathy. *Scand J Med Sci Sports*. (2013) 23:e81–88. doi: 10.1111/sms.12021

21. Ho KY, Harty M, Kellogg J, Teter K, Lee SP, Chang YJ, et al. Patellar tendon morphology in trans-tibial amputees utilizing a prosthesis with a patellar-tendon-bearing feature. *Sci Rep*. (2019) 9:16392. doi: 10.1038/s41598-019-52747-9

22. Helland C, Bojsen-Moller J, Raastad T, Seynnes OR, Moltubakk MM, Jakobsen V, et al. Mechanical properties of the patellar tendon in elite volleyball players with and without patellar tendinopathy. *Br J Sports Med*. (2013) 47:862–8. doi: 10.1136/bjsports-2013-092275

23. Ho KY, Baquet A, Chang YJ, Chien LC, Harty M, Bashford G, et al. Factors related to intra-tendinous morphology of achilles tendon in runners. *PLoS One*. (2019) 14:e0221183. doi: 10.1371/journal.pone.0221183

24. Bashford GR, Tomsen N, Arya S, Burnfield JM, Kulig K. Tendinopathy discrimination by use of spatial frequency parameters in ultrasound B-mode images. *IEEE Trans Med Imaging*. (2008) 27:608–15. doi: 10.1109/TMI.2007.912389

25. Kulig K, Chang YJ, Winiarski S, Bashford GR. Ultrasound-based tendon micromorphology predicts mechanical characteristics of degenerated tendons. *Ultrasound Med Biol*. (2016) 42:664–73. doi: 10.1016/j.ultrasmedbio.2015.11.013

26. Van Der Vlist AC, Breda SJ, Oei EHG, Verhaar JaN, De Vos R-J. Clinical risk factors for achilles tendinopathy: a systematic review. *Br J Sports Med*. (2019) 53:1352–61. doi: 10.1136/bjsports-2018-099991

27. Giai Via A, Oliva F, Padulo J, Oliva G, Maffulli N. Insertional calcific tendinopathy of the achilles tendon and dysmetabolic diseases: an epidemiological survey. *Clin J Sport Med*. (2022) 32:e68–73. doi: 10.1097/JSM.0000000000000881

28. Gaida JE, Cook JL, Bass SL. Adiposity and tendinopathy. *Disabil Rehabil*. (2008) 30:1555–62. doi: 10.1080/09638280701786864

29. Gaida JE, Alfredson L, Kiss ZS, Wilson AM, Alfredson H, Cook JL. Dyslipidemia in achilles tendinopathy is characteristic of insulin resistance. *Med Sci Sports Exerc*. (2009) 41:1194–7. doi: 10.1249/MSS.0b013e31819794c3

30. Lai C, Li R, Tang W, Liu J, Duan X, Bao D, et al. Metabolic syndrome and tendon disease: a comprehensive review. *Diabetes Metab Syndr Obes*. (2024) 17:1597–609. doi: 10.2147/DMSO.S459060

31. Thilmann AF, Fellows SJ, Ross HF. Biomechanical changes at the ankle joint after stroke. *J Neurol Neurosurg Psychiatry*. (1991) 54:134–9. doi: 10.1136/jnnp.54.2.134

32. Cheng DK, Nelson M, Brooks D, Salbach NM. Validation of stroke-specific protocols for the 10-meter walk test and 6-minute walk test conducted using 15-meter and 30-meter walkways. *Top Stroke Rehabil*. (2020) 27:251–61. doi: 10.1080/10749357.2019.1691815

33. Fugl-Meyer AR, Jaasko L, Leyman I, Olsson S, Steglind S. The post-stroke hemiplegic patient. 1. a method for evaluation of physical performance. *Scand J Rehabil Med*. (2011) 7:13–31. doi: 10.2340/1650197771331

34. Harb A, Kishner S. *Modified Ashworth Scale*. Treasure Island, FL: StatPearls Publishing (2023).

35. Ho KY, Deaver BB, Nelson T, Turner C. Using a mobile application to assess knee Valgus in healthy and post-anterior cruciate ligament reconstruction participants. *J Sport Rehabil*. (2019) 28:532–5. doi: 10.1123/jsr.2018-0278

36. Park YH, Kim W, Kim JY, Choi GW, Kim HJ. Clinical impact of metabolic syndrome on eccentric exercises for chronic insertional achilles tendinopathy. *J Foot Ankle Surg*. (2022) 61:726–9. doi: 10.1053/j.jfas.2021.03.020

37. Amir A, Hassan M, Alvi S, Mueed A, Idrees S, Ashraf J, et al. Frequency and characteristics of metabolic syndrome in patients with ischemic stroke admitted to a tertiary care hospital in Karachi. *Cureus*. (2020) 12:e9004. doi: 10.7759/cureus.9004

38. Ashtari F, Salari M, Aminoroaya A, Deljoo BK, Moeini M. Metabolic syndrome in ischemic stroke: a case control study. *J Res Med Sci*. (2012) 17:167–70.

39. Kjaer M. Role of extracellular matrix in adaptation of tendon and skeletal muscle to mechanical loading. *Physiol Rev*. (2004) 84:649–98. doi: 10.1152/physrev.00031.2003

40. Kjaer M, Langberg H, Miller BF, Boushel R, Crameri R, Koskinen S, et al. Metabolic activity and collagen turnover in human tendon in response to physical activity. *J Musculoskelet Neuronal Interact*. (2005) 5:41–52.

41. Bojsen-Møller J, Kalliokoski KK, Seppänen M, Kjaer M, Magnusson SP. Low-intensity tensile loading increases intratendinous glucose uptake in the achilles tendon. *J Appl Physiol (1985)*. (2006) 101:196–201. doi: 10.1152/japplphysiol.00004.2006

42. Christensen B, Dyrberg E, Aagaard P, Enehjelm S, Krogsgaard M, Kjaer M, et al. Effects of long-term immobilization and recovery on human triceps surae and collagen turnover in the achilles tendon in patients with healing ankle fracture. *J Appl Physiol (1985)*. (2008) 105:420–6. doi: 10.1152/japplphysiol.00201.2008

43. Ahmed IM, Lagopoulos M, Mcconnell P, Soames RW, Sefton GK. Blood supply of the achilles tendon. *J Orthop Res*. (1998) 16:591–6. doi: 10.1002/jor.1100160511

44. Stein V, Laprell H, Tinnemeyer S, Petersen W. Quantitative assessment of intravascular volume of the human achilles tendon. *Acta Orthop Scand*. (2000) 71:60–3. doi: 10.1080/00016470052943919

45. Sullivan JE, Hedman LD. Sensory dysfunction following stroke: incidence, significance, examination, and intervention. *Top Stroke Rehabil*. (2008) 15:200–17. doi: 10.1310/tsr1503-200

46. Kessner SS, Schlemm E, Gerloff C, Thomalla G, Cheng B. Grey and white matter network disruption is associated with sensory deficits after stroke. *NeuroImage: Clinical*. (2021) 31:102698. doi: 10.1016/j.nicl.2021.102698

47. Bowden MG, Balasubramanian CK, Behrman AL, Kautz SA. Validation of a speed-based classification system using quantitative measures of walking performance poststroke. *Neurorehabil Neural Repair*. (2008) 22:672–5. doi: 10.1177/1545968308318837

TYPE Original Research
PUBLISHED 12 March 2025
DOI 10.3389/fspor.2025.1545813

Check for updates

OPEN ACCESS

EDITED BY
Navrag B. Singh,
ETH Zürich, Switzerland

REVIEWED BY
Ryoichi Ema,
Shizuoka Sangyo University, Japan
Datao Xu,
Ningbo University, China

*CORRESPONDENCE
Lida Mademli
Lmademli@phed-sr.auth.gr
Adamantios Arampatzis
a.arampatzis@hu-berlin.de

RECEIVED 15 December 2024
ACCEPTED 26 February 2025
PUBLISHED 12 March 2025

CITATION
Mademli L, Nikolaidou M-E, Bohm S and Arampatzis A (2025) Postural control of sway dynamics on an unstable surface reduces similarity in activation patterns of synergistic lower leg muscles.
Front. Sports Act. Living 7:1545813.
doi: 10.3389/fspor.2025.1545813

Postural control of sway dynamics on an unstable surface reduces similarity in activation patterns of synergistic lower leg muscles

Lida Mademli[1]*, Maria-Elissavet Nikolaidou[2], Sebastian Bohm[3,4] and Adamantios Arampatzis[3,4]*

[1]Laboratory of Adapted Physical Education, School of Physical Education and Sport Science (Serres), Faculty of Physical Education and Sport Sciences, Aristotle University of Thessaloniki, Thessaloniki, Greece, [2]Sports Biomechanics Laboratory, School of Physical Education and Sport Science, Faculty of Physical Education and Sport Science, National and Kapodistrian University of Athens, Athens, Greece, [3]Department of Training and Movement Sciences, Humboldt-Universität zu Berlin, Berlin, Germany, [4]Berlin School of Movement Science, Berlin, Germany

Introduction: Diversity of activation patterns within synergistic muscles can be important for stability control in challenging conditions. This study investigates the similarity of activation patterns within the triceps surae and quadriceps femoris muscles and the effects of unstable surface during a visually guided postural task.
Methods: Eighteen healthy adults performed a visually guided anteroposterior tracking task on both stable and unstable surfaces. Electromyographic activity of triceps surae (gastrocnemius medialis, gastrocnemius lateralis, soleus) and quadriceps femoris (vastus medialis, vastus lateralis, rectus femoris) was recorded at 1,000 Hz. Cosine similarity (CS) between muscle pairs within each muscle group was calculated to assess the similarity of activation patterns of synergistic muscles for stable and unstable conditions. To compare the CS of the muscle pairs, a linear mixed model was used. For all tests the level of significance was set to $\alpha = 0.05$.
Results: Across all surface conditions, CS values within the triceps surae muscles were lower than those of the quadriceps ($p < 0.001$), indicating a greater diversity in activation patterns of the distal muscles. The unstable surface reduced CS values for both muscle groups ($p = 0.021$). No significant interaction was observed between muscle pair and surface condition ($p = 0.833$).
Discussion: The reduced similarity of activation patterns within the synergistic triceps surae and quadriceps femoris muscles on the soft surface indicates an increased flexibility of neuromotor control for the unstable condition. The lower similarity within the synergistic triceps surae muscles suggests a higher diversity of activation patterns compared to the quadriceps femoris muscles, which may increase the flexibility of neuromotor control to meet specific joint stabilization challenges during the studied tracking task.

KEYWORDS

triceps surae muscle, quadriceps femoris muscle, balance control, diversity of activation pattens, flexibility of neuromotor control

1 Introduction

Successful movement requires efficient control of the human musculoskeletal system by the central nervous system. Given the large number of degrees of freedom within the musculoskeletal system, this control process is highly complex and not yet fully understood. To address this complexity, it has been proposed that muscles are controlled through muscle modules, whereby a single neural command activates a group of muscles, thereby reducing the dimensionality of movement control (1, 2). The modular organization of lower leg muscle activation affects the efficiency, robustness and mechanical loading of the joints (3–5) and is therefore important in everyday tasks such as standing balance (6, 7), walking and running (8–11), landing (12, 13) and perturbed locomotion (14–16). During challenging conditions on unstable surfaces, such as locomotion, transitioning from double to single-leg stance, landing or lunging on unstable surfaces, the human neuromotor system compensates and enhances its robustness by modulating muscle synergies, i.e., by widening the time dependent components of muscle synergies (7, 12, 17, 18). Furthermore, muscle coactivation around knee and ankle joint increases, when performing balance or postural tasks on unstable conditions (12, 18, 19) with the increase being more pronounced around the ankle joint (12, 18). In the above mentioned studies, the triceps surae muscles were reported to be involved in the same module as synergistic muscles. The same is also true for the synergistic quadriceps muscles (3, 13, 17). However, this synergistic behavior seems to be more complicated, since it has been suggested that not all anatomically defined synergist muscles share a strong common neural drive (20–22). It has been suggested that an independent drive to certain muscles within the same muscle group enables more flexible control (21), allowing for a redistribution of neural drive across muscles to cope with challenging neuromotor task conditions (22).

For instance, while a strong common drive between muscles such as vastus lateralis and vastus medialis might be necessary to protect the knee joint from excessive internal contact stresses (23, 24), other synergistic muscles, such as the two gastrocnemii, might require a more flexible control strategy to comply with secondary goals, e.g., ankle stabilization (20, 21). These differences between synergistic muscles suggest possible diversity in the activation patterns, which may affect the force-sharing among the synergistic muscles. Lai et al. (25) and Hamard et al. (26) reported a diversity in electromyographic (EMG) activity among the synergistic triceps surae muscles during walking and running. In situations where the application of forces is required in multiple directions, such as around joints responsible for postural control like the ankle, greater flexibility in muscle control may be required to manage varying external demands. A diversity of activation patterns among synergistic muscles may improve task performance. Under conditions of increased postural demands, the need for more flexible motor control becomes greater, and a wide range of possible activation strategies among synergistic muscles may increase the versatility for effective control and regulation of body stability. However, to the best of our knowledge, no study has yet examined the similarity of activation patterns within different muscles groups in both stable and unstable balance conditions.

The primary aim of this study was to determine (1) whether the similarity of activation patterns within the distal triceps surae muscles differs from that within the more proximal quadriceps femoris muscles during a visually guided postural task, and (2) whether the similarity of activation patterns within synergistic muscles is altered under unstable conditions. To create an unstable and thus a more challenging condition for the neuromotor system, we introduced surface-related perturbations by having participants standing on a foam surface while performing a visually guided postural task. In previous research, we found that these surface-related perturbations had a significant impact on the local dynamic stability, with greater short-term maximum Lyapunov exponents (sMLE) values for the whole-body kinematics (from head to foot), implying that tracking a visual moving target on an unstable surface is a more demanding and more challenging task than doing so on a stable surface (19).

We studied the activation patterns of synergistic muscles from two different muscle groups—the ticeps surae and quadriceps femoris—as previous literature indicates that they differ in the level of common neural input shared between their muscle heads. To quantify the similarity of the activation patterns between the different muscles within the same muscle groups, we used the cosine similarity (27, 28). First, based on previous findings of increased shared neural input between the vastus lateralis (VL) and vastus medialis (VM) muscles compared to the gastrocnemius lateralis (GL) and gastrocnemius medialis (GM) muscles (22), we hypothesized that during a visually guided postural task, the similarity of the activation pattern within the quadriceps femoris muscles would be greater than that within the triceps surae muscles. Second, based on reports that unstable surfaces increase body instability during visually guided postural tasks (19, 29), we hypothesized a decrease in the similarity of activation patterns within the synergistic triceps surae and quadriceps femoris muscles in the unstable condition.

2 Methods

2.1 Participants

Twenty healthy young adults (age 32 ± 5 years, mean ± standard deviation) with no history of neuromuscular or musculoskeletal disorders participated in the study (12 male and 8 female, body height 175 ± 7 cm, body mass 69 ± 12 kg). From two participants not all data (EMG activity, ground reaction forces) were accurately collected for all examined trials, thus the data from 18 adults were used in the analysis (11 males and 7 females, body height 175 ± 8 cm, body mass 72 ± 11 kg, age 32 ± 6 years). All participants were informed about the experimental protocol and gave their informed consent before their inclusion to the study. The experiment was performed with the approval of the institution's ethics committee (HU-KSBF-EK_2018_0013, Humboldt-Universität zu Berlin) and in accordance with the Declaration of Helsinki.

2.2 Task and apparatus

Participants had to perform a postural tracking task by shifting their weight anteroposteriorly during standing on a force plate (40 × 60 cm, AMTI BP400600-2000, Advanced Mechanical Technology, Inc., Watertown, MA, USA) under two different conditions: (a) on a stable surface (floor) and (b) on an unstable surface (foam). The foam consisted of two balance beams (Sport-Thieme Balance beam EVA foam, with the dimensions 38 × 16.5 × 5.8 cm, Sport-Thieme Germany), which were oriented in the foot's anterior to posterior plane. The tracking task involved following a visual target, which was displayed on a TV screen (47″, HD LG) as a red dot and moved vertically in the center of the screen. In addition to the target, the TV screen displayed a yellow dot representing the participant's anteroposterior center of pressure (CoP) component, providing real-time feedback of the participant's position on the force plate. Both the target and CoP position feedback were synchronously shown on the TV monitor by means of a custom-built software, developed in MATLAB R2014 (Math Works Inc., USA), with 50 Hz refresh rate. Participants were instructed to follow the red dot's movement by adjusting the yellow dot through forward and backward weight shifts, while keeping their knees and hips straight. Specifically, when the red dot moved upward, participants had to shift their weight anteriorly, and when it moved downward, they had to shift their weight posteriorly. Each participant performed one trial per condition in a randomized order to prevent sequence effects, with each trial lasting 120 s. Prior to each trial, there was a familiarization period lasting maximum 30 s. During the trial, participants stood barefoot on the force plate with their feet at the center of the platform, maintaining an inter-malleolar distance equal to 10% of their body height. They held their arms akimbo to ensure stability and eliminate any influence of arm movements on balance and task performance. The experimental set up is depicted in Figure 1.

The target's motion was periodic sinusoidal with a single frequency (f) set at 0.25 Hz, which was generated in MATLAB R2014b using the sine function [$\text{sine}(t) = \sin(2\pi i \times \text{f} \times t)$]. This frequency was selected because it is the dominant frequency of natural, self-paced voluntary sway (30, 31). The maximum (peak to peak) amplitude of target motion was adjusted for each participant to 60% of his/her foot length, which corresponded to an average amplitude of 15.08 ± 1.09 cm. This amplitude was chosen based on pilot tests showing that 60% of foot length is suitable for swaying on both stable and unstable surfaces, as it effectively increases instability on the unstable surface compared to the rigid one, consistent with findings from earlier studies (7, 17).

The participants' postural behaviour was recorded using the force platform at 1,000 Hz. The EMG activity of soleus (SOL),

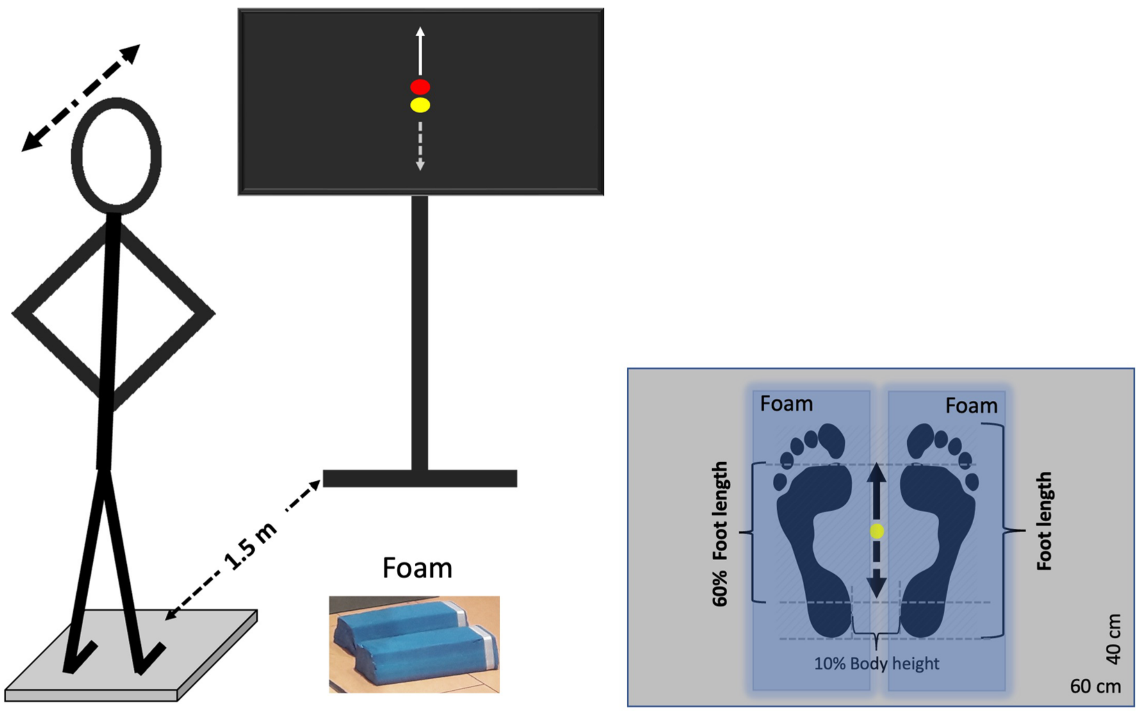

FIGURE 1
Experimental set up. Participants tracked a visual target (red dot) moving vertically using a yellow dot, which was controlled by their center of pressure (CoP) movement in the anteroposterior direction. They maintained their feet on the ground and leaned their body forward and backward. An anterior CoP shift corresponded to an upward movement of the yellow dot (solid arrow), while a posterior shift corresponded to a downward movement (dashed arrow). The target moved sinusoidally at 0.25 Hz, with an amplitude normalized to 60% of the participant's foot length. The task was performed on both a stable (rigid) surface and an unstable (foam) surface. Blue-shaded areas indicate the position of the foam pads for the unstable trial. Each tracking task lasted 120 s.

GM, GL, rectus femoris (RF), VL and VM was captured at a sampling frequency of 1,000 Hz using a wireless EMG system (myon m320, myon AG, Schwarzenberg, Switzerland). The wireless electrodes were placed at the right lower limb, on the belly of each muscle, with a 2 cm intra-electrode distance according to SENIAM recommendations. All data (force and EMG signals) were simultaneously recorded by a motion capture system (Vicon, Oxford, U.K.). An interface was created in MATLAB R2014b for triggering and synchronizing all devices with a single pulse.

2.3 EMG processing

EMG activity was processed and analysed in MATLAB R2023b. First, for each trial and participant, the data of the first cycle of the target motion were excluded for any further analysis, since most of the participants were not able to follow the target accurately during this first cycle. The raw EMG signals were high-pass filtered at 20 Hz using a 4th-order Butterworth zero-lag filter to remove movement artifacts and low-frequency noise. The signals were then full-wave rectified and smoothed with a low-pass filter at 5 Hz using a 4th order Butterworth zero-lag filter to extract the envelope of the EMG activity. The filtering was applied sequentially, with high-pass filtering preceding rectification and low-pass smoothing.

To normalise the EMG signals, a single average reference value was calculated for each participant and each muscle from the filtered and rectified EMG signal from both the stable and unstable conditions. Specifically, for each muscle the mean of the EMG signal was calculated separately for the stable and unstable condition over the entire duration (excluding the first cycle). The average of these two mean values was then used as a constant reference to normalize the EMG signals for each corresponding muscle and participant for both conditions. To account for baseline muscle activation, the minimum value of the normalized EMG signal of the two conditions (stable and unstable) was identified for each muscle. The lower of these two values was then subtracted from the normalized EMG to remove the baseline activation for each muscle and each participant in both conditions. Henceforth, any reference to the EMG signal denotes the EMG signal that has been filtered, normalized, and adjusted for baseline activation.

The cosine similarity (CS) between the EMG signals of different synergistic muscle pairs was calculated in MATLAB R2023b, to assess the similarity between the EMG signals of the examined muscle groups (27, 28) under both stable and unstable conditions. The CS between each pair of EMG signals was calculated using the dot product of the two signals divided by the product of their norms. This was done separately for the stable and unstable conditions using the following equation:

$$CS_{EMGi} = \frac{\text{dot}\,(EMG_x,\ EMG_y)}{\|\ EMG_x\ \| \quad \times \quad \|\ EMG_y\ \|}$$

where CS_{EMGi} is the cosine similarity value for a given muscle pair and condition, EMG_x and EMG_y represent the EMG signals of the two different muscles. Note that cosine similarity values can range from 1, indicating a perfect match in pattern, to −1, indicating a perfect inverse pattern.

The CS was calculated at the stable and unstable conditions for the EMG signals of the following synergistic muscle pairs: GM-GL, GM-SOL and GL-SOL for the triceps surae muscle, and RF-VL, RF-VM and VM-VL for the quadriceps femoris. Furthermore, the CS between the target signal and CoP was also calculated to evaluate the participants' accuracy in tracking of the moving visual target under both conditions. This was accordingly done by computing the dot product of the anteroposterior coordinates of the CoP during each tracking task with the sinusoidal signal representing the target's motion and then dividing the outcome by the product of their norms.

2.4 Statistical analysis

All statistical analyses were performed using IBM SPSS statistics (version 29, Armonk, NY:IBM Corp). The normality of the CS data was confirmed with the Shapiro–Wilk test. The values of CS between the CoP and Target, representing the accuracy of CoP-Target motion coupling, were compared between the stable and unstable conditions using a dependent samples *t*-test. A linear mixed model was used to compare the CS of the EMG signals between the examined muscle pairs, groups, and surface conditions. The model included one random factor (participants) to account for the repeated measures within individuals, and three fixed factors: condition (stable vs. unstable surface), muscle pair (GM-GL, GM-SOL, GL-SOL, VL-VM, VL-RF, VM-RF), and muscle group (triceps surae vs. quadriceps femoris). *post hoc* pairwise comparisons using dependent samples *t*-test with the Sidak correction to adjust for multiple comparisons were conducted to explore significant main effects and interactions between the three fixed factors. The normality of the residuals for the linear mixed model was assessed using the Shapiro–Wilk test and Q-Q plots. While the majority of the data were normally distributed, deviations were observed in the residuals for CS of two muscle pairs (RF-VM in the stable condition and VM-VL in the unstable condition). These deviations were considered acceptable given the robustness of the linear mixed model to minor violations of normality (32). Further, residual diagnostics, including Q-Q plots and scatter plots of residuals vs. predicted values, were examined to ensure that the assumptions of normality, homoscedasticity, and linearity were reasonably met. For all tests the level of significance was set to $\alpha = 0.05$.

3 Results

In Figure 2, the target signal and the group averaged CoP displacement on both stable and unstable surfaces are depicted. The CS values between CoP and Target for all participants

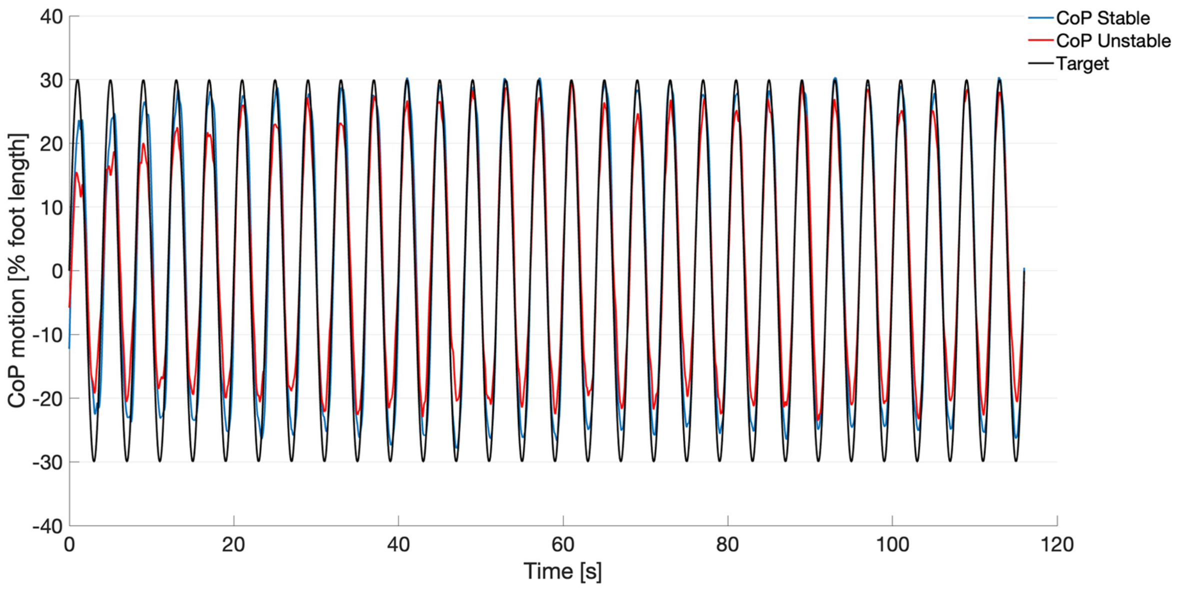

FIGURE 2
Target signal (black) and mean curves of the center of pressure (CoP) motion from all participants (n = 18) during the tracking task on the stable surface (blue) and on the unstable surface (red).

ranged from 0.79–0.95 for the stable condition and from 0.66–0.93 for the unstable condition. The pairwise comparison revealed significantly greater values for the stable condition (0.91 ± 0.04) compared to the unstable condition (0.84 ± 0.07, $t(14) = 3.917$, $p = 0.002$).

The EMG activity of the examined muscles for one participant is indicatively depicted in Figure 3 for both stable and unstable condition. Overall, there was a significant surface effect on the CS values between the EMG signal of all synergistic muscle pairs [$F(1, 187) = 5.425$, $p = 0.021$], with lower values for the unstable surface (Figure 4), indicating less synchronized muscle activation patterns on the unstable surface independent of muscle group.

For all surface conditions the quadriceps femoris muscle group demonstrated greater CS values than the triceps surae muscle group [$F(1, 187) = 56.529$, $p < 0.001$] (Figure 4). When comparing the CS between muscle pairs, significant differences were found, independent of surface condition [$F(4, 187) = 4.582$, $p = 0.001$]. *post hoc* pairwise comparisons revealed significant differences in CS between the examined muscle pairs (Figure 4). Specifically, the CS of muscle pairs GM-GL and GL-SOL was significantly lower than that of all quadriceps femoris muscle pairs (i.e., RF-VL $p < 0.01$, RF-VM $p = 0.01$, VM-VL $p < 0.01$). The CS of muscle group GM-SOL was greater compared to GL-SOL ($p = 0.019$) and lower compared to VM-VL ($p = 0.004$). Within the quadriceps femoris muscle group, no significant differences in CS values were found between muscle pairs (RF-VL vs. RF-VM: $p = 1.000$, RF-VL vs. VM-VL: $p = 0.753$, RF-VM vs. VM-VL: $p = 0.227$). There was no significant interaction between muscle pair and surface condition [$F(4, 187) = 0.366$, $p = 0.833$].

4 Discussion

In the present study, the CSs of the synergistic muscle pairs were higher in the quadriceps femoris compared to the triceps surae muscles, indicating superior similarity in the activation patterns of the more proximal muscles during the tracking task studied, confirming our first hypothesis. Greater synchronization between the quadriceps femoris compared to the triceps surae muscles was also reported by Rossato et al. (22) during isometric contractions. The authors argued (22) that more independent synaptic input within synergistic muscles may provide more flexible neuromotor control to meet specific joint stabilization challenges. The human triceps surae muscles are pennate with a substantial in series elasticity due to the long Achilles tendon. Pennate muscles with substantial in series elasticity are particularly sensitive to external perturbations and capable of flexible force generation for body stability (33, 34). Although the three synergistic triceps surae muscles are pennate with high in series elasticity, they have important differences in terms of function and geometry. The biarticular GM and GL muscles have variable frontal moment arms, resulting in different functional outcomes across the frontal plane range of motion (35). Furthermore, their activation can be altered during balancing tasks due to the different mechanical advantages of the GM and GL muscles (20). The ankle joint plays a key role in human body stability during balance tasks (36, 37). Consequently, during the studied tracking task, a greater diversity of activation patterns within the triceps surae muscles may increase the flexibility of neuromotor control to manage secondary goals, such as producing opposing ankle moments in the frontal plane to maintain mediolateral body stability.

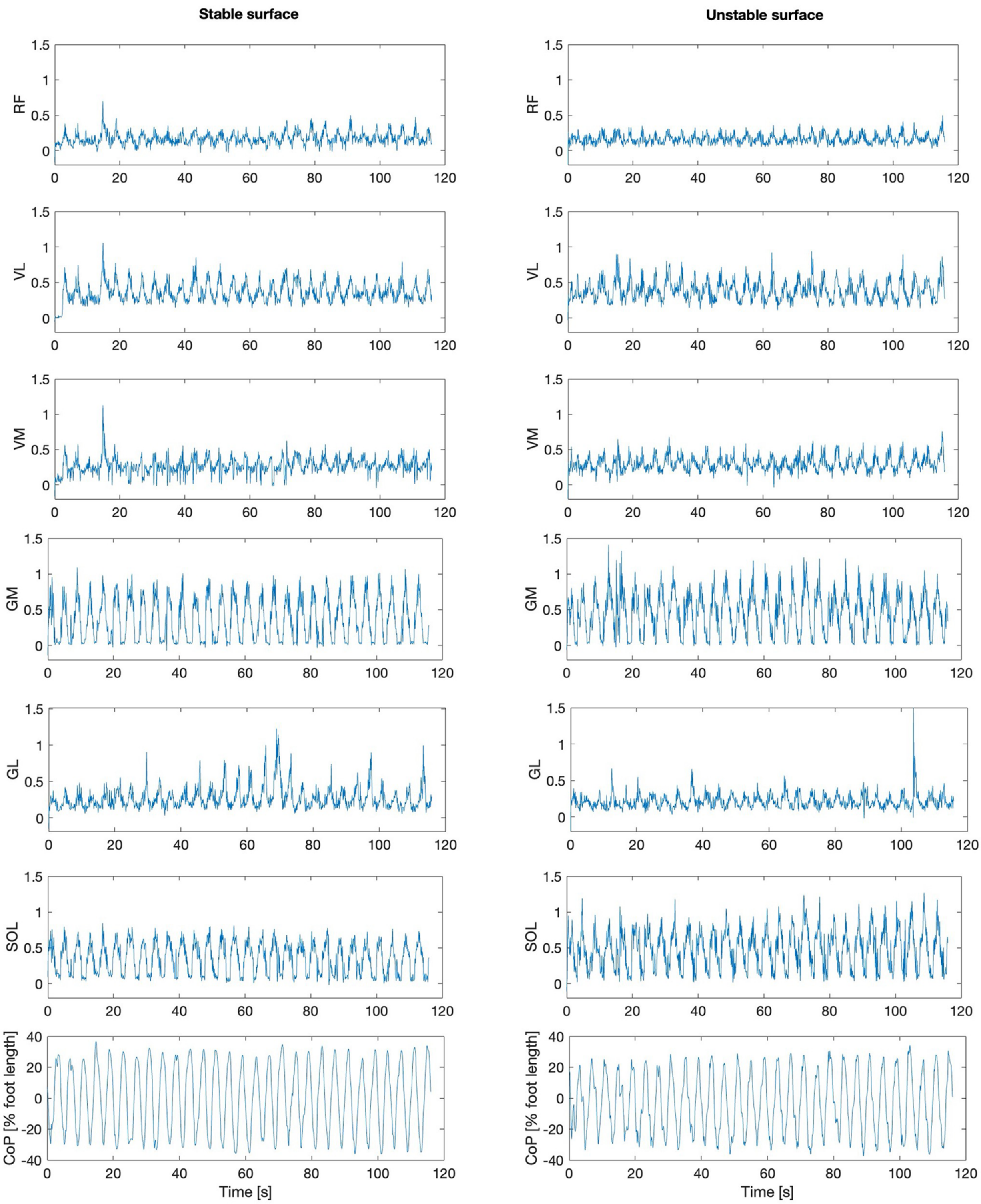

FIGURE 3
The center of pressure motion (CoP) and the processed electromyographic activity [mV] of all examined muscles: rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), gastrocnemius medialis (GM), gastrocnemius lateralis (GL), and soleus (SOL), during the postural tracking task on the stable (left) and on the unstable surface (right) are depicted exemplarily for one participant.

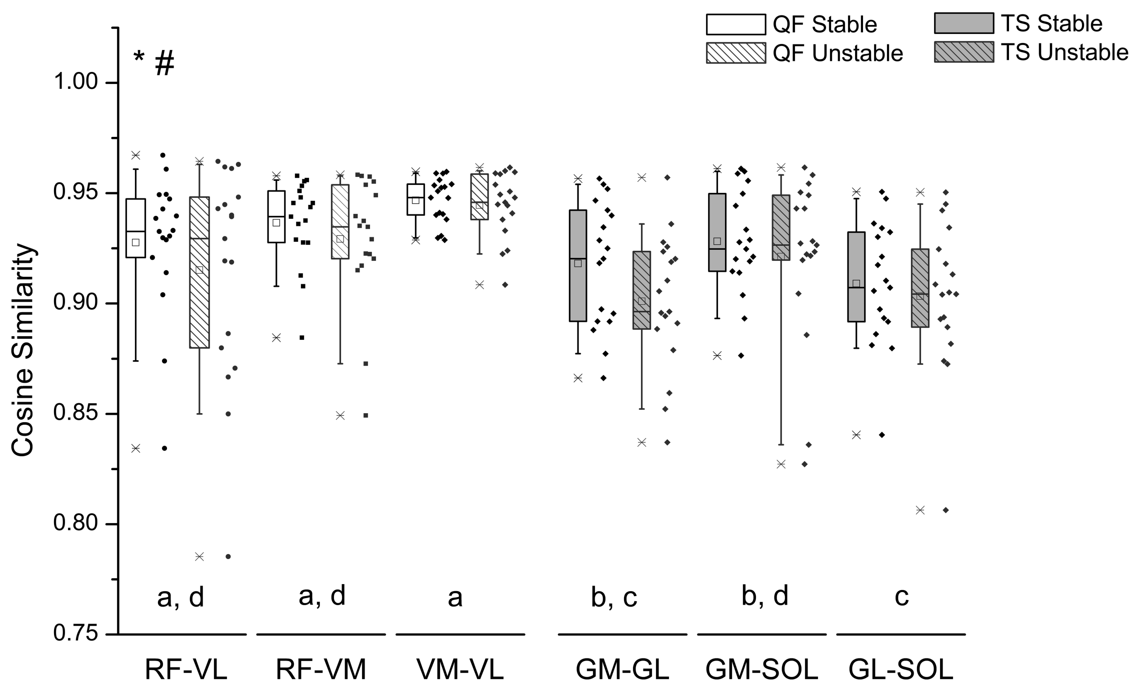

FIGURE 4
Box plot and individual data points (dots) of the cosine similarity values between the EMG activity of quadriceps femoris (QF) and triceps surae (TS) muscle pairs during postural tracking on the stable (unfilled bars) and on the unstable surface (striped bars). The center line of each box denotes the median value (50th percentile), while the box spans the interquartile range (25th to 75th percentiles). The whiskers mark the 10th and 90th percentiles. RF, rectus femoris; VL, vastus lateralis; VM, vastus medialis; GM, gastrocnemius medialis; GL, gastrocnemius lateralis; SOL, soleus. *Statistically significant differences between QF and TS muscles ($p = 0.001$). #Statistically significant differences between stable and unstable condition ($p = 0.021$). Muscle pairs sharing the same letter do not differ significantly ($p > 0.05$, *post hoc* analysis).

The quadriceps muscles are mainly involved in stabilizing the knee joint and may not be as functionally relevant as the triceps surae muscles for the chosen tracking task. During the examined postural task, quadriceps femoris muscles function primarily as damping elements to control the movement and secondarily as active components to generate force in the direction of the movement (29). It can be argued that the need for increased functional flexibility between the synergistic VL, VM and RF muscles is not as high as for the three triceps surae muscles, resulting in greater similarity in their activation patterns. The quadriceps muscles are also crucial in preventing patellofemoral pain as they control the internal stresses of the joint (23, 24). High synchronization between VM and VL minimizes mediolateral patellar force by balancing opposing forces on the patella (24). Thus, a higher similarity of the activation patterns within the quadriceps femoris muscles may represent a neural drive mechanism to protect knee joint from elevated patellofemoral joint contact stresses, which ultimately leads to pain.

We modified the surface with soft beams to create external mechanical perturbations for the participants during the tracking task. Although, there was a deterioration in tracking performance in the unstable condition, as reflected by lower CS values between target and CoP motion, all participants were still able to track the target despite the perturbations. This implies that the neuromotor system was capable of compensating for these perturbations, indicating a robust motor control. We also found a decrease in CS in both the triceps surae and quadriceps femoris muscles in the unstable condition, confirming our second hypothesis. This finding indicates adjustments in neuromotor control within the investigated synergistic muscles. The lower CS values within the triceps surae and quadriceps femoris muscles suggest increased diversity of individual muscle activation patterns and a strategy for more flexible, task-specific control of muscle activity on the unstable surface. The reduced similarity in the activation patterns of synergistic muscles may allow for greater and thus finer adjustments in force and stabilization, which are critical in unstable conditions.

Between the triceps surae muscle pairs we found different similarities in their activation patterns. The CS between GL-SOL were lower than GM-SOL, suggesting different functional contributions of the gastrocnemii during the tracking task. In stable and unstable conditions, the primary role of triceps surae muscles is to counteract forward body motion, while dorsiflexors regulate backward sway (19, 38). From the plantarflexors SOL plays a key role in postural control, acting as an antigravity muscle (39), given its larger physiological cross-sectional area (40) and predominance of type I muscle fibers (41). This stabilizing function is essential on both stable and unstable surfaces, providing sustained postural control when torque demands increase (19). The two gastrocnemii may produce different ankle moments in the frontal plane (35), and thus different patterns of activation during the tracking task may contribute to stability challenges in the mediolateral direction. It should be mentioned that in the triceps surae two biarticular muscles are involved and one in the quadriceps femoris muscle. Biarticular muscles generate moments in two joints simultaneously and can transfer energy between the two joints (42, 43). Biarticular muscles such as the gastrocnemii and rectus femoris showed greater changes in EMG-activity patterns compared to monoarticular muscles during step-down perturbations (44) and surprise loading (45) and may have a greater effect on similarity. When standing on a foam surface, the demand of postural stabilization and torque generation around the ankle joint increases due to the mechanical disturbances from surface compliance in both mediolateral and anteroposterior directions (46). Previous studies confirmed an increased EMG activity of muscles controlling anteroposterior motion, i.e., SOL and tibialis anterior, to counterbalance the demand for increased torque around the ankle joint due to the soft foam surface (19, 29). However, muscles primarily responsible for eversion, like peroneus longus, do not show this effect (29), possibly due to the anteroposterior emphasis of the task, which may allow mediolateral perturbations to be managed through control mechanisms other than increased activation, such as altered synchronization between GM-SOL and GL-SOL.

Our findings are related to the tasks investigated and the participants recruited for this study. It is possible that in more challenging tasks, such as tracking an unpredictably moving target or relying on acoustic rather than visual guidance, the effect of an unstable surface on the similarity of activation patterns within synergistic muscles could differ. Furthermore, individual skill level, training experience, and pathology may influence the degree of similarity between synergistic muscles. Nevertheless, we are confident that our main findings—reduced similarity in the triceps surae compared to the quadriceps muscles and a decrease in similarity due to surface instability—are valid. A further limitation of this study is that only a single-frequency visual target movement (0.25 Hz) was used, meaning that changes in muscle activation pattern similarity at different movement speeds were not explored. However, a previous study found that when performing the same task at a slower target speed (0.125 Hz), the examined muscles exhibited lower activation only during a small portion of the sway cycle, while the overall activation pattern remained similar across both velocities (29), indicating no relevant effects of visual target motion frequency.

In conclusion, our results show a decreased similarity in activation patterns within the synergistic triceps surae and quadriceps femoris muscle pairs on the soft surface indicating an increased flexibility of neuromotor control in the unstable condition. The lower CS in the activation patterns of triceps surae muscle pairs compared to the quadriceps femoris muscle pairs suggests a higher diversity in activation, which may increase the flexibility of neuromotor control to meet specific joint stabilization challenges during the investigated tracking task. These insights into postural control mechanisms offer potential applications for balance training and rehabilitation, where enhancing flexible motor strategies might improve stability in variable challenging environments. The critical role of the human plantar flexor muscles in controlling body stability under demanding locomotor conditions has been widely reported in the past (44, 47, 48). For example, selective EMG activity enhancement of the GM muscle has been found during drop-like perturbations (44, 48). It may be possible to influence the flexibility and diversity of synergistic muscles, in particular the triceps surae muscles, which are very important for balance performance (49, 50), by increasing the balance task complexity and difficulty using unstable conditions.

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 humans were approved by Humboldt-Universität zu Berlin, HU-KSBF-EK_2018_0013. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

LM: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Project administration, Software, Writing – original draft, Writing – review & editing. MN: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Visualization, Writing – review & editing. SB: Investigation, Methodology, Writing – review & editing. AA: Conceptualization, Formal Analysis, Methodology, Project administration, Resources, Writing – original draft, Writing – review & editing.

Funding

The author(s) declare that no financial support was received for the research, authorship, and/or publication of this article.

Acknowledgments

We acknowledge the support by the Open Access Publication Fund of the Humboldt-Universität zu Berlin.

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.

The author(s) declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision.

Generative AI statement

The author(s) declare that no Generative AI was used in the creation of this manuscript.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

References

1. Bizzi E, Cheung VCK, d'Avella A, Saltiel P, Tresch M. Combining modules for movement. *Brain Res Rev*. (2008) 57:125–33. doi: 10.1016/j.brainresrev.2007.08.004

2. Dominici N, Ivanenko YP, Cappellini G, D'Avella A, Mondì V, Cicchese M, et al. Locomotor primitives in newborn babies and their development. *Science*. (2011) 334:997–9. doi: 10.1126/science.1210617

3. Clark DJ, Ting LH, Zajac FE, Neptune RR, Kautz SA. Merging of healthy motor modules predicts reduced locomotor performance and muscle coordination complexity post-stroke. *J Neurophysiol*. (2010) 103:844–57. doi: 10.1152/jn.00825.2009

4. Brüll L, Hezel N, Arampatzis A, Schwenk M. Comparing the effects of two perturbation-based balance training paradigms in fall-prone older adults: a randomized controlled trial. *Gerontology*. (2023) 69:910–22. doi: 10.1159/000530167

5. Xu D, Zhou H, Quan W, Ma X, Chon TE, Fernandez J, et al. New insights optimize landing strategies to reduce lower limb injury risk. *Cyborg Bionic Syst*. (2024) 5:0126. doi: 10.34133/cbsystems.0126

6. Oliveria SA, Gizzi CL, Kersting UG, Farina D. Modular organization of balance control following perturbations during walking. *J Neurophysiol*. (2012) 108:1895–906. doi: 10.1152/jn.00217.2012.-Balance

7. Munoz-Martel V, Santuz A, Ekizos A, Arampatzis A. Neuromuscular organisation and robustness of postural control in the presence of perturbations. *Sci Rep*. (2019) 9:12273. doi: 10.1038/s41598-019-47613-7

8. Cappellini G, Ivanenko YP, Poppele RE, Lacquaniti F. Motor patterns in human walking and running. *J Neurophysiol*. (2006) 95:3426–37. doi: 10.1152/jn.00081.2006

9. Ting LH, Chiel HJ, Trumbower RD, Allen JL, McKay JL, Hackney ME, et al. Neuromechanical principles underlying movement modularity and their implications for rehabilitation. *Neuron*. (2015) 86:38–54. doi: 10.1016/j.neuron.2015.02.042

10. Janshen L, Santuz A, Ekizos A, Arampatzis A. Fuzziness of muscle synergies in patients with multiple sclerosis indicates increased robustness of motor control during walking. *Sci Rep*. (2020) 10:7249. doi: 10.1038/s41598-020-63788-w

11. Santuz A, Ekizos A, Kunimasa Y, Kijima K, Ishikawa M, Arampatzis A. Lower complexity of motor primitives ensures robust control of high-speed human locomotion. *Heliyon*. (2020b) 6:e05377. doi: 10.1016/j.heliyon.2020.e05377

12. Munoz-Martel V, Santuz A, Bohm S, Arampatzis A. Proactive modulation in the spatiotemporal structure of muscle synergies minimizes reactive responses in perturbed landings. *Front Bioeng Biotechnol*. (2021b) 9:761766. doi: 10.3389/fbioe.2021.761766

13. Xu D, Zhou H, Quan W, Gusztav F, Baker JS, Gu Y. Adaptive neuro-fuzzy inference system model driven by the non-negative matrix factorization-extracted muscle synergy patterns to estimate lower limb joint movements. *Comput Methods Programs Biomed*. (2023) 242:107848. doi: 10.1016/j.cmpb.2023.107848

14. Martino G, Ivanenko YP, Serrao M, Ranavolo A, Draicchio F, Conte C, et al. Locomotor patterns in cerebellar ataxia. *J Neurophysiol*. (2014) 112:2810–21. doi: 10.1152/jn.00275.2014.-Several

15. Martino G, Ivanenko YP, D'avella A, Serrao M, Ranavolo A, Draicchio F, et al. Neuromuscular adjustments of gait associated with unstable conditions. *J Neurophysiol*. (2015) 114:2867–82. doi: 10.1152/jn.00029.2015.-A

16. Santuz A, Brüll L, Ekizos A, Schroll A, Eckardt N, Kibele A, et al. Neuromotor dynamics of human locomotion in challenging settings. *iScience*. (2020a) 23:100796. doi: 10.1016/j.isci.2019.100796

17. Santuz A, Ekizos A, Eckardt N, Kibele A, Arampatzis A. Challenging human locomotion: stability and modular organisation in unsteady conditions. *Sci Rep*. (2018) 8:2740. doi: 10.1038/s41598-018-21018-4

18. Munoz-Martel V, Santuz A, Bohm S, Arampatzis A. Neuromechanics of dynamic balance tasks in the presence of perturbations. *Front Hum Neurosci*. (2021a) 14:560630. doi: 10.3389/fnhum.2020.560630

19. Mademli L, Mavridi D, Bohm S, Patikas DA, Santuz A, Arampatzis A. Standing on unstable surface challenges postural control of tracking tasks and modulates neuromuscular adjustments specific to task complexity. *Sci Rep*. (2021) 11:6122. doi: 10.1038/s41598-021-84899-y

20. Héroux ME, Dakin CJ, Luu BL, Inglis JT, Blouin J-S. Absence of lateral gastrocnemius activity and differential motor unit behavior in soleus and medial gastrocnemius during standing balance. *J Appl Physiol*. (2014) 116:140–8. doi: 10.1152/japplphysiol.00906.2013

21. Hug F, Del Vecchio A, Avrillon S, Farina D, Tucker K. Muscles from the same muscle group do not necessarily share common drive: evidence from the human triceps surae. *J Appl Physiol*. (2021) 130:342–54. doi: 10.1152/JAPPLPHYSIOL.00635.2020

22. Rossato J, Tucker K, Avrillon S, Lacourpaille L, Holobar A, Hug F. Less common synaptic input between muscles from the same group allows for more flexible coordination strategies during a fatiguing task. *J Neurophysiol*. (2022) 127:421–33. doi: 10.1152/jn.00453.2021

23. Sheehan FT, Borotikar BS, Behnam AJ, Alter KE. Alterations in *in vivo* knee joint kinematics following a femoral nerve branch block of the vastus medialis: implications for patellofemoral pain syndrome. *Clin Biomech*. (2012) 27:525–31. doi: 10.1016/j.clinbiomech.2011.12.012

24. Alessandro C, Barroso FO, Prashara A, Tentler DP, Yeh H-Y, Tresch MC, et al. Coordination amongst quadriceps muscles suggests neural regulation of internal joint stresses, not simplification of task performance. *PNAS*. (2020) 117:8135–42. doi: 10.1073/pnas.1916578117/-/DCSupplemental

25. Lai AKM, Lichtwark GA, Schache AG, Pandy MG. Differences in *in vivo* muscle fascicle and tendinous tissue behavior between the ankle plantarflexors during running. *Scand J Med Sci Sports*. (2018) 28:1828–36. doi: 10.1111/sms.13089

26. Hamard R, Aeles J, Kelp NY, Feigean R, Hug F, Dick TJM. Does different activation between the medial and the lateral gastrocnemius during walking translate into different fascicle behavior? *J Exp Biol*. (2021) 224:jeb242626. doi: 10.1242/jeb.242626

27. Mersmann F, Bohm S, Schroll A, Marzilger R, Arampatzis A. Athletic training affects the uniformity of muscle and tendon adaptation during adolescence. *J Appl Physiol*. (2016) 121:893–9. doi: 10.1152/japplphysiol.00493.2016.-With

28. Brüll L, Santuz A, Mersmann F, Bohm S, Schwenk M, Arampatzis A. Spatiotemporal modulation of a common set of muscle synergies during unpredictable and predictable gait perturbations in older adults. *J Exp Biol*. (2024) 227:jeb247271. doi: 10.1242/jeb.247271

29. Patikas DA, Papavasileiou A, Ekizos A, Hatzitaki V, Arampatzis A. Swaying slower reduces the destabilizing effects of a compliant surface on voluntary sway dynamics. *PLoS One*. (2019) 14:e0226263. doi: 10.1371/journal.pone.0226263

30. Sotirakis H, Kyvelidou A, Mademli L, Stergiou N, Hatzitaki V. Aging affects postural tracking of complex visual motion cues. *Exp Brain Res*. (2016) 234:2529–40. doi: 10.1007/s00221-016-4657-x

31. Sotirakis H, Hatzitaki V, Munoz-Martel V, Mademli L, Arampatzis A. Center of pressure feedback modulates the entrainment of voluntary sway to the motion of a visual target. *Appl Sci*. (2019) 9:3952. doi: 10.3390/app9193952

32. Jacqmin-Gadda H, Sibillot S, Proust C, Molina JM, Thiébaut R. Robustness of the linear mixed model to misspecified error distribution. *Comput Stat Data Anal*. (2007) 51:5142–54. doi: 10.1016/j.csda.2006.05.021

33. Daley MA, Biewener AA. Running Over Rough Terrain Reveals Limb control for Intrinsic Stability (2006). Available online at: www.pnas.orgcgidoi10.1073pnas.0601473103

34. Biewener AA, Daley MA. Unsteady locomotion: integrating muscle function with whole body dynamics and neuromuscular control. *J Exp Biol*. (2007) 210:2949–60. doi: 10.1242/jeb.005801

35. Lee SSM, Piazza SJ. Inversion-eversion moment arms of gastrocnemius and tibialis anterior measured *in vivo*. *J Biomech*. (2008) 41:3366–70. doi: 10.1016/j.jbiomech.2008.09.029

36. Winter DA. Human balance and posture control during standing and walking. *Gait Posture*. (1995) 3:193–214. doi: 10.1016/0966-6362(96)82849-9

37. Loram ID, Lakie M. Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for stability. *J Physiol*. (2002) 545:1041–53. doi: 10.1113/jphysiol.2002.025049

38. Loram ID, Maganaris CN, Lakie M. Paradoxical muscle movement during postural control. *Med Sci Sports Exerc*. (2009) 41:198–204. doi: 10.1249/MSS.0b013e318183c0ed

39. Paillard T. Relationship between muscle function, muscle typology and postural performance according to different postural conditions in young and older adults. *Front Physiol*. (2017) 8:585. doi: 10.3389/fphys.2017.00585

40. Albracht K, Arampatzis A, Baltzopoulos V. Assessment of muscle volume and physiological cross-sectional area of the human triceps surae muscle *in vivo*. *J Biomech*. (2008) 41:2211–8. doi: 10.1016/j.jbiomech.2008.04.020

41. Winkler T, Mersmann F, Von Roth P, Dietrich R, Bierbaum S, Arampatzis A. Development of a non-invasive methodology for the assessment of muscle fibre composition. *Front Physiol*. (2019) 10:174. doi: 10.3389/fphys.2019.00174

42. Schenau GJVI. From rotation to translation: constraints on multi-joint movements and the unique action of bi-articular muscles. *Hum Mov Sci*. (1989) 8:301–37. doi: 10.1016/0167-9457(89)90037-7

43. Arampatzis A, Kharazi M, Theodorakis C, Mersmann F, Bohm S. Biarticular mechanisms of the gastrocnemii muscles enhance ankle mechanical power and work during running. *R Soc Open Sci*. (2023) 10:230007. doi: 10.1098/rsos.230007

44. Nakazawa K, Kawashima N, Akai M, Yano H. On the reflex coactivation of ankle flexor and extensor muscles induced by a sudden drop of support surface during walking in humans. *J Appl Physiol*. (2004) 96:604–11. doi: 10.1152/japplphysiol.00670.2003.-Recent

45. McDonagh MJN, Duncan A. Interaction of pre-programmed control and natural stretch reflexes in human landing movements. *J Physiol*. (2002) 544:985–94. doi: 10.1113/jphysiol.2002.024844

46. Patel M, Fransson PA, Lush D, Gomez S. The effect of foam surface properties on postural stability assessment while standing. *Gait Posture*. (2008) 28:649–56. doi: 10.1016/j.gaitpost.2008.04.018

47. Dick TJM, Clemente CJ, Punith LK, Sawicki GS. Series elasticity facilitates safe plantar flexor muscle-tendon shock absorption during perturbed human hopping. *Proc R Soc B*. (2021) 288:20210201. doi: 10.1098/rspb.2021.0201

48. Bunz EK, Haeufle DFB, Remy CD, Schmitt S. Bioinspired preactivation reflex increases robustness of walking on rough terrain. *Sci Rep*. (2023) 13:13219. doi: 10.1038/s41598-023-39364-3

49. Dick TJM, Punith LK, Sawicki GS. Humans falling in holes: adaptations in lower-limb joint mechanics in response to a rapid change in substrate height during human hopping. *J R Soc Interface*. (2019) 16:20190292. doi: 10.1098/rsif.2019.0292

50. Golyski PR, Sawicki GS. Which lower limb joints compensate for destabilizing energy during walking in humans? *J R Soc Interface*. (2022) 19:20220024. doi: 10.1098/rsif.2022.0024

frontiers | Frontiers in Sports and Active Living

TYPE Original Research
PUBLISHED 01 April 2025
DOI 10.3389/fspor.2025.1524489

OPEN ACCESS

EDITED BY
Huub Maas,
VU Amsterdam, Netherlands

REVIEWED BY
Kuan Wang,
Shanghai University of Medicine and Health Sciences, China
Jia Han,
Shanghai University of Medicine and Health Sciences, China

*CORRESPONDENCE
Adamantios Arampatzis
a.arampatzis@hu-berlin.de

RECEIVED 07 November 2024
ACCEPTED 19 March 2025
PUBLISHED 01 April 2025

CITATION
Fischer L, Schroll A, Schmidt H and Arampatzis A (2025) Sex-specific trunk movement coordination in participants with low-back pain and asymptomatic controls. Front. Sports Act. Living 7:1524489. doi: 10.3389/fspor.2025.1524489

Sex-specific trunk movement coordination in participants with low-back pain and asymptomatic controls

Lukas Fischer[1,2], Arno Schroll[1,2], Hendrik Schmidt[3] and Adamantios Arampatzis[1,2]*

[1]Department of Training and Movement Sciences, Humboldt-Universität zu Berlin, Berlin, Germany, [2]Berlin School of Movement Science, Humboldt-Universität zu Berlin, Berlin, Germany, [3]Julius Wolff Institute, Berlin Institute of Health at Charité—Universitätsmedizin Berlin, Berlin, Germany

Background: Trunk posture and lumbo-pelvic coordination can influence spinal loading and are commonly used as clinical measures in the diagnosis and management of low-back pain and injury risk. However, sex and pain specific characteristics have rarely been investigated in a large cohort of both healthy individuals and low-back pain patients. It has also been suggested that the motor control of trunk stability and trunk movement variability is altered in individuals with low-back pain, with possible implications for pain progression. Nonetheless, clear links to low-back pain are currently lacking.

Objective: To investigate trunk posture, lumbo-pelvic coordination, trunk dynamic stability and trunk movement variability in an adequately large cohort of individuals with low-back pain and asymptomatic controls and to explore specific effects of sex, pain intensity and pain chronicity.

Methods: We measured lumbo-pelvic kinematics during trunk flexion and trunk dynamic stability and movement variability during a cyclic pointing task in 306 adults (156 females) aged between 18 and 64 years, reporting either no low-back pain or pain in the lumbar area of the trunk. Participants were grouped based on their characteristic pain intensity as asymptomatic (ASY, $N = 53$), low to medium pain (LMP, $N = 185$) or medium to high pain (MHP, $N = 68$). Participants with low-back pain that persisted for 12 weeks or longer were categorized as chronic ($N = 104$). Data were analyzed using linear mixed models in the style of a two way anova.

Results: Female participants showed a higher range of motion in both the trunk and pelvis during trunk flexion, as well as an increased lumbar lordosis in standing attributed to a higher pelvic angle that persisted throughout the entire trunk flexion movement, resulting in a longer duration of lumbar lordosis. The intensity and chronicity of the pain had a negligible effect on trunk posture and the lumbo-pelvic coordination. Pain chronicity had an effect on trunk dynamic stability (i.e., increased trunk instability), while no effects of sex and pain intensity were detected in trunk dynamic stability and movement variability.

Conclusions: Low-back pain intensity and chronicity was not associated with lumbo-pelvic posture and kinematics, indicating that lumbo-pelvic posture and kinematics during a trunk flexion movement have limited practicality in the clinical diagnosis and management of low-back pain. On the other hand, the increased local instability of the trunk during the cyclic coordination task studied indicates control errors in the regulation of trunk movement in participants with chronic low-back pain and could be considered a useful diagnostic tool in chronic low-back pain.

KEYWORDS

lumbo-pelvic rhythm, spine alignment, trunk stability, trunk variability, lordosis

1 Introduction

Low-back pain (LBP) is a leading cause of years lived with disability for 568 million people world wide (1) and is the leading health condition contributing to the need for rehabilitation services globally (2), with immense health care costs and loss of productivity regardless of the treatment indicating the need for nuanced interventions (3). Range of motion (RoM), posture and alignment of the spine during trunk flexion have often been used as key components in the clinical diagnosis of people with low-back pain (4, 5). Trunk flexion in the sagittal plane is a frequently performed movement especially in work-related activities (6) and earlier studies have associated trunk flexion movements with an increased load on the spine by measuring intradiscal pressures *in vivo* (7–9) and using musculoskeletal models (10–12). Trunk flexion involves both lumbar and pelvic tilt, and quantifying lumbo-pelvic coordination has potential applications in load and injury-risk assessment (13–16) and low-back pain therapy (17, 18). Although the lumbo-pelvic coordination during trunk flexion and extension movements has been extensively studied (19–22), determinants such as sex- or pain-specific characteristics have rarely been examined. Some studies (19) found no differences in lumbo-pelvic coordination between female and male participants with and without low-back pain, whereas others (23) reported sex-specific differences in lumbo-pelvic coordination in asymptomatic participants. A recent systematic review (24) found an effect of sex on lumbar lordosis and lumbar RoM, but noted that the studies available for synthesis were limited and more evidence was needed. Considering the importance of lumbar lordosis in spinal loading (11, 16, 25) and its possible association with low-back pain and spondylolysis (26, 27) an assessment of lumbo-pelvic kinematics in a large number of healthy participants and patients with low-back pain may be of clinical relevance.

It is generally accepted that the risk of developing low-back pain is multifactorial (28–31) and that besides psychosocial (32–34) and personal factors (35–37), a deterioration of the motor control of spinal stability may be related to the onset and progression of low-back pain (13, 38–40). Non-linear analyses can be used to characterize the motor control of trunk stability during repetitive dynamic trunk movements (41). The local dynamic stability of the trunk can be assessed from kinematic data using the maximum finite-time Lyapunov exponent (41). Positive values of this exponent show a divergence of the nearest neighbors in state space over time, while larger values reveal a faster divergence (42), thus indicating a greater effect of small perturbations in trunk kinematics. While there are indications of an association between pathological conditions and trunk movement variability (43, 44) it should be noted that movement variability may also be of functional relevance and indicate a skilled repertoire of a healthy motor system (45) that is needed to cope with perturbations (46). Investigations that incorporate the temporal structure of the movement, such as non-linear time series analyses, can provide insight into the motor control of a systems stability (39, 47) and help to understand the effect of low-back pain on movement (48). However, the effect of low-back pain on trunk dynamic stability and trunk movement variability is currently not well understood (49). Previous studies found no clear effect of low-back pain on trunk dynamic stability (50) and movement variability (51). Current reviews find insufficient evidence (51) and report inconsistent results (52) probably due to differences in methods, task demands and inclusion criteria. It has been suggested that subgroups according to pain characteristics (52, 53), analyses of possible confounders and a large number of participants included (51) may help to detect low-back specific adjustments in spinal motion, thereby increasing the clinical utility of certain measurement variables.

The purpose of the current study was to investigate trunk posture, lumbo-pelvic coordination, trunk dynamic stability and trunk movement variability in an adequately large cohort of participants with low-back pain and asymptomatic controls and to explore specific effects of sex, pain intensity and pain chronicity. Based on literature reports of sex-specific characteristics in lumbar lordosis, sacrum orientation and lumbo-pelvic rhythm (23, 54) we hypothesized that females would have higher lumbar lordosis, higher pelvic RoM and lower lumbo-pelvic ratio than males. Furthermore, based on the reported inconsistent findings of low-back pain on spinal posture and movement (55), we hypothesized that pain intensity would not affect the investigated variables, but that chronic low-back pain would, due to possible control errors in the regulation of trunk movement.

2 Methods

2.1 Experimental design

To determine an appropriate sample size for six groups (i.e., three levels of pain intensity, two levels of sex), we conducted an *a priori* power analysis (G*Power 3.1.9.7) using the outcomes in measures of lumbo-pelvic coordination from an earlier study by our group (56), where we observed medium effect sizes from 0.5 to 0.7 (Cohen's *d*) between male and female participants. Assuming a more conservative medium effect size of $f = 0.20$, an alpha error of 0.05 and a statistical power of 0.8 for a balanced group design, a total of 244 participants should suffice to detect specific differences between pain groups or interaction effects ($df = 2$) in measures of lumbo-pelvic coordination. To detect differences between male and female participants ($df = 1$), a total of 199 participants would be sufficient. Based on this power analysis and an assumed data loss ratio of at least 20% due to drop out or data quality, we recruited 306 adults for this study (156 females, 150 males, Table 1). We included participants aged between 18 and 64 years, reporting either no low-back pain or pain in the lumbar area of the trunk. Exclusion criteria were as follows: body mass index (BMI) >28, central or peripheral neurological impairments, prior spine surgery, malposition or aberration of lower extremities, pregnancy, medication with opioids or muscle relaxants, rheumatism, osteoporosis or acute infection, cardiac diseases. Study participants were recruited

TABLE 1 Participants' anthropometric characteristics.

	ASY		LMP		MHP		p-values			
	Male (N = 26)	Female (N = 27)	Male (N = 91)	Female (N = 94)	Male (N = 33)	Female (N = 35)	cLBP	Sex	Pain	Int
Age [y]	36.50 ± 10.08	38.15 ± 14.75	42.60 ± 11.02	42.02 ± 12.63	41.93 ± 14.13	39.69 ± 12.30	0.582	0.797	0.103	0.513
Height [m]*	1.80 ± 0.08	1.66 ± 0.07	1.80 ± 0.08	1.67 ± 0.06	1.81 ± 0.14	1.69 ± 0.07	0.350	**<0.001**	0.698	0.662
Mass [kg]*,‡,$	78.13 ± 11.35	62.66 ± 7.67	79.39 ± 9.79	63.48 ± 9.32	80.51 ± 12.12	68.49 ± 10.48	0.525	**<0.001**	**0.040**	0.264
BMI [kg/m^2]*,‡,$	24.07 ± 2.85	22.71 ± 2.57	24.47 ± 2.72	22.70 ± 2.97	24.90 ± 4.26	23.89 ± 3.32	0.144	**<0.001**	**0.022**	0.531
CPI†,‡,$	0.00 ± 0.00	0.00 ± 0.00	25.31 ± 12.63	26.21 ± 12.10	60.30 ± 8.71	61.62 ± 10.27	**<0.001**	0.627	**<0.001**	0.955

*Significant differences ($p < 0.05$) between male and female.
†Significant differences ($p < 0.05$) between ASY and LMP, *post-hoc* analysis.
‡Significant differences ($p < 0.05$) between ASY and MHP, *post-hoc* analysis.
$Significant differences ($p < 0.05$) between LMP and MHP, *post-hoc* analysis.
Values are presented as mean ± standard deviation.
ASY: asymptomatic no pain (CPI = 0); LMP: low to medium pain (CPI = 1–49); MHP: medium to high pain (CPI = 50–100).
p-values denote effects of pain chronicity (cLBP), sex (Sex), pain intensity (Pain) or a sex-by-pain interaction (Int).
Bold values highlight significant differences ($p < 0.05$).

within the ongoing "Berlin Back Study"—a prospective cross-sectional investigation registered with the German Clinical Trial Register (DRKS-ID: DRKS00027907, DRKS00029361). The recruitment of participants was conducted through multiple channels, including local promotion at Charité-Universitätsmedizin Berlin (via mailed flyers, notice boards, online platforms, and social media), public outreach (including newspapers, magazines, podcasts, and television), collaborations with local businesses and administrative bodies, and word-of-mouth referrals. The study protocol adheres to the ethical principles outlined in the Helsinki Declaration (57). The study follows the STROBE guidelines (58) and was approved by the Ethics Committee of the Humboldt-Universität zu Berlin (HU-KSBF-EK_2021_0006). Written informed consent was obtained from all participants.

2.2 Pain assessment

Pain intensity was assessed using the validated German version (59) of the Graded Chronic Pain Scale (60). The characteristic pain intensity (CPI) (61) was calculated as the average of 0–10 ratings of pain right now, average pain and worst pain in the last three months multiplied by 10 resulting in a score between 0 and 100. CPI scores were then used to classify all participants as either (1) asymptomatic (ASY, CPI = 0) (2), participants with low to medium pain (LMP, CPI = 1–49), or (3) participants with medium to high pain (MHP, CPI = 50–100) (60). The chronicity of low-back pain was assessed for all participants during the clinical examination conducted by an orthopedic and trauma surgery specialist. Low-back pain that persisted for 12 weeks or longer was defined as chronic (62).

2.3 Lumbo-pelvic kinematics

All participants performed three consecutive trunk forward bending movements. They were instructed to perform the movement in a slow but comfortable pace without bending their knees and until the individual maximal trunk flexion position. The lumbo-pelvic alignment and range of motion of both the trunk and pelvis were measured using two 3-dimensional accelerometers (Biovision, Berlin; size 1 × 1 × 1 cm, 500 Hz). The sensors were attached to the skin at the level of thoracic vertebrae 12 (T12) and sacral vertebrae 1 (S1) using double-sided adhesive tape (Figure 1A).

The obtained acceleration data were low pass filtered at 5 Hz using a 2nd order IIR Butterworth zero-phase filter and processed using a moving average in a time window of 500 ms. The orientation of the local coordinate system of each sensor in reference to the global space coordinate system was determined using the gravitational field. To minimize the influence of accelerations additional to the gravitational acceleration, participants were encouraged to execute the movement at a low, controlled velocity and only trials in which the norm of the gravitational vector was 1 ± 0.10 g were included in further calculations. During the trunk forward bending movement the angles of the pelvis and the trunk were measured in the sagittal plane using the orientation of the local coordinate systems of the sensors at S1 and T12 with respect to the global coordinate system. Positive values of the two angles indicate a forward-rotated orientation (with respect to the vertical), while negative values indicate a backward-rotated orientation of the pelvis or the trunk (Figure 1B). The difference between the trunk angle and the pelvic angle was used to calculate the lumbar angle with negative values indicating the magnitude of lumbar lordosis. Finally, a moving average filter was applied to the calculated angle values.

To identify start and end points of each cycle, we calculated secant slopes of the trunk angle in time increments of 100 ms. The start of a cycle was determined by the secant slope value falling below 0.1 and the end of a cycle was defined by the maximum angle value. Each cycle was time normalized to 1,000 data points and from the three performed cycles a mean trial was calculated. The range of motion of the pelvis ($pelvic_{RoM}$), the trunk ($trunk_{RoM}$) and the lumbar spine ($lumbar_{RoM}$) was calculated as the difference of the respective maximum angle

10.3389/fspor.2025.1524489

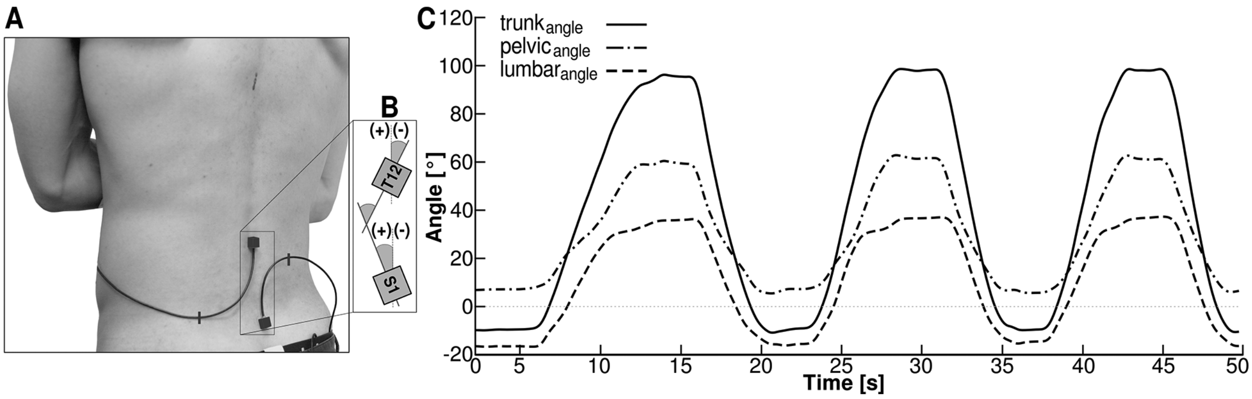

FIGURE 1
(A) Two 3-dimensional accelerometers positioned at sacral vertebrae 1 (S1) and thoracic vertebrae 12 (T12) were used to measure the pelvic ($pelvic_{angle}$) and trunk ($trunk_{angle}$) angles during three trunk flexion movements. (B) Positive values (+) indicate a forward-rotated orientation with respect to the vertical, while negative values (−) indicate a backward-rotated orientation of the pelvis or trunk. The lumbar angle ($lumbar_{angle}$) was calculated as the difference between $trunk_{angle}$ and $pelvic_{angle}$. (C) Angles of trunk, pelvis and lumbar spine during the three trunk flexion movements. The range of motion of the pelvis ($pelvic_{RoM}$), the trunk ($trunk_{RoM}$) and the lumbar spine ($lumbar_{RoM}$) was calculated as the difference of the respective maximum angle during the movement and the respective angle in the upright standing position. Values of the lumbar angle less than zero indicate the phase of lumbar lordosis while values of the lumbar angle above zero indicate the phase of lumbar kyphosis.

value during the trunk forward bending movement and the respective angle value in the upright standing position (Figure 1C). During the forward bending movement the duration of the lumbar lordosis and the duration of the lumbar lordosis normalized to the movement time was quantified from movement start to the first occurrence of a positive value in the lumbar angle. Further, we quantified the changes of the pelvic angle during the lumbar lordosis ($pelvic_{RoM-Lordosis}$), during the lumbar kyphosis ($pelvic_{RoM-Kyphosis}$) and the pelvic angle at the end of the lordotic phase to identify the orientation of the pelvis during the transition from lordosis to kyphosis. Finally, the lumbo-pelvic ratio (LPR) was calculated as the ratio of the changes in lumbar spine orientation to the changes in pelvic orientation for the whole movement (LPR_{full}) and for the lumbar lordosis ($LPR_{Lordosis}$) and kyphosis ($LPR_{Kyphosis}$) phases of the movement.

2.4 Trunk dynamic stability

The participants performed a cyclic movement that comprised trunk rotation and flexion by reaching targets positioned on the left and right side of the participant (Figure 2A). The targets were mounted on metal frames and adjusted to match the participants height at eye level and the proximal border of the patella. The frames were adjusted in width to the arm span width of the respective participant. The movement was explained and demonstrated to the participants and the movement frequency of 10 cycles per minute (0.17 Hz) was introduced by a digital metronome. After a short familiarization phase participants continuously performed a total of 32 movement cycles.

The local trunk dynamic stability was examined using the maximum finite-time Lyapunov exponent (λ_{max}). Kinematic data were collected using a 3-dimensional accelerometer (Biovision, Berlin; size 1 × 1 × 1 cm, 500 Hz) attached to the skin at the level of the second thoracic vertebrae (T2). The obtained acceleration data were low pass filtered at 20 Hz using a 2nd order IIR Butterworth zero-phase filter. We calculated the norm of the measured 3-dimensional accelerations after subtracting the respective minima. The resulting one-dimensional time-series was demeaned and a total of 30 cycles were used for subsequent analysis (Figure 2B). The time-series was time normalized to 18,000 data points (600 data points per cycle) and the reconstruction of the trunk motion in state space (Figures 2C,D) was performed using the method of delay embedding by choosing an appropriate time delay τ and embedding dimension m as follows (Equation 1):

$$S(t) = [x(t), x(t+\tau), x(t+2\tau), \ldots, x(t+(m-1)\tau)] \quad (1)$$

with $S(t)$ representing the m-dimensional reconstructed state vector, $x(t)$ the one-dimensional Euclidean norm series, τ the time delay, and m the embedding dimension. For each time series a constant time delay of $\tau = 75$ was appropriate based on average mutual information analysis (63). Global false nearest neighbor's analysis (64) revealed dimension $m = 3$ to be sufficient for the reconstruction of the current data. Finally, the maximum Lyapunov exponent was calculated as the slope of the logarithmic average divergence curve using the algorithm of Kantz (65) for the first 40 values (Figure 2E). This parameter describes the average logarithmic divergence between initially neighboring trajectories in state space. Thus, the higher the maximum Lyapunov exponent, the more unstable the system responds locally to external mechanical induced perturbations (66) or to internal motor control perturbations (67). To quantify the variability of the trunk movement, the time series of the norm of

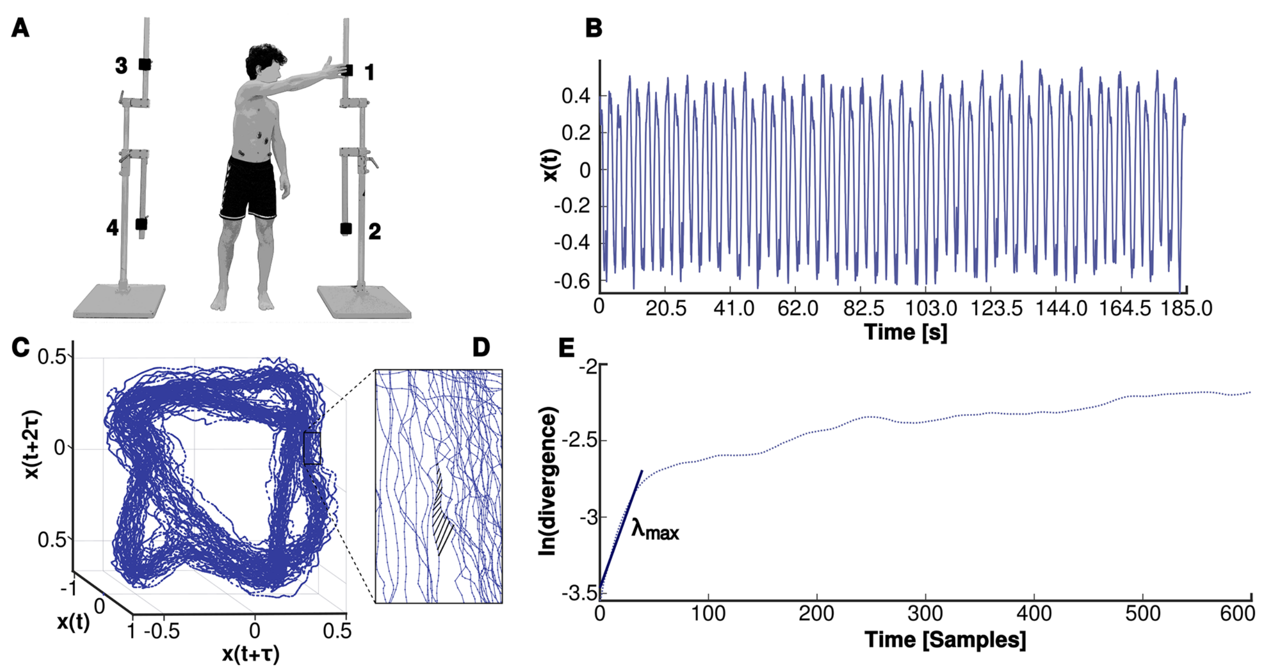

FIGURE 2
Movement task and assessment of the local dynamic stability of the trunk. **(A)** Participants repeatedly executed a pointing task from position 1 over 2 with the right arm and 3–4 with the left arm with a frequency of 10 cycles per minute (0.17 Hz); **(B)** one-dimensional time-series $x(t)$ derived from the norm of the 3-dimensional accelerations of the trunk (accelerometer at the level of the second thoracic vertebrae); **(C)** reconstructed state space of the trunk movement using dimension $m = 3$ and time delay $\tau = 75$; **(D)** diverging Euclidean distances of a nearest neighbor pair in the reconstructed state space; **(E)** average logarithmic rate (ln) of divergence of all nearest neighbor pairs over time and the maximum Lyapunov exponent (λ_{max}) as slope of the linear fit to the resulting ln(divergence) curve for 0–40 samples.

the measured 3-dimensional accelerations during all 30 cycles was used. Each cycle was time normalized to 101 data points (0%–100%) and standard deviations were calculated across all cycles at every time point. Finally, the mean variability (*MeanSD*) was calculated over all values for each trial (68).

3 Statistics

To account for possible effects of sex, pain intensity and pain chronicity on lumbo-pelvic coordination, trunk dynamic stability and trunk movement variability, we analyzed the data using linear mixed models in the style of a two way anova with interaction term with the factors sex and pain intensity and the covariate pain chronicity for the anthropometric characteristics (i.e., age, height, mass, BMI) and the kinematic parameters (i.e., movement time, $\text{trunk}_{\text{RoM}}$, $\text{pelvic}_{\text{RoM}}$, $\text{lumbar}_{\text{RoM}}$, lumbar angle at beginning and end of trunk flexion, pelvic angle at beginning and end of trunk flexion and lumbar lordosis, $\text{pelvic}_{\text{RoM}-\text{Lordosis}}$, $\text{pelvic}_{\text{RoM}-\text{Kyphosis}}$, duration of lumbar lordosis, normalized duration of lumbar lordosis, LPR_{full}, $\text{LPR}_{\text{Lordosis}}$, $\text{LPR}_{\text{Kyphosis}}$, variability, maximum Lyapunov Exponent) by using the generalized least squares (gls) approach (69) implemented in the package “nlme” (70). In the case of a significant main or interaction effect we performed a *post-hoc* analysis with controlling the false discovery rate using the approach by Benjamini and Hochberg (71). All the significance levels were set to $\alpha = 0.05$ and analyses were conducted in R (72).

4 Results

Participants showed no significant differences in age ($p = 0.103$) or body height ($p = 0.698$) between the different pain groups (Table 1). However, male participants were significantly taller ($p < 0.001$), heavier ($p < 0.001$) and had a higher BMI ($p < 0.001$) than females. In the body mass ($p = 0.040$) and BMI ($p = 0.022$) of the participants we identified a significant main effect of pain intensity. Participants with medium to high pain had a higher BMI than asymptomatic participants ($p = 0.027$) and participants with low to medium pain ($p = 0.027$). No effect of pain chronicity was detected in age ($p = 0.582$), body height ($p = 0.350$), body mass ($p = 0.525$), and BMI ($p = 0.144$) of the participants.

In the time of the forward trunk bending movement no effect of sex ($p = 0.132$), pain intensity ($p = 0.052$), pain chronicity ($p = 0.274$) or a sex-by-pain interaction ($p = 0.105$) was detected (Table 2). In the $\text{trunk}_{\text{RoM}}$ we identified a significant main effect of sex ($p < 0.001$) and pain intensity ($p = 0.009$) but no effect of pain chronicity ($p = 0.440$) or a sex-by-pain interaction ($p = 0.479$). Females demonstrated a higher $\text{trunk}_{\text{RoM}}$ compared to males ($p < 0.001$). In participants with medium to high pain

TABLE 2 Movement time ($time_{mov}$), range of motion of the trunk ($trunk_{RoM}$), pelvis ($pelvic_{RoM}$) and lumbar spine ($lumbar_{RoM}$) of participants with different pain levels.

	ASY		LMP		MHP		*p*-values			
	Male (*N* = 26)	Female (*N* = 27)	Male (*N* = 91)	Female (*N* = 94)	Male (*N* = 33)	Female (*N* = 35)	cLBP	Sex	Pain	Int
$time_{mov}$ [s]	8.38 ± 1.67	7.95 ± 2.43	8.41 ± 2.96	9.28 ± 3.01	8.04 ± 2.56	8.09 ± 2.78	0.274	0.132	0.052	0.105
$trunk_{RoM}$ [°]*,‡	110.81 ± 12.45	117.94 ± 13.93	103.71 ± 17.09	114.59 ± 14.61	101.15 ± 17.47	109.45 ± 17.32	0.440	**<0.001**	**0.009**	0.479
$pelvic_{RoM}$ [°]*	68.79 ± 13.79	75.74 ± 16.92	64.90 ± 16.19	76.80 ± 16.82	64.07 ± 16.91	68.48 ± 13.77	0.998	**<0.001**	0.068	0.223
$lumbar_{RoM}$ [°]	42.03 ± 12.01	42.20 ± 12.36	38.81 ± 12.80	37.79 ± 13.22	37.08 ± 11.33	40.97 ± 12.75	0.346	0.594	0.052	0.675

*Significant differences ($p < 0.05$) between male and female.
‡Significant differences ($p < 0.05$) between ASY and MHP, *post-hoc* analysis.
Values are presented as mean ± standard deviation.
ASY: asymptomatic no pain (CPI = 0); LMP: low to medium pain (CPI = 1–49); MHP: medium to high pain (CPI = 50–100).
p-values denote effects of pain chronicity (cLBP), sex (Sex), pain intensity (Pain) or a sex-by-pain interaction (Int).
Bold values highlight significant differences ($p < 0.05$).

TABLE 3 Angle of the lumbar spine in the upright standing position and at the end of the movement, pelvic angle during standing, at the end of the lumbar lordosis and at the end of the movement, trunk angle during standing and at the end of the movement, changes in pelvic angle during the phases of lumbar lordosis ($pelvic_{ROM-lordosis}$) and lumbar kyphosis ($pelvic_{ROM-kyphosis}$), mean duration of the lumbar lordosis (LOR_{dur}), mean duration of the lumbar lordosis relative to the movement time ($LOR_{dur-norm}$) for participants with different pain levels.

	ASY		LMP		MHP		*p*-values			
	Male (*N* = 26)	Female (*N* = 27)	Male (*N* = 91)	Female (*N* = 94)	Male (*N* = 33)	Female (*N* = 35)	cLBP	Sex	Pain	Int
lumbar angle stand [°]*,†,§	−20.30 ± 10.26	−29.38 ± 16.14	−17.63 ± 10.54	−23.84 ± 13.78	−19.38 ± 9.91	−29.91 ± 12.30	0.921	**<0.001**	**0.002**	0.437
lumbar angle end [°]*	21.72 ± 8.76	12.82 ± 9.39	21.18 ± 11.33	13.95 ± 10.07	17.71 ± 11.34	11.06 ± 10.42	0.217	**<0.001**	0.203	0.654
pelvic angle stand [°]*,†	7.89 ± 8.07	15.66 ± 13.58	6.30 ± 9.94	11.97 ± 11.66	7.52 ± 8.63	17.47 ± 9.45	0.157	**<0.001**	**0.008**	0.364
pelvic angle lordosis [°]*	36.61 ± 19.53	59.85 ± 27.63	33.96 ± 21.32	52.13 ± 27.52	38.01 ± 19.09	56.43 ± 25.06	0.931	**<0.001**	0.201	0.945
pelvic angle end [°]*,‡	76.68 ± 12.64	91.40 ± 13.78	71.20 ± 15.47	88.77 ± 14.85	71.59 ± 13.72	85.95 ± 16.07	0.340	**<0.001**	**0.036**	0.654
trunk angle stand [°]	−12.42 ± 4.57	−13.72 ± 6.85	−11.34 ± 6.49	−11.87 ± 6.58	−11.85 ± 4.79	−12.44 ± 7.66	**0.042**	0.292	0.252	0.888
trunk angle end [°]*,‡	98.40 ± 11.10	104.22 ± 12.59	92.38 ± 16.39	102.72 ± 14.68	89.30 ± 18.04	97.01 ± 15.31	0.947	**<0.001**	**0.003**	0.400
$pelvic_{RoM-Lordosis}$ [°]*	28.72 ± 15.45	41.49 ± 24.28	27.00 ± 16.22	38.76 ± 25.14	29.40 ± 16.76	38.55 ± 20.00	0.399	**<0.001**	0.583	0.840
$pelvic_{RoM-Kyphosis}$ [°]§	40.07 ± 18.58	37.32 ± 29.76	38.80 ± 21.79	39.69 ± 27.57	36.45 ± 22.69	31.04 ± 20.88	0.322	0.391	**0.027**	0.733
LOR_{dur} [s]*	2.95 ± 1.16	3.88 ± 2.57	3.05 ± 1.85	4.40 ± 3.01	3.11 ± 1.79	4.09 ± 2.23	0.234	**<0.001**	0.299	0.616
$LOR_{dur-norm}$ [%]*	35.78 ± 12.92	52.45 ± 21.09	38.89 ± 19.64	48.21 ± 23.96	43.34 ± 21.38	52.26 ± 22.25	0.837	**<0.001**	0.103	0.590

*Significant differences ($p < 0.05$) between male and female.
†Significant differences ($p < 0.05$) between ASY and LMP, *post-hoc* analysis.
‡Significant differences ($p < 0.05$) between ASY and MHP, *post-hoc* analysis.
§Significant differences ($p < 0.05$) between LMP and MHP, *post-hoc* analysis.
Values are presented as mean ± standard deviation.
ASY: asymptomatic no pain (CPI = 0); LMP: low to medium pain (CPI = 1–49); MHP: medium to high pain (CPI = 50–100).
p-values denote effects of pain chronicity (cLBP), sex (Sex), pain intensity (Pain) or a sex-by-pain interaction (Int).
Bold values highlight significant differences ($p < 0.05$).

$trunk_{RoM}$ was significantly reduced when compared to asymptomatic participants ($p = 0.006$). We did not detect statistically significant differences in $trunk_{RoM}$ between participants with low to medium pain and participants with medium to high pain ($p = 0.087$). In the $pelvic_{RoM}$ we detected a significant main effect of sex ($p < 0.001$) but no effect of pain intensity ($p = 0.069$), pain chronicity ($p = 0.998$) or a sex-by-pain interaction ($p = 0.223$). Females demonstrated a higher $pelvic_{RoM}$ compared to males ($p < 0.001$). In the $lumbar_{RoM}$ we did not find any significant main effect of sex ($p = 0.594$), pain intensity ($p = 0.052$), pain chronicity ($p = 0.346$) or a sex-by-pain interaction ($p = 0.675$). The lumbar angle in the upright standing position demonstrated a significant main effect of sex ($p < 0.001$) and pain intensity ($p = 0.002$), but no effect of pain chronicity ($p = 0.921$) or a sex-by-pain interaction ($p = 0.437$; Table 3). Female participants had a significantly greater lumbar angle in the upright standing position (i.e., higher lumbar lordosis; $p < 0.001$). Participants with low to medium pain showed a significantly reduced lumbar angle in the upright standing position (i.e., smaller lumbar lordosis) when compared to asymptomatic participants ($p = 0.049$) and participants with medium to high pain ($p = 0.049$). We did not detect statistically significant differences in the lumbar angle during upright standing between participants with medium to high pain and asymptomatic participants ($p = 0.929$). In the lumbar angle at the end of the movement we detected a significant main effect of sex ($p < 0.001$) but no effect of pain intensity ($p = 0.203$), pain chronicity ($p = 0.217$) or a or a sex-by-pain interaction ($p = 0.654$). Females showed a significantly smaller lumbar angle at the end of the movement (i.e., less kyphosis) than males ($p < 0.001$).

In the pelvic angle during the upright standing position we identified a significant main effect of sex ($p < 0.001$) and pain intensity ($p = 0.008$), but no effect of pain chronicity ($p = 0.157$) or a sex-by-pain interaction ($p = 0.364$). Female participants had a significantly higher pelvic angle in the upright standing position ($p < 0.001$). In female participants with low to medium pain the pelvic angle in the upright standing position was significantly smaller than in asymptomatic female participants ($p = 0.013$) and female participants with medium to high pain ($p = 0.047$). In the pelvic angle at the end of the lordotic phase we identified a significant main effect of sex ($p < 0.001$) but no effect of pain intensity ($p = 0.201$), pain chronicity ($p = 0.931$) or a sex-by-pain interaction ($p = 0.945$). Female participants showed significantly higher pelvic angles at the end of the lordotic phase than male participants ($p < 0.001$). In the pelvic angle at the end of the movement we identified a significant main effect of sex ($p < 0.001$) and pain intensity ($p = 0.036$), but no effect of pain chronicity ($p = 0.340$) or a sex-by-pain interaction ($p = 0.654$). Female participants showed significantly higher pelvic angles at the end of the movement than male participants ($p < 0.001$). Participants with medium to high pain showed a significantly reduced pelvic angles at the end of the movement when compared to asymptomatic participants ($p = 0.031$).

In the trunk angle during the upright standing position we identified a significant effect of pain chronicity ($p = 0.042$), but no effect of sex ($p = 0.292$), pain intensity ($p = 0.252$) or a sex-by-pain interaction ($p = 0.888$). Participants with chronic pain had a significantly lower trunk angle in the upright standing position ($p = 0.042$). In the trunk angle at the end of the movement we identified a significant main effect of sex ($p < 0.001$) and pain intensity ($p = 0.003$), but no effect of pain chronicity ($p = 0.947$) or a sex-by-pain interaction ($p = 0.400$). Female participants showed significantly higher trunk angles at the end of the movement than male participants ($p < 0.001$). Participants with medium to high pain showed a significantly reduced trunk angle at the end of the movement when compared to asymptomatic participants ($p = 0.011$).

In pelvic$_{RoM}$ during the phase of lumbar lordosis we found a significant main effect of sex ($p < 0.001$), but no effect of pain intensity ($p = 0.068$), pain chronicity ($p = 0.339$) or a sex-by-pain interaction ($p = 0.840$). Females demonstrated significantly higher values in pelvic$_{RoM}$ during the phase of lumbar lordosis than males ($p < 0.001$). In pelvic$_{RoM}$ during the phase of lumbar kyphosis we detected a significant main effect of pain intensity ($p = 0.027$), but no effects of sex ($p = 0.391$), pain chronicity ($p = 0.322$) or a sex-by-pain interaction ($p = 0.733$). *post-hoc* analysis revealed no significant differences between the different pain groups. In the duration of lumbar lordosis and the duration of lumbar lordosis normalized to the movement time we detected a significant main effect of sex ($p < 0.001$), but no effect of pain intensity ($p = 0.299$ and $p = 0.103$), pain chronicity ($p = 0.234$ and $p = 0.837$) or a sex-by-pain interaction ($p = 0.616$ and $p = 0.590$). Females were significantly longer in a lordotic position when compared to males ($p < 0.001$).

In the LPR we detected a significant main effect of sex ($p < 0.001$) during the phase of the lumbar kyphosis, with females demonstrating a lower LPR during this phase of the movement (Table 4, Figure 3). No effect of pain intensity ($p > 0.05$), pain chronicity ($p > 0.05$) or a sex-by-pain interaction ($p > 0.05$) was evident in LPR$_{full}$, LPR$_{Lordosis}$ and LPR$_{Kyphosis}$. Finally, in movement variability and the maximum Lyapunov Exponent during the cyclic pointing task no effect of sex ($p = 0.822$ and $p = 0.543$), pain intensity ($p = 0.214$ and $p = 0.719$) or a sex-by-pain interaction ($p = 0.812$ and $p = 0.661$) was detected. However, in the maximum Lyapunov Exponent we found a significant effect of pain chronicity ($p = 0.008$), with higher values in participants with chronic pain. In movement variability no effect of pain chronicity was detected ($p = 0.247$).

5 Discussion

In this study we investigated lumbo-pelvic coordination, dynamic stability and movement variability of the trunk in a large sample of participants with low-back pain and asymptomatic controls. We identified sex-specific characteristics in measures of lumbo-pelvic coordination while pain intensity and pain chronicity only had a minor impact. Pain chronicity had an effect on trunk dynamic stability, while variability and dynamic stability of the trunk were not affected by sex and pain intensity.

During the trunk forward bending movement, female participants had a higher lumbar angle (i.e., a higher lumbar lordosis) in the upright standing position due to a higher pelvic angle. The higher pelvic angle in females was maintained throughout the entire movement, resulting in a longer duration

TABLE 4 Lumbo-pelvic ratio of the full movement (LPR$_{full}$), during the phase of lumbar lordosis (LPR$_{Lordosis}$) and lumbar kyphosis (LPR$_{Kyphosis}$), variability of the trunk (VAR) and maximum lyapunov exponent (MLE) of the participants with different pain levels.

	ASY		LMP		MHP		*p*-values			
	Male (*N* = 26)	Female (*N* = 27)	Male (*N* = 91)	Female (*N* = 94)	Male (*N* = 33)	Female (*N* = 35)	cLBP	Sex	Pain	Int
LPR$_{full}$	0.67 ± 0.35	0.61 ± 0.29	0.66 ± 0.35	0.54 ± 0.30	0.67 ± 0.51	0.63 ± 0.28	0.273	0.051	0.226	0.776
LPR$_{Lordosis}$	0.84 ± 0.57	0.93 ± 0.47	0.85 ± 0.60	1.12 ± 1.52	1.11 ± 1.22	1.01 ± 0.64	0.109	0.410	0.939	0.440
LPR$_{Kyphosis}$*	0.79 ± 0.71	0.46 ± 0.30	0.78 ± 0.71	0.48 ± 0.50	0.65 ± 0.45	0.50 ± 0.35	0.054	**<0.001**	0.945	0.676
VAR [m/s^2]	0.38 ± 0.05	0.37 ± 0.06	0.38 ± 0.05	0.37 ± 0.06	0.36 ± 0.06	0.36 ± 0.07	0.247	0.822	0.214	0.812
MLE	2.44 ± 0.35	2.53 ± 0.42	2.54 ± 0.36	2.55 ± 0.38	2.52 ± 0.39	2.56 ± 0.31	**0.008**	0.543	0.719	0.661

*Significant differences ($p < 0.05$) between male and female.
Values are presented as mean ± standard deviation.
ASY: asymptomatic no pain (CPI = 0); LMP: low to medium pain (CPI = 1–49); MHP: medium to high pain (CPI = 50–100).
p-values denote effects of pain chronicity (cLBP), sex (Sex), pain intensity (Pain) or a sex-by-pain interaction (Int).
Bold values highlight significant differences ($p < 0.05$).

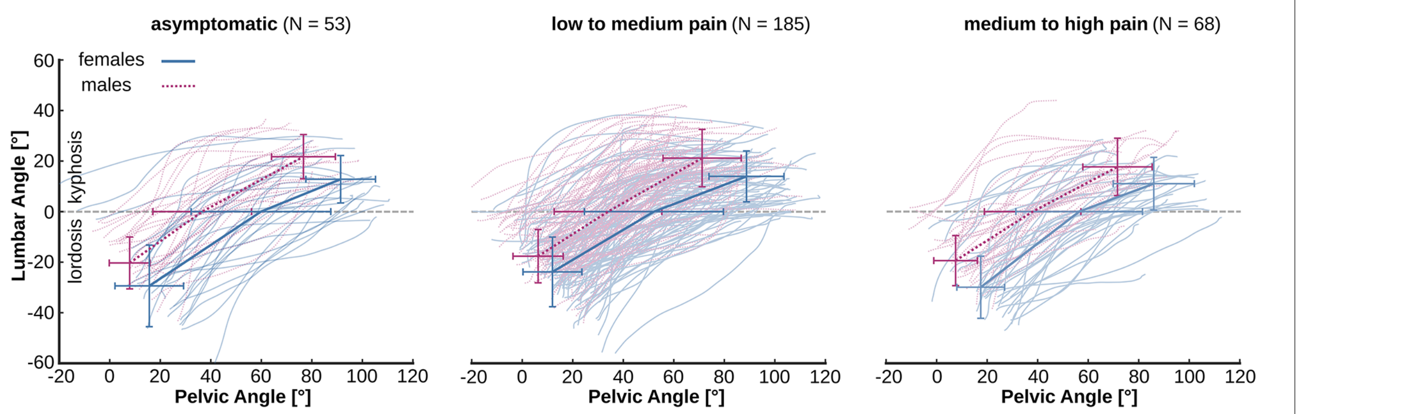

FIGURE 3
Individual lumbar angle as a function of pelvic angle for participants with different low-back pain intensities during trunk flexion. Crosses represent means ± standard deviations at upright standing, end of lumbar lordosis and end of movement, respectively. The average slope of each segment represents the corresponding lumbo-pelvic ratio.

of the lumbar lordosis. The fact that the lumbo-pelvic ratio did not differ between females and males during the lordotic phase suggests that the relative position of the pelvis and trunk is maintained in both females and males during this phase. Low-back pain had no effect on pelvic angle or lumbar lordosis, which indicates that the identified lumbo-pelvic posture and kinematics were sex-specific. A similar sex-specific behavior of the lumbo-pelvic coordination during trunk forward bending has been reported for healthy adults (23) and adolescent athletes (56). The lower lumbo-pelvic ratio found in females during the phase of lumbar kyphosis was the reason for a tendency towards a lower lumbo-pelvic ratio in females during the full range of motion ($p = 0.051$). When comparing asymptomatic males and females during a full trunk flexion, females generally displayed a smaller lumbo-pelvic ratio than males (23, 54, 73), mainly due to a greater pelvic range of motion and a greater angle of sacrum orientation in standing and full flexion. Sex-specific differences in sacral shape and orientation (74) or hamstring flexibility (75) might affect the lumbo-pelvic kinematics resulting in a higher pelvic angle in females. It has been argued that in female participants lower muscular strength capacities (76–79) and sex-specific characteristics in muscle morphology (80, 81) may also affect lumbar lordosis and pelvic inclination (26). However, several studies failed to support a relationship between trunk muscle strength, lumbar lordosis and pelvic inclination (56, 82, 83) suggesting that muscle strength may not be the reason for the sex-specific higher lumbar lordosis. The transmission of muscular forces and the resulting stability of the pelvis is dependent on the ligamentous network (84–86) and there is evidence that the pre-tension of the sacrotuberous ligament differ between males and females (87). Hence, sex-specific variations in the pre-tension of the sacrotuberous ligament may have the potential to influence the posture and kinematics of the lumbo-pelvic region (88–90).

A lordotic posture and higher pelvic rotation is associated with an increased loading of the spine (11, 16, 25). Arjmand and Shirazi-Adl (25) reported that lordotic postures increased pelvic rotation and trunk extensor muscle forces with concomitant increases in spine loading. The found effects of sex on lumbar lordosis and pelvic motion may increase trunk loading and can be interpreted as a possible risk factor for a low-back injury in females. Epidemiological studies reporting a higher prevalence of low-back pain in females (91–93) support the higher risk of a low-back injury in females. However, in our results, low-back pain intensity was not associated with lumbo-pelvic posture and kinematics (i.e., no effects of pain on lumbar lordosis, pelvic RoM, duration of lumbar lordosis, lumbo-pelvic ratio), indicating that lumbo-pelvic posture and kinematics may not be suitable measures to differentiate low-back pain patients and may not be a clear risk factor for a low-back injury. Participants with medium to high pain showed a reduced trunk RoM compared to asymptomatic controls, due to a lower trunk angle at the end of the forward flexion, but without differences in lumbar lordosis or pelvic rotation. Similarly, other studies (53, 55) observed no differences in lumbar lordosis and pelvic angle between participants with and without low-back pain. Some previous studies have reported an effect of low-back pain on the lumbo-pelvic ratio, with a relatively greater lumbar contribution during full trunk flexion in participants with low-back pain (19, 22, 94). When classifying patients with low-back pain into specific subgroups, Kim et al. (21) found differences in the lumbo-pelvic ratio between healthy participants and specific subgroups of low-back pain. It has therefore been suggested, that the relative contributions of the lumbar spine and the pelvis to a flexion movement may be of clinical relevance (55) with applications in diagnostics (19, 21, 22, 95), load and injury-risk estimation (13–16) and therapy of low-back pain (17, 18). When pooling data from previous experimental studies in a meta-analytic approach, Laird et al. (55), similar to our study with the large number of participants, found no significant differences in lumbo-pelvic coordination for participants with and without low-back pain. In our study, participants with low to medium pain intensity had a reduced lumbar spine angle when standing upright compared to both asymptomatic controls and participants with medium to high pain intensity. However, the participants with medium to high pain did not differ from the asymptomatic controls, so the results did not show a consistent effect of pain intensity on lumbar spine alignment. We argue that, besides the influence of lumbo-pelvic posture and kinematics on spinal loading (11), it remains unclear whether these variables can influence the prevalence of low-back pain.

We did not detect any significant main effect of sex and pain intensity on trunk dynamic stability and trunk movement variability during the used movement task. There are indications of differences in the variability of trunk movements between people with low-back pain and asymptomatic controls (43, 44). However, the characterization of these modifications is currently not well understood (49). Heterogeneous metrics and inclusion criteria limit the quality of evidence, making it difficult to draw conclusive inferences. Several previous studies also reported no main effect of low-back pain on the spatial variability and local dynamic stability of trunk kinematics during repetitive reaching tasks (39, 96). When reducing pain intensity by implementing an exercise therapy, Arampatzis et al. (50) reported unchanged local dynamic stability of trunk motion despite a significant reduction in low-back pain. Recent reviews, however, show inconsistent results (52) reporting differences (97) or no differences (51) in movement variability between participants with low-back pain and asymptomatic controls. It appears that both the variability and the local dynamic stability of trunk movement may not be sensitive enough to detect an association with low-back pain intensity. Although pain intensity did not affect local dynamic stability, we found higher values of the short-term Lyapunov exponent in chronic low-back-pain patients, indicating higher local trunk instability in these participants. It seems that chronic pain, in our case at least 12 weeks of continuous pain, may induce internal control errors in the regulation of trunk movement during cyclic coordinative tasks involving flexion-extension and rotation, worsening the local dynamic stability of the trunk. These findings indicate that non-linear analysis of trunk movement kinematics may be a useful tool in chronic low-back pain patients and could be considered in the clinical diagnosis and management of low-back-pain.

There are some limitations associated with the approach used in the current study. It is worth noting that with the accelerometers in our study we estimated the sacrum orientation as part of the pelvic girdle and not the pelvic tilt or the curvature of the lumbar spine directly. Still, our results are in agreement with those from previous studies that assessed lumbo-pelvic coordination using noninvasive methods (11, 20, 22, 56, 98) and it has been argued that sacral and pelvic ranges of motion can be assumed to be equal (23). Further, it has been demonstrated that the curvature measured on the back surface significantly correlates with angles measured from x-rays (99) and provides a reasonable accurate measurement of the total lumbar motion (100). To further account for possible differences between sensor position and subdermal anatomical structures, subjects with a BMI >28.0 kg/m^2 were excluded, yet limiting our results to only a subgroup of the general population. Finally, it can be argued that the age of the participants may influence lumbo-pelvic kinematics. We further analyzed our data regarding participants age and detected an effect of age in measures of lumbo-pelvic kinematics such as the lumbar angle during upright standing (i.e., lumbar lordosis, $p = 0.039$), the lumbo-pelvic ratio (LPR) during the full movement ($p < 0.001$) and the phase of lumbar kyphosis ($p = 0.006$), range of motion of the lumbar spine ($p < 0.001$) and the pelvis during full movement ($p = 0.004$) and during the phase of lumbar lordosis ($p = 0.032$). Importantly, the previously detected main effects of sex, pain intensity and pain chronicity did remain. The findings show, as also reported earlier (23, 101), that age may influence lumbo-pelvic kinematics.

In summary this current study highlights the effects of sex and low-back pain intensity on lumbo-pelvic coordination and trunk kinematics in standing and during trunk flexion. The presented results emphasize sex-specific characteristics in measures of lumbo-pelvic coordination while pain intensity and pain chronicity appear to have a minor impact. Our results suggest that lumbo-pelvic posture and kinematics during a trunk flexion movement have limited practicality in the clinical diagnosis and management of low-back pain. The increased local instability of the trunk during the cyclic coordination task studied indicates control errors in the regulation of trunk movement in participants with chronic low-back pain and could be considered a useful diagnostic tool in chronic low-back pain.

Data availability statement

The datasets presented in this study can be found in online repositories. The names of the repository/repositories and accession number(s) can be found below: https://zenodo.org/records/15044691.

Ethics statement

The studies involving humans were approved by Ethics Committee of the Humboldt-Universität zu Berlin (HU-KSBF-EK_2021_0006). The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

LF: Conceptualization, Methodology, Formal analysis, Investigation, Data curation, Writing – original draft, Writing – review & editing, Project administration. AS: Data curation, Formal analysis, Investigation, Methodology, Project administration, Software, Writing – review & editing. HS: Funding acquisition, Project administration, Writing – review & editing. AA: Conceptualization, Formal analysis, Funding acquisition, Methodology, Project administration, Resources, Supervision, Writing – original draft, Writing – review & editing.

Funding

The author(s) declare that financial support was received for the research and/or publication of this article. This study is part of the Research Unit 5177 and was financed by the German Research Foundation (DFG, AR 809/2-1, SCHM 2572/12-1, SCHM 2572/11-1). The funders had no role in the study design, data collection and analysis, decision to publish or preparation of the manuscript. Open Access funding was enabled by the Open Access Publication Fund of Humboldt-Universität zu Berlin.

Acknowledgments

We acknowledge the support by Theresa Ahnert and Barbara Schiller for the organization of the recruitment.

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.

The author(s) declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision.

Generative AI statement

The author(s) declare that no Generative AI was used in the creation of this manuscript.

Publisher's note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

References

1. GBD 2017 Disease and Injury Incidence and Prevalence Collaborators. Global, regional, and national incidence, prevalence, and years lived with disability for 354 diseases and injuries for 195 countries and territories, 1990–2017: a systematic analysis for the Global Burden of Disease Study 2017. *Lancet*. (2018) 392(10159):1789–858. doi: 10.1016/S0140-6736(18)32279-7

2. Cieza A, Causey K, Kamenov K, Hanson SW, Chatterji S, Vos T. Global estimates of the need for rehabilitation based on the Global Burden of Disease Study 2019: a systematic analysis for the Global Burden of Disease Study 2019. *Lancet*. (2020) 396(10267):2006–17. doi: 10.1016/S0140-6736(20)32340-0

3. Solumsmoen S, Poulsen G, Kjellberg J, Melbye M, Munch TN. The impact of specialised treatment of low back pain on health care costs and productivity in a nationwide cohort. *EClinicalMedicine*. (2022) 43:101247. doi: 10.1016/j.eclinm.2021.101247

4. Aksekili MAE. Low back pain. In: Longo UG, Denaro V, editors. *Textbook of Musculoskeletal Disorders*. Cham: Springer International Publishing (2023). p. 329–37. doi: 10.1007/978-3-031-20987-1_37

5. Al Ameri K, Al Memari S. Performing back physical examination. In: Lari S, Al Memari S, Al Marzooqi D, editors. *Family Medicine OSCE: First Aid to Objective Structured Clinical Examination*. Singapore: Springer Nature (2024). p. 299–300. doi: 10.1007/978-981-99-5530-5_78

6. Jansen JP, Burdorf A, Steyerberg E. A novel approach for evaluating level, frequency and duration of lumbar posture simultaneously during work. *Scand J Work Environ Health*. (2001) 27(6):373–80. doi: 10.5271/sjweh.629

7. Nachemson A. The effect of forward leaning on lumbar intradiscal pressure. *Acta Orthop Scand*. (1965) 35(1–4):314–28. doi: 10.3109/17453676508989362

8. Nachemson AL. Disc pressure measurements. *Spine*. (1981) 6(1):93–7. doi: 10.1097/00007632-198101000-00020

9. Wilke H, Neef P, Hinz B, Seidel H, Claes L. Intradiscal pressure together with anthropometric data–a data set for the validation of models. *Clin Biomech*. (2001) 16(Suppl 1):S111–126. doi: 10.1016/S0268-0033(00)00103-0

10. Takahashi I, Kikuchi S-i, Sato K, Sato N. Mechanical load of the lumbar spine during forward bending motion of the trunk-a biomechanical study. *Spine*. (2006) 31(1):18–23. doi: 10.1097/01.brs.0000192636.69129.fb

11. Tafazzol A, Arjmand N, Shirazi-Adl A, Parnianpour M. Lumbopelvic rhythm during forward and backward sagittal trunk rotations: combined *in vivo* measurement with inertial tracking device and biomechanical modeling. *Clin Biomech*. (2014) 29(1):7–13. doi: 10.1016/j.clinbiomech.2013.10.021

12. Liu T, Khalaf K, Adeeb S, El-Rich M. Effects of lumbo-pelvic rhythm on trunk muscle forces and disc loads during forward flexion: a combined musculoskeletal and finite element simulation study. *J Biomech*. (2019) 82:116–23. doi: 10.1016/j.jbiomech.2018.10.009

13. Cholewicki J, McGill SM. Mechanical stability of the *in vivo* lumbar spine: implications for injury and chronic low back pain. *Clin Biomech*. (1996) 11(1):1–15. doi: 10.1016/0268-0033(95)00035-6

14. Arjmand N, Plamondon A, Shirazi-Adl A, Larivière C, Parnianpour M. Predictive equations to estimate spinal loads in symmetric lifting tasks. *J Biomech*. (2011) 44(1):84–91. doi: 10.1016/j.jbiomech.2010.08.028

15. Firouzabadi A, Arjmand N, Zhang T, Pumberger M, Schmidt H. Effect of low back pain on the kinetics and kinematics of the lumbar spine—a combined *in vivo* and in silico investigation. *J Biomech*. (2024) 164:111954. doi: 10.1016/j.jbiomech.2024.111954

16. Ghezelbash F, Shirazi-Adl A, Arjmand N, El-Ouaaid Z, Plamondon A, Meakin JR. Effects of sex, age, body height and body weight on spinal loads: sensitivity analyses in a subject-specific trunk musculoskeletal model. *J Biomech*. (2016) 49(14):3492–501. doi: 10.1016/j.jbiomech.2016.09.026

17. Ikeda DM, McGill SM. Can altering motions, postures, and loads provide immediate low back pain relief: a study of 4 cases investigating spine load, posture, and stability. *Spine*. (2012) 37(23):E1469–1475. doi: 10.1097/BRS.0b013e31826c97e5

18. O'Sullivan P. Diagnosis and classification of chronic low back pain disorders: maladaptive movement and motor control impairments as underlying mechanism. *Man Ther*. (2005) 10(4):242–55. doi: 10.1016/j.math.2005.07.001

19. Esola MA, McClure PW, Fitzgerald GK, Siegler S. Analysis of lumbar spine and hip motion during forward bending in subjects with and without a history of low back pain. *Spine*. (1996) 21(1):71–8. doi: 10.1097/00007632-199601010-00017

20. Granata KP, Sanford AH. Lumbar-pelvic coordination is influenced by lifting task parameters. *Spine*. (2000) 25(11):1413–8. doi: 10.1097/00007632-200006010-00014

21. Kim MH, Yi CH, Kwon OY, Cho SH, Cynn HS, Kim YH, et al. Comparison of lumbopelvic rhythm and flexion-relaxation response between 2 different low back pain subtypes. *Spine*. (2013) 38(15):1260–7. doi: 10.1097/BRS.0b013e318291b502

22. Porter JL, Wilkinson A. Lumbar-hip flexion motion. A comparative study between asymptomatic and chronic low back pain in 18- to 36-year-old men. *Spine*. (1997) 22(13):1508–14. doi: 10.1097/00007632-199707010-00017

23. Pries E, Dreischarf M, Bashkuev M, Putzier M, Schmidt H. The effects of age and gender on the lumbopelvic rhythm in the sagittal plane in 309 subjects. *J Biomech*. (2015) 48(12):3080–7. doi: 10.1016/j.jbiomech.2015.07.030

24. Arshad R, Pan F, Reitmaier S, Schmidt H. Effect of age and sex on lumbar lordosis and the range of motion. A systematic review and meta-analysis. *J Biomech*. (2019) 82:1–19. doi: 10.1016/j.jbiomech.2018.11.022

25. Arjmand N, Shirazi-Adl A. Biomechanics of changes in lumbar posture in static lifting. *Spine*. (2005) 30(23):2637–48. doi: 10.1097/01.brs.0000187907.02910.4f

26. Been E, Kalichman L. Lumbar lordosis. *Spine J*. (2014) 14(1):87–97. doi: 10.1016/j.spinee.2013.07.464

27. Christie HJ, Kumar S, Warren SA. Postural aberrations in low back pain. *Arch Phys Med Rehabil*. (1995) 76(3):218–24. doi: 10.1016/S0003-9993(95)80604-0

28. Kerr MS, Frank JW, Shannon HS, Norman RW, Wells RP, Neumann WP, et al. Biomechanical and psychosocial risk factors for low back pain at work. *Am J Public Health*. (2001) 91(7):1069–75. doi: 10.2105/ajph.91.7.1069

29. Parreira P, Maher CG, Steffens D, Hancock MJ, Ferreira ML. Risk factors for low back pain and sciatica: an umbrella review. *Spine J*. (2018) 18(9):1715–21. doi: 10.1016/j.spinee.2018.05.018

30. Stevenson JM, Weber CL, Smith JT, Dumas GA, Albert WJ. A longitudinal study of the development of low back pain in an industrial population. *Spine*. (2001) 26(12):1370–7. doi: 10.1097/00007632-200106150-00022

31. Taylor JB, Goode AP, George SZ, Cook CE. Incidence and risk factors for first-time incident low back pain: a systematic review and meta-analysis. *Spine J*. (2014) 14(10):2299–319. doi: 10.1016/j.spinee.2014.01.026

32. Bigos SJ, Battie MC, Spengler DM, Fisher LD, Fordyce WE, Hansson T, et al. A longitudinal, prospective study of industrial back injury reporting. *Clin Orthop Relat Res*. (1992) 279:21–34. doi: 10.1097/00003086-199206000-00004

33. Carroll LJ, Cassidy JD, Côté P. Depression as a risk factor for onset of an episode of troublesome neck and low back pain. *Pain*. (2004) 107(1–2):134–9. doi: 10.1016/j.pain.2003.10.009

34. Hoogendoorn WE, Bongers PM, de Vet HC, Houtman IL, Ariëns GA, van Mechelen W, et al. Psychosocial work characteristics and psychological strain in relation to low-back pain. *Scand J Work Environ Health*. (2001) 27(4):258–67. doi: 10.5271/sjweh.613

35. Battié MC, Bigos SJ, Fisher LD, Spengler DM, Hansson TH, Nachemson AL, et al. Anthropometric and clinical measures as predictors of back pain complaints in industry: a prospective study. *J Spinal Disord*. (1990) 3(3):195–204.

36. Battié MC, Bigos SJ, Fisher LD, Hansson TH, Nachemson AL, Spengler DM, et al. A prospective study of the role of cardiovascular risk factors and fitness in industrial back pain complaints. *Spine*. (1989) 14(2):141–7. doi: 10.1097/00007632-198902000-00001

37. Feldman DE, Rossignol M, Shrier I, Abenhaim L. Smoking. A risk factor for development of low back pain in adolescents. *Spine*. (1999) 24(23):2492–6. doi: 10.1097/00007632-199912010-00011

38. Cholewicki J, Silfies SP, Shah RA, Greene HS, Reeves NP, Alvi K, et al. Delayed trunk muscle reflex responses increase the risk of low back injuries. *Spine*. (2005) 30(23):2614–20. doi: 10.1097/01.brs.0000188273.27463.bc

39. Graham RB, Oikawa LY, Ross GB. Comparing the local dynamic stability of trunk movements between varsity athletes with and without non-specific low back pain. *J Biomech*. (2014) 47(6):1459–64. doi: 10.1016/j.jbiomech.2014.01.033

40. Willigenburg NW, Kingma I, Hoozemans MJ, van Dieën JH. Precision control of trunk movement in low back pain patients. *Hum Mov Sci*. (2013) 32(1):228–39. doi: 10.1016/j.humov.2012.12.007

41. Granata KP, England SA. Stability of dynamic trunk movement. *Spine*. (2006) 31(10):E271–6. doi: 10.1097/01.brs.0000216445.28943.d1

42. Rosenstein MT, Collins JJ, De Luca CJ. A practical method for calculating largest lyapunov exponents from small data sets. *Phys D Nonlinear Phenom*. (1993) 65(1):117–34. doi: 10.1016/0167-2789(93)90009-P

43. Dideriksen JL, Gizzi L, Petzke F, Falla D. Deterministic accessory spinal movement in functional tasks characterizes individuals with low back pain. *Clin Neurophysiol*. (2014) 125(8):1663–8. doi: 10.1016/j.clinph.2013.11.037

44. van den Hoorn W, Bruijn SM, Meijer OG, Hodges PW, van Dieën JH. Mechanical coupling between transverse plane pelvis and thorax rotations during gait is higher in people with low back pain. *J Biomech*. (2012) 45(2):342–7. doi: 10.1016/j.jbiomech.2011.10.024

45. Harbourne RT, Stergiou N. Movement variability and the use of nonlinear tools: principles to guide physical therapist practice. *Phys Ther*. (2009) 89(3):267–82. doi: 10.2522/ptj.20080130

46. Stergiou N, Decker LM. Human movement variability, nonlinear dynamics, and pathology: is there a connection? *Hum Mov Sci*. (2011) 30(5):869–88. doi: 10.1016/j.humov.2011.06.002

47. Dingwell JB, Cusumano JP, Cavanagh PR, Sternad D. Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. *J Biomech Eng*. (2001) 123(1):27–32. doi: 10.1115/1.1336798

48. Lockhart T, Stergiou N. New perspectives in human movement variability. *Ann Biomed Eng*. (2013) 41(8):1593–4. doi: 10.1007/s10439-013-0852-0

49. van Dieën JH, Reeves NP, Kawchuk G, van Dillen LR, Hodges PW. Analysis of motor control in patients with low back pain: a key to personalized care? *J Orthop Sports Phys Ther*. (2019) 49(6):380–8. doi: 10.2519/jospt.2019.7916

50. Arampatzis A, Schroll A, Moreno Catala M, Laube G, Schuler S, Dreinhofer K. A random-perturbation therapy in chronic non-specific low-back pain patients: a randomised controlled trial. *Eur J Appl Physiol*. (2017) 117(12):2547–60. doi: 10.1007/s00421-017-3742-6

51. Bakar F A, Staal JB, van Cingel R, Saito H, Ostelo R, van Dieën JH. Not all movements are equal: differences in the variability of trunk motor behavior between people with and without low back pain-A systematic review with descriptive synthesis. *PLoS One*. (2023) 18(9):e0286895. doi: 10.1371/journal.pone.0286895

52. Saito H, Watanabe Y, Kutsuna T, Futohashi T, Kusumoto Y, Chiba H, et al. Spinal movement variability associated with low back pain: a scoping review. *PLoS One*. (2021) 16(5):e0252141. doi: 10.1371/journal.pone.0252141

53. Laird RA, Kent P, Keating JL. How consistent are lordosis, range of movement and lumbo-pelvic rhythm in people with and without back pain? *BMC Musculoskelet Disord*. (2016) 17(1):403. doi: 10.1186/s12891-016-1250-1

54. Pan F, Firouzabadi A, Zander T, Schmidt H. Sex-dependent differences in lumbo-pelvic coordination for different lifting tasks: a study on asymptomatic adults. *J Biomech*. (2020) 102:109505. doi: 10.1016/j.jbiomech.2019.109505

55. Laird RA, Gilbert J, Kent P, Keating JL. Comparing lumbo-pelvic kinematics in people with and without back pain: a systematic review and meta-analysis. *BMC Musculoskelet Disord*. (2014) 15:229. doi: 10.1186/1471-2474-15-229

56. Arampatzis A, Frank J, Laube G, Mersmann F. Trunk muscle strength and lumbo-pelvic kinematics in adolescent athletes: effects of age and sex. *Scand J Med Sci Sports*. (2019) 29(11):1691–8. doi: 10.1111/sms.13503

57. WMA—The World Medical Association. Declaration of Helsinki (2013). Verfügbar unter: https://www.wma.net/what-we-do/medical-ethics/declaration-of-helsinki/ (zitiert Juni 13, 2023).

58. von Elm E, Altman DG, Egger M, Pocock SJ, Gøtzsche PC, Vandenbroucke JP, et al. The strengthening the reporting of observational studies in epidemiology (STROBE) statement: guidelines for reporting observational studies. *Lancet*. (2007) 370(9596):1453–7. doi: 10.1016/S0140-6736(07)61602-X

59. Klasen BW, Hallner D, Schaub C, Willburger R, Hasenbring M. Validation and reliability of the German version of the chronic pain grade questionnaire in primary care back pain patients. *Psychosoc Med*. (2004) 1:Doc07.

60. Von Korff M, Ormel J, Keefe FJ, Dworkin SF. Grading the severity of chronic pain. *Pain*. (1992) 50(2):133–49. doi: 10.1016/0304-3959(92)90154-4

61. Dworkin SF, Von Korff M, Whitney CW, Le Resche L, Dicker BG, Barlow W. Measurement of characteristic pain intensity in field research. *Pain*. (1990) 41:S290. doi: 10.1016/0304-3959(90)92696-N

62. Cohen SP, Vase L, Hooten WM. Chronic pain: an update on burden, best practices, and new advances. *Lancet*. (2021) 397(10289):2082–97. doi: 10.1016/S0140-6736(21)00393-7

63. Fraser AM, Swinney HL. Independent coordinates for strange attractors from mutual information. *Phys Rev A Gen Phys*. (1986) 33(2):1134–40. doi: 10.1103/PhysRevA.33.1134

64. Kennel MB, Brown R, Abarbanel HD. Determining embedding dimension for phase-space reconstruction using a geometrical construction. *Phys Rev A*. (1992) 45(6):3403–11. doi: 10.1103/PhysRevA.45.3403

65. Kantz H. A robust method to estimate the maximal lyapunov exponent of a time series. *Phys Lett A*. (1994) 185(1):77–87. doi: 10.1016/0375-9601(94)90991-1

66. Santuz A, Ekizos A, Eckardt N, Kibele A, Arampatzis A. Challenging human locomotion: stability and modular organisation in unsteady conditions. *Sci Rep*. (2018) 8(1):2740. doi: 10.1038/s41598-018-21018-4

67. Janshen L, Santuz A, Ekizos A, Arampatzis A. Fuzziness of muscle synergies in patients with multiple sclerosis indicates increased robustness of motor control during walking. *Sci Rep*. (2020) 10(1):7249. doi: 10.1038/s41598-020-63788-w

68. Dingwell JB, Marin LC. Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. *J Biomech*. (2006) 39(3):444–52. doi: 10.1016/j.jbiomech.2004.12.014

69. Pinheiro J, Bates DM. *Mixed-Effects Models in S and S-PLUS*. New York: Springer-Verlag (2000). (Statistics and Computing). Verfügbar unter: http://link.springer.com/10.1007/b98882 (zitiert November 16, 2023).

70. Pinheiro J, Bates D. nlme: linear and nonlinear mixed effects models. Version 3.1-167 (2025). Available at: https://CRAN.R-project.org/package=nlme (Accessed March 01, 2025).

71. Benjamini Y, Hochberg Y. Controlling the false discovery rate: a practical and powerful approach to multiple testing. *J R Stat Soc Ser B*. (1995) 57(1):289–300. doi: 10.1111/j.2517-6161.1995.tb02031.x

72. R Core Team. *R: A Language and Environment for Statistical Computing*. Vienna, Austria: R Foundation for statistical Computing (2023). Verfügbar unter: https://www.R-project.org/.

73. Shojaei I, Vazirian M, Croft E, Nussbaum MA, Bazrgari B. Age related differences in mechanical demands imposed on the lower back by manual material handling tasks. *J Biomech*. (2016) 49(6):896–903. doi: 10.1016/j.jbiomech.2015.10.037

74. Vialle R, Levassor N, Rillardon L, Templier A, Skalli W, Guigui P. Radiographic analysis of the sagittal alignment and balance of the spine in asymptomatic subjects. *J Bone Joint Surg Am*. (2005) 87(2):260–7. doi: 10.2106/JBJS.D.02043

75. Blackburn JT, Bell DR, Norcross MF, Hudson JD, Kimsey MH. Sex comparison of hamstring structural and material properties. *Clin Biomech*. (2009) 24(1):65–70. doi: 10.1016/j.clinbiomech.2008.10.001

76. Gomez T, Beach G, Cooke C, Hrudey W, Goyert P. Normative database for trunk range of motion, strength, velocity, and endurance with the isostation B-200 lumbar dynamometer. *Spine*. (1991) 16(1):15–21. doi: 10.1097/00007632-199101000-00003

77. Kumar S, Dufresne RM, Van Schoor T. Human trunk strength profile in flexion and extension. *Spine*. (1995) 20(2):160–8. doi: 10.1097/00007632-199501150-00006

78. Kumar S, Chaffin DB, Redfern M. Isometric and isokinetic back and arm lifting strengths: device and measurement. *J Biomech*. (1988) 21(1):35–44. doi: 10.1016/0021-9290(88)90189-3

79. Langrana NA, Lee CK, Alexander H, Mayott CW. Quantitative assessment of back strength using isokinetic testing. *Spine*. (1984) 9(3):287–90. doi: 10.1097/00007632-198404000-00011

80. Jorgensen MJ, Marras WS, Granata KP, Wiand JW. MRI-derived moment-arms of the female and male spine loading muscles. *Clin Biomech*. (2001) 16(3):182–93. doi: 10.1016/S0268-0033(00)00087-5

81. Marras WS, Jorgensen MJ, Granata KP, Wiand B. Female and male trunk geometry: size and prediction of the spine loading trunk muscles derived from MRI. *Clin Biomech*. (2001) 16(1):38–46. doi: 10.1016/S0268-0033(00)00046-2

82. Arampatzis A, Laube G, Schroll A, Frank J, Bohm S, Mersmann F. Perturbation-based exercise for prevention of low-back pain in adolescent athletes. *Transl Sports Med*. (2021) 4:128–37. doi: 10.1002/tsm2.191

83. Youdas JW, Garrett TR, Egan KS, Therneau TM. Lumbar lordosis and pelvic inclination in adults with chronic low back pain. *Phys Ther*. (2000) 80(3):261–75. doi: 10.1093/ptj/80.3.261

84. Böhme J, Lingslebe U, Steinke H, Werner M, Slowik V, Josten C, et al. The extent of ligament injury and its influence on pelvic stability following type II anteroposterior compression pelvic injuries–A computer study to gain insight into open book trauma. *J Orthop Res*. (2014) 32(7):873–9. doi: 10.1002/jor.22618

85. Henyš P, Ramezani M, Schewitz D, Höch A, Möbius D, Ondruschka B, et al. Sacrospinous and sacrotuberous ligaments influence in pelvis kinematics. *J Anat*. (2022) 241(4):928–37. doi: 10.1111/joa.13739

86. Snijders CJ, Hermans PFG, Niesing R, Spoor CW, Stoeckart R. The influence of slouching and lumbar support on iliolumbar ligaments, intervertebral discs and sacroiliac joints. *Clin Biomech*. (2004) 19(4):323–29. doi: 10.1016/j.clinbiomech.2004.01.006

87. Steinke H, Kurz S, Fischer B, Schuberth S, Hobusch C, Heyde CE. The sacrotuberous ligament is preloaded in situ. *J Mech Behav Biomed Mater*. (2022) 134:105368. doi: 10.1016/j.jmbbm.2022.105368

88. Aldabe D, Hammer N, Flack NAMS, Woodley SJ. A systematic review of the morphology and function of the sacrotuberous ligament. *Clin Anat*. (2019) 32(3):396–407. doi: 10.1002/ca.23328

89. Hammer N, Scholze M, Kibsgård T, Klima S, Schleifenbaum S, Seidel T, et al. Physiological *in vitro* sacroiliac joint motion: a study on three-dimensional posterior pelvic ring kinematics. *J Anat*. (2019) 234(3):346–58. doi: 10.1111/joa.12924

90. Hammer N, Steinke H, Lingslebe U, Bechmann I, Josten C, Slowik V, et al. Ligamentous influence in pelvic load distribution. *Spine J*. (2013) 13(10):1321–30. doi: 10.1016/j.spinee.2013.03.050

91. Fett D, Trompeter K, Platen P. Back pain in elite sports: a cross-sectional study on 1,114 athletes. *PLoS One*. (2017) 12(6):e0180130. doi: 10.1371/journal.pone.0180130

92. Keogh E. Sex and gender differences in pain: past, present, and future. *Pain*. (2022) 163(Suppl 1):S108–16. doi: 10.1097/j.pain.0000000000002738

93. Kjaer P, Wedderkopp N, Korsholm L, Leboeuf-Yde C. Prevalence and tracking of back pain from childhood to adolescence. *BMC Musculoskelet Disord*. (2011) 12:98. doi: 10.1186/1471-2474-12-98

94. McClure PW, Esola M, Schreier R, Siegler S. Kinematic analysis of lumbar and hip motion while rising from a forward, flexed position in patients with and without a history of low back pain. *Spine*. (1997) 22(5):552–8. doi: 10.1097/00007632-199703010-00019

95. Larivière C, Gagnon D, Loisel P. The effect of load on the coordination of the trunk for subjects with and without chronic low back pain during flexion-extension and lateral bending tasks. *Clin Biomech*. (2000) 15(6):407–16. doi: 10.1016/S0268-0033(00)00006-1

96. Wildenbeest MH, Kiers H, Tuijt M, van Dieën JH. Associations of low-back pain and pain-related cognitions with lumbar movement patterns during repetitive seated reaching. *Gait Posture*. (2022) 91:216–22. doi: 10.1016/j.gaitpost.2021.10.032

97. Alsubaie AM, Mazaheri M, Martinez-Valdes E, Falla D. Is movement variability altered in people with chronic non-specific low back pain? A systematic review. *PLoS One*. (2023) 18(6):e0287029. doi: 10.1371/journal.pone.0287029

98. Nelson JM, Walmsley RP, Stevenson JM. Relative lumbar and pelvic motion during loaded spinal flexion/extension. *Spine*. (1995) 20(2):199–204. doi: 10.1097/00007632-199501150-00013

99. Adams MA, Dolan P, Marx C, Hutton WC. An electronic inclinometer technique for measuring lumbar curvature. *Clin Biomech*. (1986) 1(3):130–4. doi: 10.1016/0268-0033(86)90002-1

100. Stokes IA, Bevins TM, Lunn RA. Back surface curvature and measurement of lumbar spinal motion. *Spine*. (1987) 12(4):355–61. doi: 10.1097/00007632-198705000-00009

101. Pan F, Arshad R, Zander T, Reitmaier S, Schroll A, Schmidt H. The effect of age and sex on the cervical range of motion—a systematic review and meta-analysis. *J Biomech*. (2018) 75:13–27. doi: 10.1016/j.jbiomech.2018.04.047

TYPE Original Research
PUBLISHED 30 April 2025
DOI 10.3389/fspor.2025.1546821

OPEN ACCESS

EDITED BY
Maria-Elissavet Nikolaidou,
National and Kapodistrian University of Athens, Greece

REVIEWED BY
Tilman Engel,
University of Potsdam, Germany
Datao Xu,
Ningbo University, China

*CORRESPONDENCE
Steven Simon
✉ steven.simon@rptu.de
Jonas Dully
✉ jonas.dully@rptu.de

RECEIVED 17 December 2024
ACCEPTED 07 April 2025
PUBLISHED 30 April 2025

CITATION
Simon S, Dully J, Ludwig O, Dindorf C, Bartaguiz E, Fröhlich M and Becker S (2025) Kinematic effects of sensorimotor foot orthoses on the gait of patients with patellofemoral pain—a randomized controlled trial.
Front. Sports Act. Living 7:1546821.
doi: 10.3389/fspor.2025.1546821

Kinematic effects of sensorimotor foot orthoses on the gait of patients with patellofemoral pain—a randomized controlled trial

Steven Simon*, Jonas Dully*, Oliver Ludwig, Carlo Dindorf, Eva Bartaguiz, Michael Fröhlich and Stephan Becker

Department of Sports Science, RPTU University of Kaiserslautern-Landau, Kaiserslautern, Rheinland-Pfalz, Germany

Introduction: Foot orthoses (FOs) are a noninvasive and cost-effective medical treatment that positively influence biomechanical factors, such as the kinematics of the lower extremities. Nevertheless, there is a research gap regarding the influence of FOs, particularly sensorimotor foot orthoses (SMFOs), on joint kinematics of the lower extremity in gait. Therefore, this randomized controlled clinical trial addressed the impact of SMFOs on the ankle, knee, and hip joint kinematics of patients with patellofemoral pain (PFP) in comparison to that of biomechanical foot orthoses (BMFOs).

Methods: A total of 20 participants (6 men; 14 women) were part of a three-month intervention with stratified random assignment to custom-made SMFO or BMFO treatment. In the pre- and post-tests, three 12-meter gait walks were assessed by inertial measurement units (IMUs) with the patients wearing no FOs, SMFOs, and BMFOs. For each joint in all three dimensions, three-way repeated-measures statistical parametric mapping (SPM) was performed using analysis of variance (ANOVA)-like statistics. *post-hoc*, the significant results were checked using *post-hoc* *t*-test-like SPMs.

Results: Results show that SMFOs and BMFOs both significantly changed ankle and knee kinematic parameters in patients with PFP in long-term. No significant immediate effects of FOs were detected; however, there were significant interaction effects between the time of measurement and the groups. In the pre-post comparison, the SMFO-treated group showed less dorsiflexion in the initial contact and terminal stance, less knee flexion in the mid stance, terminal stance, and pre-swing, as well as a more neutral knee movement in the frontal plane. The BMFO-treated group showed slightly more knee abduction in the terminal stance, greater knee flexion at initial contact, and less hip adduction at initial contact.

Conclusions: Overall, the results of this trial support the assumption that temporal adaptation processes play a vital role in the application of custom-made orthopedic FOs and highlight the long-term effects on the kinematics of the lower extremities.

KEYWORDS
sensorimotor system, insoles, chondropathia patellae, gait analysis, inertial measurement units

1 Introduction

Orthopedic foot orthoses (FOs) are medical aids to help patients who have pain in the lower extremities (1), especially patients with distal ankle instability (2), high pronation in gait and running (3, 4), and knee pain (5, 6). Depending on the cause and symptoms, FOs are customized to achieve more favorable kinematic (e.g., joint angle) and kinetic (e.g., force peaks) conditions for lower extremity joint pain relief (7, 8). Further, FOs have been proven to reduce pain in different pathologies of the lower extremities (9, 10). Patellofemoral pain (PFP) syndrome is one of the most common causes of anterior knee pain in adolescents and adults (11, 12). Static and dynamic components, including axial and rotational errors of the lower extremity (13) and foot malalignment (14) might be the causative factors for PFP. Foot malalignments with excessive or insufficient pronation of the foot influence knee abduction moment in the frontal plane and ground reaction forces (15). Therefore, from a biomechanical perspective, the aim of the practice is to redirect the forces acting on the femoropatellar joint. Saxena et al. (16) concluded that FOs are an effective treatment option for relieving the clinical symptoms of PFP, particularly in young people. Lewinson et al. (17) investigated the potential of modifying the angular impulse magnitude of knee abduction through lateral and medial wedged FOs to alleviate pain in runners with patellofemoral pain (PFP) and found a clinically significant pain reduction. Gross and Foxworth (18) stated that FOs have a positive impact on patients with PFP with excessive foot pronation, lower extremity alignment, and an increased Q-angle.

In orthopedic care, a distinction is made between the two main approaches of custom-made FOs: biomechanical (BMFOs) and sensorimotor FOs (SMFOs) (10, 19, 20). BMFOs are characterized by supporting, bedding, and shell elements that are primarily intended to provide support, correction, and relief. In contrast, SMFOs primarily influence the activity of defined muscles via the corresponding elements at specific time intervals in the step cycle in a targeted manner (19, 21). Studies have measured the influence of SMFOs on muscle activity (22, 23), joint kinematics of the foot (24) and tibia, femur rotation (25) during walking, and postural parameters (26) in different samples with and without pathologies. Chondropathia patellae, associated with impaired patellofemoral kinematics (27), was listed by Greitemann et al. (21) as an indication for SMFO. The major targets of SMFOs in patients with PFP are improved motor control and muscle activation during movement (6), improved knee guidance through motion control of the hindfoot, and consequently, reduced retropatellar pressure (19). Kerkhoff et al. (6, 28) examined prefabricated BMFOs and SMFOs and their effects on the muscle activity of the lower extremities in participants with nonspecific knee pain. Their results showed that prefabricated BMFOs and custom-made SMFOs led to different activation patterns compared to shoes without FOs during a single-leg landing test. In contrast, SMFOs increased the influence of the semitendinosus and peroneus longus muscles on the gait. Ludwig et al. (22) measured the activation effect of sensorimotor foot orthoses (SMFO) on the peroneus longus muscle, which is primarily responsible for dynamic balance control and ankle joint stability. However, the biomechanical efficiency of the FOs in patients with knee pain remains unclear (22, 28). There is a lack of clinical studies that address the different mechanisms of action on lower extremity kinematics and their clinical effect on pathologies (19).

Therefore, the authors addressed the following research question: Does wearing custom-made SMFOs alter the kinematics of the ankle, knee, and hip in patients' gait in the short- (immediate) and long-term (three months) differently from custom-made BMFOs? It was hypothesized that both FOs immediately change the kinematics of the ankle, knee, and hip joints of the diseased knee side, and based on their different mechanisms of action, SMFOs differ in their long-term effects on the patient's gait.

2 Methods

2.1 Study design

This study represents a double-blinded, randomized-controlled clinical trial (RCT) with pre- and post-testing. The intervention period was three months. The sample was randomly assigned to an orthopedic device (SMFO or BMFO) over the intervention period after diagnostic and orthopedic anamnesis by the physician, considering the inclusion and exclusion criteria (see Chapter 2.2. and Figure 1). Placebo foot orthoses were not included in the three-month intervention due of ethical restrictions.

Scientific evaluation was carried out using the standard procedure of the physician and orthopedic shoe technician, who treat patients with corresponding orthopedic indications. The study was conducted in accordance with the local legislation and institutional requirements in accordance with the Declaration of Helsinki and approved by the Ethics Committee of the Kaiserslautern-Landau Sozialwissenschaften (Nr. 70, 16 February 2024). The study was registered in the German Clinical Register and the World Health Organization Clinical Trials Registry Platform (DRKS00035082). The participants provided their written informed consent to participate in this study.

2.2 Participants

A total of 26 participants were included in the baseline and follow-up gait assessments. Perceived anterior knee pain was included in another study using the same sample (29). In contrast to (29), after data processing, 20 participants were included in statistical analysis. An explanation of the dropouts in the data analysis is provided in Section 2.4.

As part of this project, another study has already been published, focusing on clinical pain development (29). A sample size of 24 participants was determined for an analysis of variance (ANOVA) to assess the interaction effect (effect size $f = 0.25$, 2 groups, 2 ToMs, α error probability = 0.05, correlation among

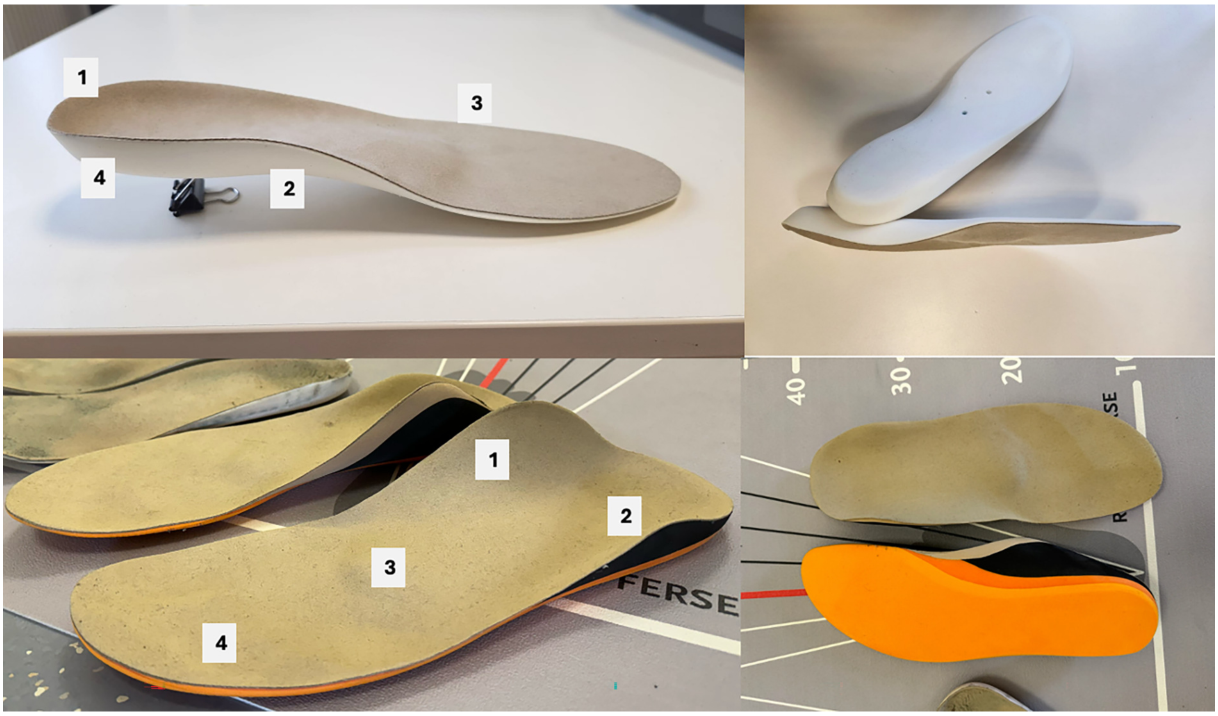

FIGURE 1
Foot orthoses (FOs) of the control group (above; BMFO, biomechanical foot orthoses) and intervention group (below; SMFO, sensorimotor foot orthoses) and. BMFO: 1 = light shell and heel pad, 2 = supination wedge, 3 = metatarsal pad, 4 = injection molded foam, 25 shore; SMFO: 1 = medial spot, 2 = lateral spot, 3 = retrocapital element, 4 = toe bar, 5 = sandwich material [ethylenvinylacetat (EVA), 35–25–35 shore].

TABLE 1 Anthropometric data of sample ($n = 20$).

Groups	Descriptive statistics	Age (y)	BMI (kg/m^2)	NI left	NI right	AI left	AI right
IG	Mean	24.40	22.34	0.16	0.16	0.20	0.20
	SD	7.65	3.78	0.02	0.03	0.07	0.06
	Max	38	31.17	0.18	0.20	0.31	0.27
	Min	15	17.26	0.12	0.13	0.09	0.09
CG	Mean	26.36	24.02	0.20	0.21	0.24	0.24
	SD	12.13	6.59	0.06	0.06	0.06	0.02
	Max	54	42.52	0.33	0.33	0.30	0.27
	Min	16	16.14	0.13	0.13	0.10	0.21

NI, navicular index (55); AI, arch index (56).

repeated measures: 0.8) using G*Power 3.1 (30). The anthropometric data are shown in Table 1.

The inclusion and exclusion criteria were defined and assessed by the same physician for all participants.

Inclusion criteria:

- Age between 15 and 60 years
- Discomfort in the knee joint area during at least two weight-bearing activities (walking stairs, squatting, standing up) for at least three weeks: pain during these activities on most days in the last month that is ≥ 30 mm on a 100-mm Visual Analog Scale (VAS)
- Foot malalignment
- Indication (at least 1 out of 5):
  - Femoropatellar pain syndrome
  - Chondropathia patellae up to grade 3 with pathological alignment and femoral antetorsion
  - Runner's knee, jumper's knee
  - Osteochondral defects, inflammation and impingement of the Hoffa fat body,
  - Tendinopathies of the patellar or quadriceps tendon, patellofemoral osteoarthritis, plica syndrome
  - Altered Q-angle of the lower extremity ("malalignment")/ recognizable rotational abnormality of ankle joints, tibia and femur during gait

Exclusion criteria:

- Medical history of knee joint arthroplasty or osteotomy
- Previous (surgical) treatment (<12 months) of ankle, knee, or hip joints
- Radiographic evidence of fixed bone deformity or joint erosion
- Moderate or severe concomitant tibiofemoral osteoarthritis [Kellgren and Lawrence grade 3 on anteroposterior radiograph (31)]
- Underlying neurological pathology
- Known underlying rheumatic disease with drug treatment
- Previous treatment with orthopedic foot orthoses
- Acute muscle/ligament injury (<4 weeks)

All participants were physically active. As a termination criterion during the intervention, an increase in subjectively perceived pain by two points or more during the intervention period was

10.3389/fspor.2025.1546821

defined. Statistical evaluation of pain perception and development, comfort and effectiveness rating was part of another study (29). In addition, additional physiotherapeutic treatment was documented (assessed in 7 out of 20 participants), daily steps and wearing time was documented by questionnaires weekly (12 times) and the participants were asked about their dominantly pain-affected leg in case of knee pain on both sides.

2.3 Procedure

After anamnesis and diagnosis by a physician, the participants were instructed to attend a foot measurement appointment with an orthopedic technician. All parameters for medically indicated FO fitting were determined according to German medical standards and commissioned based on 2D foot scans and 3D foot molding. The orthopedic shoe technician was responsible for all patients. Two pairs of FOs (BMFO and SMFO) were produced for each patient and subjected to fitting and dispensing appointments. Both types of FO treatments (Figure 1) were manufactured and individually adapted to the patients' pain, foot, and knee conditions. In SMFO treatment, a medial element is positioned to apply targeted pressure along the tendon path of the tibialis posterior muscle. The lateral element is placed dorsally on the calcaneus, exerting pressure on the tendon paths of the peroneus longus and brevis muscles. The retrocapital element is positioned just behind the metatarsophalangeal joints of toes two to five. Additionally, a toe bar under the middle and distal phalanges of toes two to five provides a comfortable resting area for the distal phalanges. The design of the BMFO prioritized the height of the supination and pronation wedges, as well as the metatarsal pad, tailored to the individual's foot type (longitudinal and transversal arch posture) and clinical indication. The key advantage of these FOs lies in their soft padding, which distributes pressure preventing pressure peaks and ensuring continuous cushioning.

The orthopedic shoe technician responsible for the individual, custom-made FOs was not informed which FO was assigned to the participant during the intervention period.

First, all body dimensions (foot, leg length and width, pelvis width, and sternal height) were measured. Anamnesis, including body height, weight, activity level, and medical history, was performed. The primary research objective was to analyze various established parameters of habitual gait in the lower extremities using sensors. Therefore, Xsens inertial measurement units (IMUs) (Movella, Enschede, the Netherlands) were used. The results of Al-Amri et al. (32) suggest that the MVN BIOMECH system can be used by clinicians to quantify lower limb joint angles in clinically relevant movements. Nijmeijer et al. (2023) (33) demonstrated that the Xsens IMU system delivers highly comparable angular curves for sagittal lower-body joint kinematics during sports-specific movements, such as jump landing and change-of-direction tasks, to Vicon (Vicon Motion Systems, Ltd.).

A lower-body model [60 Hz, Xsens MVN Analyze Pro 2024.2; Xsens Technologies B.V. (Enschede, The Netherlands)] represented by eight IMUs was used to measure the kinematics of the lower extremity and pelvis (see Figure 2).

- 2x foot (each side)
- 2x lower leg (each side)
- 2x upper leg (each side)
- 1 pelvis (see Figure 3)
- 1 sternum (not shown in Figure 2)

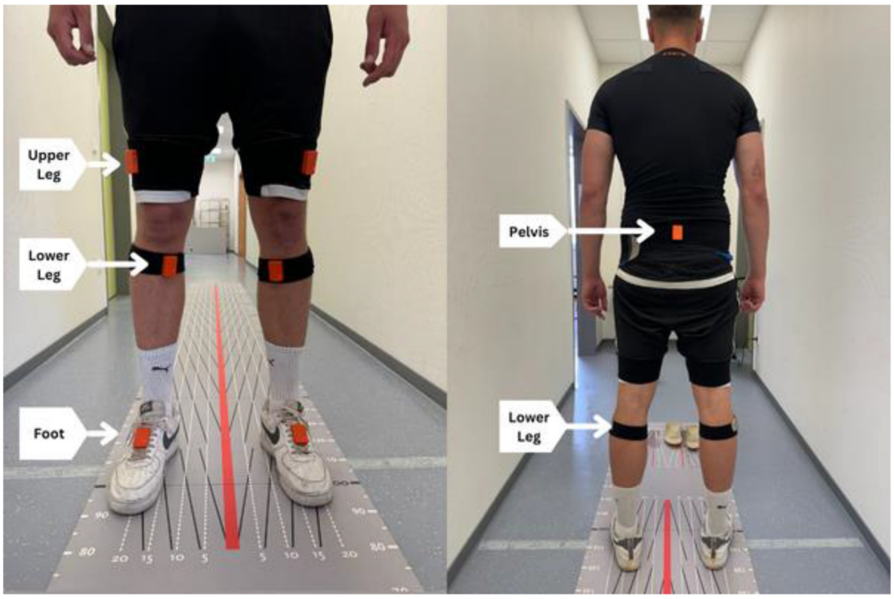

FIGURE 2
IMU-supported gait analysis with a 12-meter walking distance.

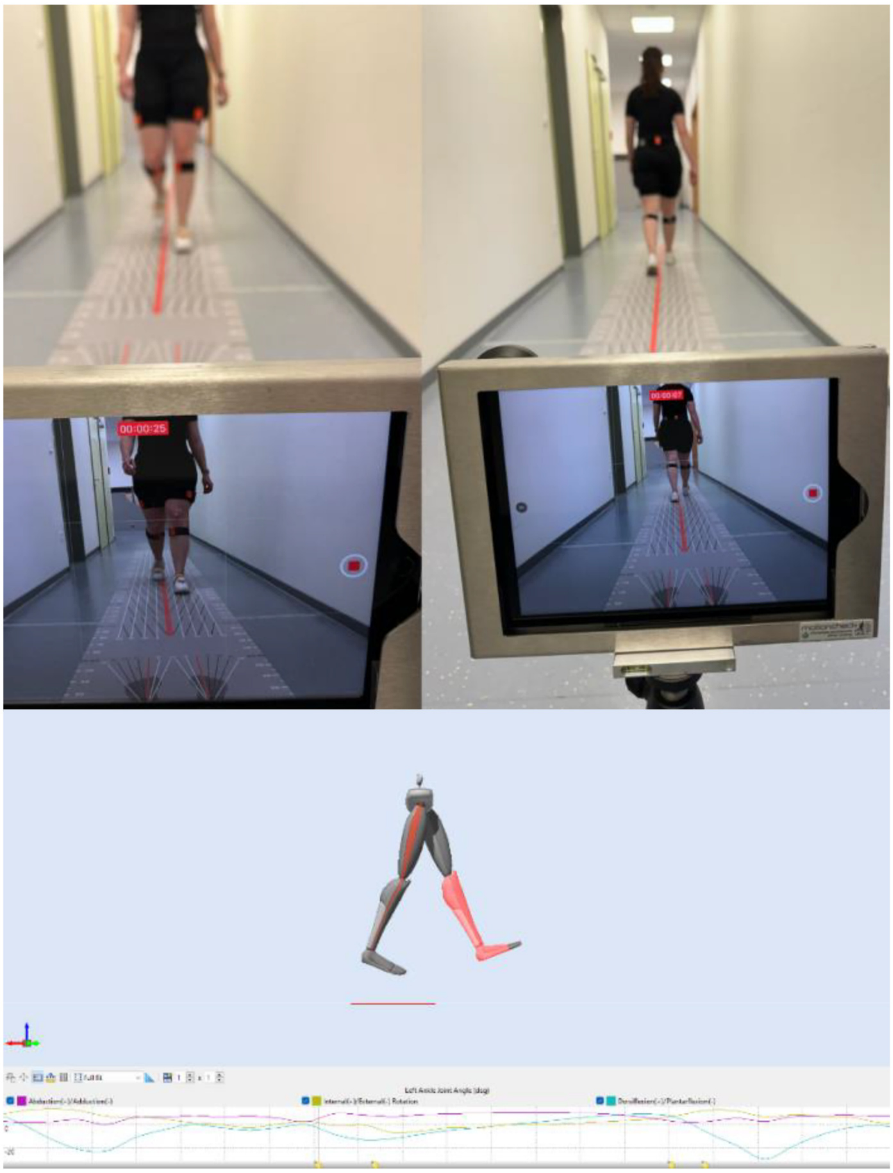

FIGURE 3
IMU placement on the patient's body (Xsens Movella, Enschede, Netherlands).

Each IMU comprised a three-axis accelerometer (±16 g), three-axis gyroscope (±2,000 degrees/s), and three-axis magnetometer (±1.9 Gauss) (31). These axes represent a robust and precise reference system for reconstructing three-dimensional (3D) motion. The same test leader attached the IMUs at both measurement times and fixed it to the body with a hook-and-loop fastener. They were attached to a shoe with adhesive tape. A measuring tape was used to measure the uniform distance from the knee and ankle for both measurements. The hook-and-loop fasteners protected the IMU from clothing-induced movement.

As part of the gait test, the test participant had to walk along a 12-meter gait path three times at a self-selected gait speed (Figure 3). The participant was allowed to walk the course four times in advance to become accustomed to the distance and setting. The gait speed should correspond to a patient's normal daily comfort speed. The gait distance was chosen according to the results of a systematic review by Hortobágyi et al. (34); therefore, it can be seen as a representative setting for enabling normal gait. Each participant was allowed to wear their own footwear, but the same footwear was worn during the posttest. The gait test was performed using three settings.

- 3 × 12 m with SMFOs (sensorimotor foot orthoses)
- 3 × 12 m with BMFOs (biomechanical foot orthoses)
- 3 × 12 m with NFOs (no foot orthoses)

The order of the FO in the individual shoes was randomly assigned to each participant for the pretest and structured identically in the posttest. Five gait cycles, each side cutting off the first two steps of each gait measurement, were included in the data analysis to exclude variability in the initiation and termination steps at the beginning and end of the gait cycle. After each twelve-meter distance, the patient had to stand still for three seconds to stop the measurement in the XSens Software, turn around in the starting position, and perform a new measurement. After the FO set was recorded, the test supervisor changed the FOs according to a random procedure. The participants had a five-minute break between settings.

2.4 Data processing and statistical analysis

After medical examination, pedographic foot measurement and motion capture were completed, and the stance phases of the gait were labeled by the test supervisor (heel strike: low point of the heel; toe off: lifting of the toe) in the Xsens MVN Analyze Pro 2024.2 software and exported in CLS files. Next, the data were merged using MATLAB (R2024b, MathWorks, Massachusetts, USA). Data acquisition resulted in three trials for each of five steps. Some participants did not have three valid trials; therefore, only two trials were used for further data processing, which were available for all subjects. This resulted in ten steps for the (dominantly) affected leg in each setting (pre- and post-training). To compare the kinematic variables (angular curves), each stance phase was normalized to 100%. The IMU-based data from the dominant PFP-affected leg were time-normalized.

For each variable, i.e., the ankle, knee, and hip angles in the sagittal and frontal planes, three-way repeated measures statistical parametric mapping (SPM) was calculated using ANOVA-like statistics with an alpha of 0.05. The two repeated measurement factors were the time of measurement (pre- and post-intervention) and the foot orthotic worn (SMFO, BMFO, NFO), whereas the non-repeated factor was the treatment group (intervention, control). Because the SPM requires a balanced design for all factors, ten participants were chosen (minimum number of participants in each group). The subjects with the most similar anthropometric characteristics between the intervention groups (SMFO and BMFO) were chosen using the Euclidean distance on z-standardized anthropometrics.

Post hoc for each FO, a two-way repeated-measures SPM was calculated with time and FO intervention factors. Significant results were checked using *post-hoc* *t*-test-like SPMs with (a) unpaired statistics for the intervention groups (SMFO and BMFO) and (b) paired statistics for the time factor. All steps were performed in MathLab (MathWorks, Natick, MA, USA) using the spm1d package (35). The results were visualized as the time-dependent group mean and standard deviation, as well as the F- or t-scores, including their critical threshold and the area under the significant results.

In addition, the anthropometric data of the groups (intervention and control) were checked with an independent Welch's *t*-test using JASP (Version 0.19.0, JASP Team, Netherlands), as few violations of sphericity and variance homogeneity were found [body mass index (BMI), navicular index].

3 Results

3.1 Consort

This study adheres to CONSORT guidelines. 34 participants were assessed for eligibility, owing to expected dropouts. 27 participants were recruited by the physician. Finally, one dropout occurred in the control group during the measurement period and six participants had to be excluded due to missing data and statistical analysis (see Figure 4 and Section 2.4).

3.2 Short-term effects

Statistical analysis revealed no significant kinematic effects in both planes and all joints induced by custom-made SMFOs and BMFOs in the pretest. The descriptive data showed slightly different average angular curves in the stance phases in the frontal plane movement of the ankle. Both types of FO increased supination non-significantly in initial contact, loading response, and pre-swing and non-significantly reduced eversion in mid stance and terminal stance, whereas SMFOs showed a slightly non-significant stronger impact than BMFOs (Figure 5).

The angular curves of the knee and hip joints, as well as those in the other planes, are available in the Supplementary Material of this study. Bonferroni-corrected independent Welch's *t*-tests showed no significant differences in anthropometric data between the intervention and control groups (age: $p_b > .99$; BMI: $p_b = .82$; NI_left: $p_b = .25$; NI_right: $p_b = .19$; AI_left: $p_b > .99$; AI_right: $p_b = .72$).

3.3 Long-term effects

3.3.1 Ankle

In the frontal plane, based on the three-way SPM ANOVA-like statistics, no significant effects between groups, time of measurement, or their interaction were detected (see Figure 6).

Regarding sagittal plane movement, three-way SPM ANOVA-like statistics showed a significant difference between foot orthoses and time of measurement, and a significant interaction effect between groups and time of measurement (Figure 7). The *post-hoc* test results are presented in Tables 2, 3. The SMFO-treated group showed reduced dorsiflexion and faster plantar flexion in the pre-swing phase compared to the BMFO-treated group. Furthermore, a significant interaction effect between the groups and the time of measurement was detected. The SMFO-treated group showed lower ankle joint dorsiflexion in the pre-swing phase. The posttest showed a significant decrease in dorsiflexion of the ankle joint in the terminal stance across all groups. The initial contact with the loading response and terminal stance showed a significant interaction effect between the groups and ToM. The SMFO-treated group exhibited the lowest dorsiflexion of the ankle joint in the posttest. In the pre-

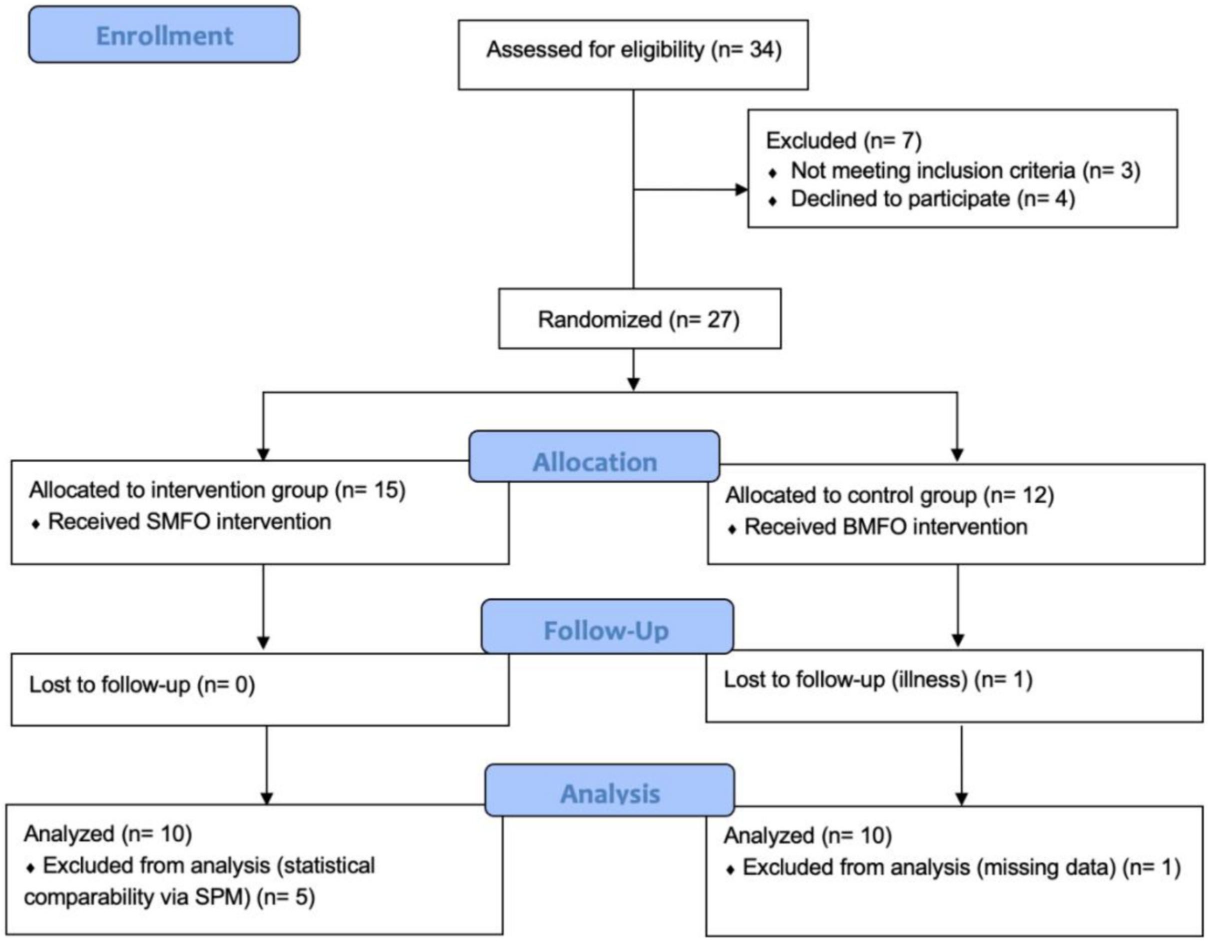

FIGURE 4
CONSORT flow diagram of this blinded randomized controlled trial (RCT). SMFO, sensorimotor foot orthoses; BMFO, biomechanical foot orthoses; SPM, statistical parametric mapping.

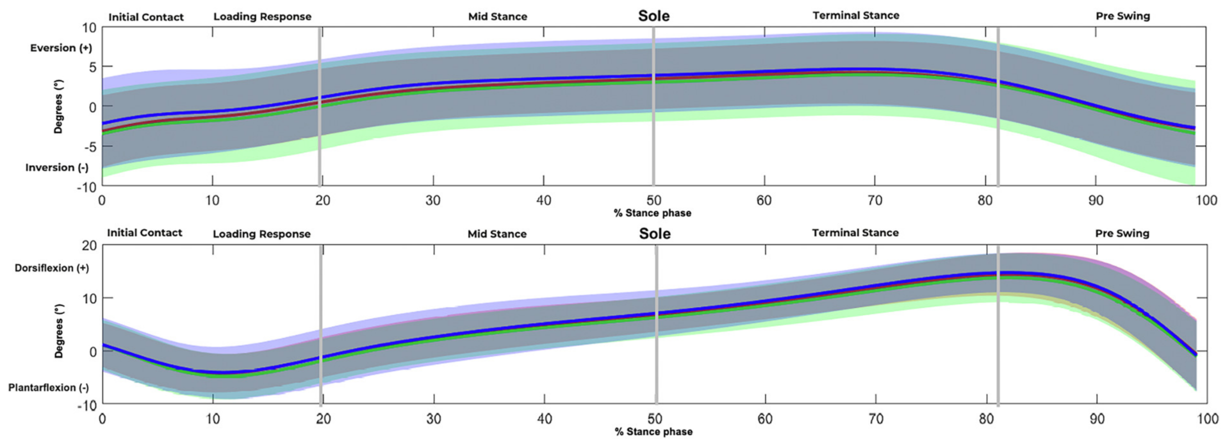

FIGURE 5
Angular curves of the ankle in the (1) frontal plane [eversion (positive), inversion (negative)] and (2) sagittal plane (dorsiflexion, plantar flexion) in the stance phases. *Y*-axis = degrees, *x*-axis = % of the stance phase. Green = SMFO, red = BMFO, blue = NFO.

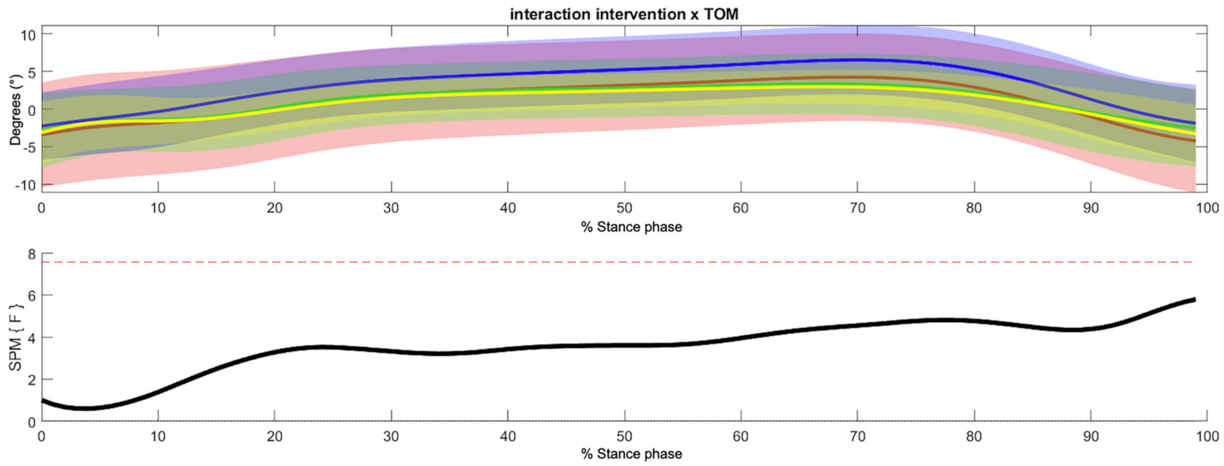

FIGURE 6
Non-significant interaction effect of intervention group and time of measurement within the 3-way repeated measures statistical parametric mapping (SPM) with averaged angular curves of the ankle joint in frontal plane. Statistical significance is present if the SPM value exceeds the threshold (red dotted line). Colors: Interaction Intervention × ToM: yellow = SMFO_pre, green = SMFO_post, red = BMFO_pre, blue = BMFO_post. Black: *F*-value.

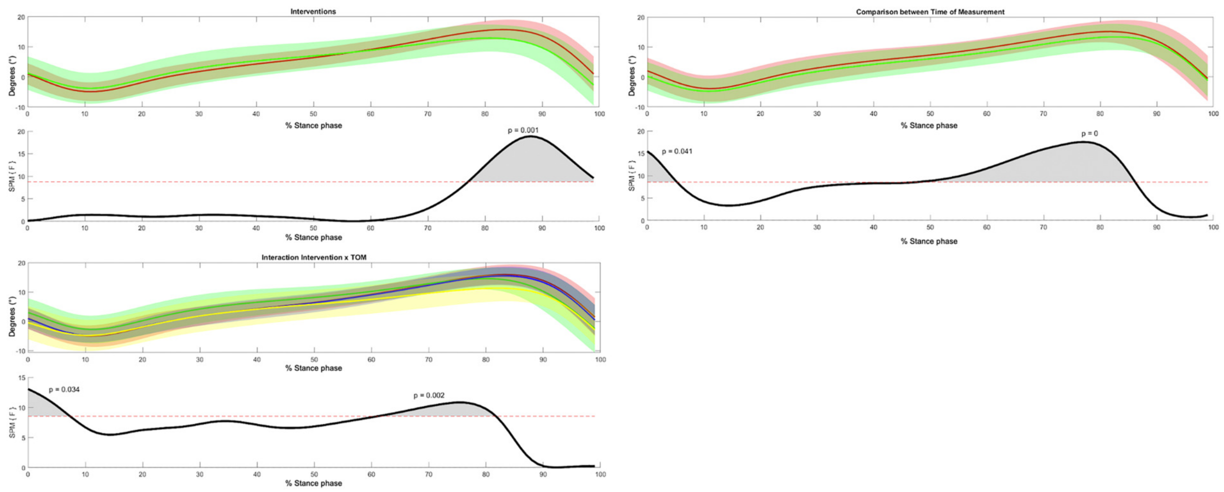

FIGURE 7
3-way repeated measures statistical parametric mapping (SPM) with averaged angular curves of the ankle joint in the sagittal plane. Statistical significance is present if the SPM value exceeds the threshold (red dotted line) represented by gray shaded areas. Colors: Interventions, green = SMFO, red = BMFO; ToM: red = pre-test, green = post-test; Interaction Intervention × ToM: yellow = SMFO_pre, green = SMFO_post, red = BMFO_pre, blue = BMFO_post. Black: *F*-value.

test, dorsiflexion in the SMFO group was still most pronounced at initial contact but decreased at pre-swing.

3.3.2 Knee

In the frontal plane, statistical analysis revealed a significant interaction effect between the intervention and time of measurement in the initial contact, loading response, and beginning of mid-stance. Particularly at the end of the loading response and beginning of mid-stance, the BMFO-treated group showed an increase in knee adduction. In contrast, the SMFO-treated group showed a normalized waveform in the range of 1–3 degrees of knee abduction (Figure 8).

Regarding the sagittal plane movement, significant intervention, time of measurement, and interaction effects were detected (see Figure 9). While knee flexion was reduced in the SMFO-treated group, the BMFO-treated group showed slightly stronger knee flexion. The SMFO-treated group exhibited an angular curve near the *x*-axis (0 degrees). The SMFO-treated group exhibited less knee flexion during the pre-swing period.

TABLE 2 Significant *post-hoc* group differences (intervention, control) in post-testing.

Comparison between intervention and control group in post-test wearing SMFO, BMFO and NFO						
Joints	Setting[a]	Initial Contact	Loading Response	Mid Stance	Terminal Stance	Pre-Swing
Ankle (frontal, sagittal, transversal plane)	NFO				Dorsiflexion: SMFO < BMFO	Plantarflexion: SMFO > BMFO
Knee (frontal, sagittal, transversal plane)	NFO		Abduction: SMFO < BMFO	Abduction: SMFO < BMFO	Flexion: SMFO < BMFO (end of TS)	Flexion: SMFO < BMFO
Hip (frontal, sagittal, transversal plane)	SMFO	Adduction: SMFO > BMFO				
	BMFO	Adduction: SMFO > BMFO	Adduction: SMFO > BMFO			
	NFO	Adduction: SMFO > BMFO			Adduction: SMFO < BMFO	

[a]The column "Setting" represents the condition in which the participants wore SMFO (sensorimotor foot orthoses), BMFO (biomechanical foot orthoses) or NFO (no foot orthoses) in the gait post-test.

3.3.3 Hip

In the frontal plane movement, a significant interaction effect between the intervention and ToM was detected at the initial contact, loading response, and beginning of mid-stance. The BMFO intervention group showed less hip abduction during initial contact in the post-test, whereas in the pre-test, there was slight hip adduction. *post hoc* tests, such as the SPM (Tables 2, 3), showed that the SMFO-treated group had more hip flexion in the terminal stance and early pre-swing.

4 Discussion

To the best of our knowledge, no study to date has examined the kinematic effects of both BMFOs and SMFOs in patients with PFP using IMUs based on an RCT study design. Results show that both SMFOs and BMFOs significantly changed ankle and knee kinematic parameters in PFP patients' gait in long-term (three months) but not in short-term.

4.1 Short-term effects

Although slight differences were observed in the angular curves of the assessed lower extremity joints (ankle, knee, and hip), no significant kinematic differences between the settings (SMFO, BMFO, and NFO) were detected immediately (pretest). The short-term results are in line with Laštovička et al. (36) who found no significant effects of SMFOs on the gait of asymptomatic healthy adults on the lower extremity kinematics besides hip adduction using a three-dimensional motion analysis system. Generally, FOs are intended to bring about changes in the kinematics of the lower extremities by altering the position and arches of the foot; however, there is still a lack of evidence regarding their effectiveness on kinematic changes (20, 22, 28, 37). In contrast, Klein et al. (24) observed an immediate reduction in rearfoot eversion when inserting SMFOs. Moisan et al. (38) concluded that FOs affect the biomechanics of the distal segments of the knee during most functional tasks such as step-up and step-down tests, jump landing, or stair ascent/descent. The necessity of a familiarization phase might be a possible explanation, as lower-extremity joints and their movements adapt after a certain period of wear (39, 40). The long-term results of this trial supported this hypothesis (Chapter 3). In contrast, Leung and Merseley have shown that FOs have an immediate impact on kinematics and lead to temporal improvements in gait in adults with hemiplegia (41). Nevertheless, the sample size of this study was not comparable to that of the PFP participants in this RCT. Highlighting the descriptive data of the frontal plane of the ankle, less eversion in the whole stance phase was detected when wearing BMFOs and SMFOs compared to NFOs, potentially induced by the medial support of the BMFO and the medial spot of the SMFO; however, this was not significant.

4.2 Long-term effects

In contrast to the short-term analysis, significant effects between groups, time of measurement, and their interactions were observed. The three-way ANOVA-like SPM revealed a significant difference between groups (intervention and control) in the sagittal angles of the ankle and knee. The BMFO-treated group showed greater dorsiflexion of the ankle in the terminal stance and pre-swing phases, while the SMFO-treated group showed reduced dorsiflexion and earlier plantar flexion of the ankle in the late stance phase. A possible explanation for that might be the proprioceptive and tactile stimulation of the SMFOs' elements on the forefoot, influencing the movement of the ankle in the phase where the most pressure is on the retro capital element and toe bar (19). Regarding the sagittal plane of the knee, significantly lower knee flexion was observed in the SMFO group than in the BMFO-treated group during the transition from the terminal stance to the beginning of the pre-swing. By reducing excessive medial or lateral knee loading, the

TABLE 3 Significant *post-hoc* time effects in intervention (SMFO) and control (BMFO) group.

Time effects (Pre vs. Post)		SMFO-treated group					BMFO-treated group				
Joints	Plane	Initial Contact	Loading Response	Mid Stance	Terminal Stance	Pre-Swing	Initial Contact	Loading Response	Mid Stance	Terminal Stance	Pre-Swing
Ankle	Sagittal	Dorsiflexion ↓			Dorsiflexion ↓						
Knee	Frontal	Abduction ↑	Abduction ↑	Adduction ↓						Abduction ↑	
	Sagittal			Flexion ↓	Flexion ↓	Flexion ↓	Flexion ↑				
Hip	Frontal						Adduction ↓				

SMFO-treated group = Three-month treatment with sensorimotor foot orthoses; BMFO-treated group = Three-month treatment with biomechanical foot orthoses.

SMFOs may shift the body's mechanics toward less reliance on knee flexion to absorb and manage forces during terminal stance. Less knee flexion during terminal stance to pre-swing could be a protective adaptation; nevertheless, both groups showed different effects because a significant interaction effect was found between the groups and the time of measurement. Generally, reduced peak knee flexion during gait is associated with PFP (11). However, studies are required to investigate this effect.

Further, time effects were detected in the sagittal angles of the ankle, hip, and knee. Both groups showed slightly less ankle dorsiflexion in the initial contact and terminal stance, less hip extension in the terminal stance, and less knee flexion in the mid-stance after the treatment period. Generally, adaptation processes may be a causative factor for these results. Hsu et al. (39) found that, after long-term use of laterally-wedged insoles, pain and physical function improved, along with a decrease in the peak knee adduction moment. The authors summarized that laterally wedged insoles provide both immediate support for walking and long-term gait adaptations that reduce stress on the knee joint in individuals with bilateral medial knee osteoarthritis. Exteroceptors and proprioceptors, as part of the sensorimotor system as well as the relevant foot and ankle muscles, may adapt movements of the lower extremity' joints according to FO treatment (19). Considering the possible biomechanical chain of the lower extremities, there appears to be a contradiction between the results of ankle, knee, and hip movements. The authors expected that reduced dorsiflexion and increased knee extension were associated with greater hip extension. In contrast, less hip extension was observed in our study, which could not be fully explained. It must be considered that measurements at multiple time points might have an influence on the results. Al-Amri et al. (32) found that for walking, the between- and within-rater reliability of discrete kinematic parameters provided by the MVN BIOMECH system ranged from fair to excellent. The ICC values for the system in the study of Al-Amri et al. were between 0.65 and 0.99, with a small standard error of measurement (SEM) of less than 3.0° for ankle and knee joints and all planes (32). These results indicate a good measurement accuracy; however, despite this high technical precision, biomechanical variability remains a crucial factor in interpreting the effects of FOs.

Significant interaction effects were observed in the sagittal and frontal angles of the knee and the frontal angle of the hip. Regarding the frontal plane of the hip joint, a slight hip abduction in the hip joint could be constantly observed with SMFO, which increased, whereas there was a slight hip adduction (in the initial contact and beginning of the loading response) with BMFO, which increased slightly. There is limited evidence highlighting the association between the peak hip adduction angle and the development of PFP in runners (11). Hoglund et al. (42) summarized that patients with PFP have altered movements during the step down test compared with asymptomatic males. Specifically, they found that PFP participants had increased hip and pelvis range of motion in the frontal and transverse planes during a step-down test in the frontal and transverse planes but reduced or nearly equal range

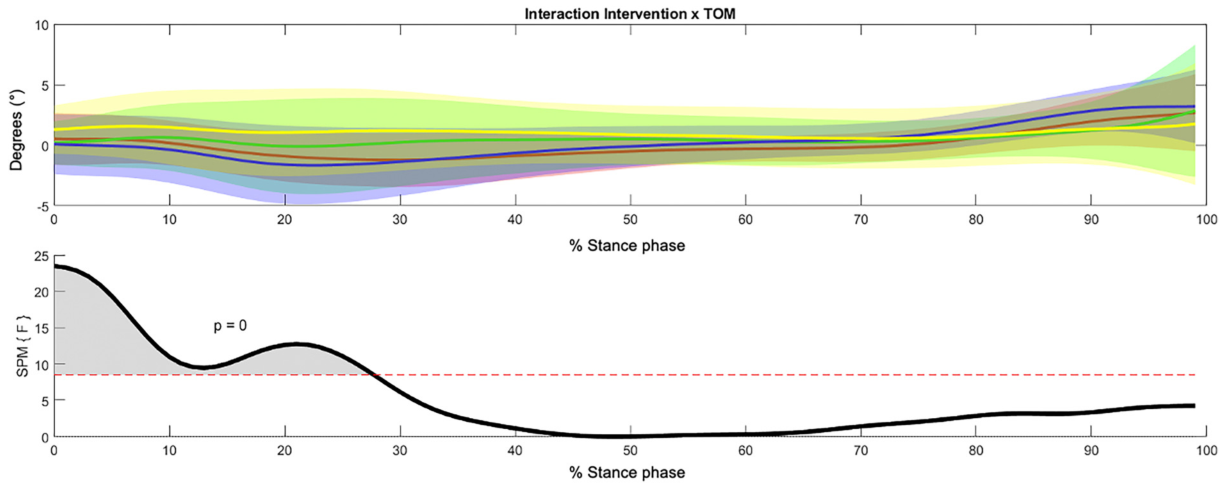

FIGURE 8
Significant interaction effect (initial contact, loading response, beginning of mid stance) of intervention group and time of measurement within the 3-way repeated measures statistical parametric mapping (SPM) with averaged angular curves of the knee joint in frontal plane. Statistical significance is present if the SPM value exceeds the threshold (red dotted line) represented by gray shaded areas. Colors: Interaction Intervention × ToM: yellow = SMFO_pre, green = SMFO_post, red = BMFO_pre, blue = BMFO_post. Black: *F*-value.

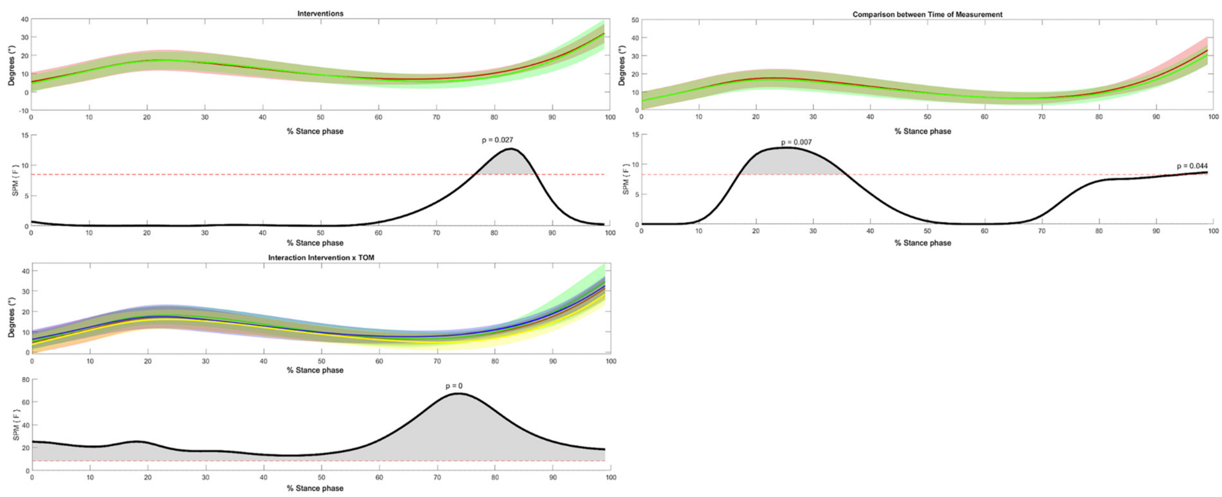

FIGURE 9
3-way repeated measures statistical parametric mapping (SPM) with average angular curves of the knee joint in the sagittal plane. Statistical significance is present if the SPM value exceeds the threshold (red dotted line) represented by gray shaded areas. Colors: Time of measurement: red = pre-test, green = post-test; interventions: green = SMFO, red = BMFO; Interaction Intervention × ToM: yellow = SMFO_pre, green = SMFO_post, red = BMFO_pre, blue = BMFO_post. Black: *F*-value.

of motion for these variables during single-leg squats. While it seems plausible that correcting foot alignment with FO makes a therapeutic contribution to the treatment of foot and Achilles tendon complaints, the extent to which FOs can contribute to knee pain due to their biomechanical or sensorimotor effects (10).

Following the *post hoc* test results of the statistical analysis, the SMFO intervention group showed reduced knee flexion at the terminal stance and pre-swing. Nevertheless, in our study the SMFO-treated group showed more neutralized frontal plane knee movement in mid-stance in pre-post comparison and less knee abduction than the BMFO-treated group in the post-test while wearing no foot orthoses. This could be explained by the different mechanisms of action of both FO concepts, whereby the biomechanical approach could guide the knee laterally via the medial wedge, whereas SMFO takes the functional chain into account by probably activating the foot supinator muscles. However, further studies are required to confirm this hypothesis.

Kinematic changes should always be interpreted in line with the therapeutic targets. This RCT included patients with PFP and multiple foot malalignments. Therefore, multifaceted kinematic effects are desirable. This RCT was additionally controlled by measuring perceived pain. The results are presented in detail in another study (29). Both interventions resulted in a significant reduction in pain between baseline and follow-up measurements, as well as over the 12-week period assessed by VAS. SMFO was perceived as more effective ($Mean_{Diff} = 1.42$) and slightly more comfortable ($Mean_{Diff} = 0.36$) than BMFO on an 11-item VAS; however, statistical analysis did not show significant differences for either parameter. Both types of FOs demonstrated a high level of comfort (BMFO: 7.91 ± 1.87; SMFO: 8.27 ± 1.10). The induced kinematic effects might help the scientific community and orthopedic aid sectors better understand the impact of different FO approaches to further improve the medical care of patients.

The participants were randomly assigned to an FO intervention group, and the anthropometric data in the group comparison showed no significant differences. Therefore, it can be assumed that age, BMI, and foot posture did not influence the statistical results and differences between the intervention groups. Physiotherapeutic treatment was documented and only conducted in 7 out of 20 participants included in this study. The statistics of (29) did not show any significant influence on the results and is not expected to influence the gait kinematics after three-month intervention. The self-selected gait speed of the participants may have been criticized for a lack of standardization. Step and stride lengths were not included in the statistical analysis of this approach; however, the supervisor inspected the participants for a comparable gait speed in all settings. Therefore, the authors assumed that gait speed might have only a minor influence on the possible differences between the FO settings. Studies such as that by Takayanagi et al. (43) showed that the average daily gait speed was lower than the average gait speed in the laboratory, but the review of Fukuchi et al. (44) highlighted that the amplitude of spatiotemporal parameters increased at faster speeds. Assuming that the participants walked relatively faster than in everyday life, this could have an amplifying effect on the kinematic findings. In the gait test, the authors did not standardize the footwear of the participants, which might have a biomechanical impact, but the same shoes had to be worn at both time of measurements.

4.3 Strengths and limitations

This RCT provides valuable insights into the short- and long-term kinematic effects of FOs in individuals with PFP. A major strength is the inclusion of both SMFO and BMFO, allowing for a comparative assessment of different FO approaches. The study highlights the role of adaptation processes in gait mechanics, suggesting the necessity of a familiarization phase when assessing the long-term impact of FOs. Medical examinations included the necessary diagnostic tools and examinations by the physician as proposed by Fulkerson (45). However, the diagnosis of PFP involves different symptoms and manifestations, and it is ultimately impossible to definitively prove that the functional causes of the disease are found in movements, such as altered tibiofemoral or patellofemoral mechanics (46). A major limitation, which is why studies on custom-made FOs in general and SMFOs in particular are limited, is the individuality of human anatomy and movement. Custom-made FOs must be individually adapted to the anatomical and physiological conditions of the patients. This makes comparability between participants even more difficult (19). While the current study has focused on the temporal effects in the sagittal plane of each joint, the specific contribution of each joint to gait pattern differences remains unclear. In addition, kinetic changes play a major role in PFP patients (11). Xu et al. (47) have shown that kinetic metrics play a greater than 50% role in identifying differences in gait patterns. Xu et al. (48) demonstrated that the ankle and knee joints, especially in the sagittal and transverse planes, provide crucial information for distinguishing gait features. These findings hold relevance for future studies examining FO effectiveness.

From a methodological perspective, in the biomechanical gait analysis of patients, optical Motion Capture (MoCap) systems (OMC), such as the 3D OMC from Vicon (Vicon Motion Systems Limited, Yarnton, England) or Qualisys (Qualisys AB, Göteborg, Sweden) are considered the gold standard (49). IMUs are one of the main tools used for instrumented gait analysis (50) and have been shown to have high accuracy in the sagittal plane and moderate accuracy in the frontal and transverse planes (51). Kobsar et al. (52) found that IMUs provide more accurate estimates of sagittal joint angles in the lower limbs compared to frontal or transverse angles, though it's important to note that much of this evidence remains limited. While joint kinematics generally show good-to-excellent validity and reliability in the sagittal and frontal planes, the data often come from small studies with weak statistical measures. However, the use of IMUs has the advantage that gait analyses can be carried out quickly and on site in doctors' offices, which increases the practicality and compliance of patients. In contrast to optical MoCap systems, the IMU measurement method has the disadvantage of representing the foot only as a single segment, which provides an incomplete picture of the ankle joint movement. Because FOs are expected to have a biomechanical influence on the ankle joints and this influence might also be different between the hindfoot and tibia, as well as the forefoot and hindfoot, the use of a 2-segment foot model as developed by Bauer et al. (50) might be interesting. A detailed analysis of hindfoot and midfoot motion is currently not feasible with IMU technology, as the foot and ankle are represented by just one sensor, which models them as a single rigid segment (50). To advance this field, future research must focus on improving measurement methods to generate higher-quality evidence and recommendations for these kinematic outcomes (52).

4.4 Future studies

In general, more RCTs are needed to investigate the kinematic changes achieved through FOs in patients with isolated foot deformities, such as pes valgus and/or planus, to compare a one-size-fits-all approach with FOs. Therefore, in case reports that do not meet the statistical standards of this study, the discrete data of individual cases with a homogeneous foot deformity pattern needs to be further investigated. Future study methodologies might be enhanced by incorporating machine learning techniques such as (53).

Additionally, larger sample sizes with balanced sexes and older age groups should be investigated in future studies. Further, more research must be conducted regarding different indications for FO treatment, as there is still no consensus in science regarding when and to what extent FOs can be used for the orthopedic treatment of different lower extremity pathologies. Musculoskeletal modeling and simulations significantly deepens the understanding of human movement and should be integrated into future efforts aiming to explore the kinetic impact of FOs on the knee joint (47, 54), particularly in patients with PFP. There is a need for further longitudinal studies investigating not only the short- but also long-term effects on the clinical and biomechanical parameters of patients.

5 Conclusion

In conclusion, it can be assumed that SMFOs and BMFOs both showed significant long-term effects on the ankle, knee and hip joints in PFP patients. The extent to which these changes in movement have a positive effect in the treatment of patellofemoral pain syndrome and contribute to pain reduction requires further investigation. In contrast to our hypotheses, no significant short-term effects were statistically assessed. Therefore, temporal adaption processes for custom-made FOs should be considered in clinical care. The results of this RCT further enhance the evidence base for improving the care of patients with PFP and foot malalignment using custom-made FOs. Future studies should investigate kinematic adaptions in the lower extremities induced by FOs with consistent foot malalignment and isolated pathologies

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 authors.

Ethics statement

The studies involving humans were approved by Ethics Committee of the Kaiserslautern-Landau Sozialwissenschaften. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s), and minor(s)' legal guardian/next of kin, for the publication of any potentially identifiable images or data included in this article.

Author contributions

SS: Conceptualization, Data curation, Formal analysis, Funding acquisition, Investigation, Methodology, Project administration, Resources, Software, Supervision, Validation, Visualization, Writing – original draft, Writing – review & editing. JD: Data curation, Formal analysis, Methodology, Software, Validation, Visualization, Writing – original draft, Writing – review & editing. OL: Conceptualization, Writing – review & editing. CD: Formal analysis, Visualization, Writing – review & editing. EB: Conceptualization, Writing – review & editing. MF: Resources, Supervision, Writing – review & editing. SB: Conceptualization, Project administration, Supervision, Writing – review & editing.

Funding

The author(s) declare that no financial support was received for the research and/or publication of this article.

Acknowledgments

The authors want to thank Andreas Heine for patient acquisition and medical examination and Stefan Woltring for manufacturing the foot orthoses.

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.

Generative AI statement

The author(s) declare that no Generative AI was used in the creation of this manuscript.

Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2025.1546821/full#supplementary-material

References

1. Skou ST, Hojgaard L, Simonsen OH. Customized foot insoles have a positive effect on pain, function, and quality of life in patients with medial knee osteoarthritis. *J Am Podiatr Med Assoc.* (2013) 103:50–5. doi: 10.7547/1030050

2. Sesma AR, Mattacola CG, Uhl TL, Nitz AJ, McKeon PO. Effect of foot orthotics on single-and double-limb dynamic balance tasks in patients with chronic ankle instability. *Foot Ankle Spec.* (2008) 1:330–7. doi: 10.1177/1938640008327516

3. Castro-Méndez A, Munuera PV, Albornoz-Cabello M. The short-term effect of custom-made foot orthoses in subjects with excessive foot pronation and lower back pain: a randomized, double-blinded, clinical trial. *Prosthet Orthot Int.* (2013) 37:384–90. doi: 10.1177/0309364612471370

4. Shih Y-F, Wen Y-K, Chen W-Y. Application of wedged foot orthosis effectively reduces pain in runners with pronated foot: a randomized clinical study. *Clin Rehabil.* (2011) 25:913–23. doi: 10.1177/0269215511411938

5. Werd MB, Knight EL, Langer PR. *Athletic Footwear and Orthoses in Sports Medicine*. Cham: Springer (2010).

6. Kerkhoff A, Wagner H, Nagel A, Möller M, Peikenkamp K. Effects of two different foot orthoses on muscle activity in female during single-leg landing. *Ger J Exerc Sport Res.* (2017) 4:305–14. doi: 10.1007/s12662-017-0474-z

7. Mündermann A, Nigg BM, Stefanyshyn DJ, Humble RN. Development of a reliable method to assess footwear comfort during running. *Gait Posture.* (2002) 16:38–45. doi: 10.1016/S0966-6362(01)00197-7

8. McCulloch MU, Brunt D, Vander Linden D. The effect of foot orthotics and gait velocity on lower limb kinematics and temporal events of stance. *J Orthop Sports Phys Ther.* (1993) 17:2–10. doi: 10.2519/jospt.1993.17.1.2

9. Crossley KM, van Middelkoop M, Callaghan MJ, Collins NJ, Rathleff MS, Barton CJ. 2016 patellofemoral pain consensus statement from the 4th international patellofemoral pain research retreat, Manchester. Part 2: recommended physical interventions (exercise, taping, bracing, foot orthoses and combined interventions). *Br J Sports Med.* (2016) 50:844–52. doi: 10.1136/bjsports-2016-096268

10. Becker S, Simon S, Mühlen J, Dindorf C, Fröhlich M. Assessing the subjective effectiveness of sensorimotor insoles (SMIs) in reducing pain: a descriptive multicenter pilot study. *J Funct Morphol Kinesiol.* (2023) 8:66. doi: 10.3390/jfmk8020066

11. Bazett-Jones DM, Neal BS, Legg C, Hart HF, Collins NJ, Barton CJ. Kinematic and kinetic gait characteristics in people with patellofemoral pain: a systematic review and meta-analysis. *Sports Med.* (2023) 53:519–47. doi: 10.1007/s40279-022-01781-1

12. Kellish AS, Kellish P, Hakim A, Miskiel S, Shahi A, Kellish A. What is the effect on kinesio taping on pain and gait in patients with patellofemoral pain syndrome? *Cureus.* (2020) 12:1–7. doi: 10.7759/cureus.8982

13. Fairbank J, Pynsent P, van Poortvliet JA, Phillips H. Mechanical factors in the incidence of knee pain in adolescents and young adults. *J Bone Joint Surg Br.* (1984) 66:685–93. doi: 10.1302/0301-620X.66B5.6501361

14. Collins N, Crossley K, Beller E, Darnell R, McPoil T, Vicenzino B. Foot orthoses and physiotherapy in the treatment of patellofemoral pain syndrome: randomised clinical trial. *Br Med J.* (2008) 337:1–8. doi: 10.1136/bmj.a1735

15. Stefanyshyn DJ, Stergiou P, Lun VM, Meeuwisse WH, Worobets JT. Knee angular impulse as a predictor of patellofemoral pain in runners. *Am J Sports Med.* (2006) 34:1844–51. doi: 10.1177/0363546506288753

16. Saxena A, Haddad J. The effect of foot orthoses on patellofemoral pain syndrome. *J Am Podiatr Med Assoc.* (2003) 93:264–71. doi: 10.7547/87507315-93-4-264

17. Lewinson RT, Wiley JP, Humble RN, Worobets JT, Stefanyshyn DJ. Altering knee abduction angular impulse using wedged insoles for treatment of patellofemoral pain in runners: a six-week randomized controlled trial. *PLoS One.* (2015) 10:e0134461. doi: 10.1371/journal.pone.0134461

18. Gross MT, Foxworth JL. The role of foot orthoses as an intervention for patellofemoral pain. *J Orthop Sports Phys Ther.* (2003) 33:661–70. doi: 10.2519/jospt.2003.33.11.661

19. Becker S, Ludwig O, Woltring S, Simon S, Fröhlich M. *Sensomotorische Einlagen: Grundlagen und Funktionen*. Berlin, Heidelberg: Springer-Verlag (2024).

20. Ranker A. Sensomotorische schuheinlagen–wirkprinzipien und evidenz. *manuelletherapie.* (2020) 24:168–72. doi: 10.1055/a-1213-0208

21. Greitemann B, Franzen M, Stinus H, Walther M, Dierolf W, Schievink F, et al. DGOOC-Beratungsausschuss Orthopädieschuhtechnik. *Orthopädie Unfallchirurgie-Mitteilungen Nachrichten.* (2016) 5:283–6. doi: 10.1055/s-0042-111222

22. Ludwig O, Kelm J, Fröhlich M. The influence of insoles with a peroneal pressure point on the electromyographic activity of tibialis anterior and peroneus longus during gait. *J Foot Ankle Res.* (2016) 9:1–9. doi: 10.1186/s13047-016-0162-5

23. Schmitt AP-L, Liebau K-H, Hamm A, Hacke C, Mittelmeier W, Schulze C. Comparison of the influence of supportive and sensorimotor insoles in the muscle activity of Tibialis anterior and peroneus longus in combat boots. *The Foot.* (2022) 52:101910. doi: 10.1016/j.foot.2022.101910

24. Klein T, Lastovicka O, Janura M, Svoboda Z, Chapman GJ, Richards J. The immediate effects of sensorimotor foot orthoses on foot kinematics in healthy adults. *Gait Posture.* (2021) 84:93–101. doi: 10.1016/j.gaitpost.2020.11.022

25. Mabuchi A, Kitoh H, Inoue M, Hayashi M, Ishiguro N, Suzuki N. The biomechanical effect of the sensomotor insole on a pediatric intoeing gait. *Int Sch Res Notices.* (2012) 2012:1–5. doi: 10.5402/2012/396718

26. Dankerl P, Keller AK, Häberle L, Stumptner T, Pfaff G, Uder M, et al. Effects on posture by different neuromuscular afferent stimulations and proprioceptive insoles: rasterstereographic evaluation. *Prosthet Orthot Int.* (2016) 40:369–76. doi: 10.1177/0309364614554031

27. Aysin IK, Askin A, Mete BD, Guvendi E, Aysin M, Kocyigit H. Investigation of the relationship between anterior knee pain and chondromalacia patellae and patellofemoral malalignment. *Eurasian J Med.* (2018) 50:28. doi: 10.5152/eurasianjmed.2018.17277

28. Kerkhoff A, Wagner H, Peikenkamp K. Different effects of sensorimotor and soft bedding foot orthoses on muscle activity during single-leg landing in sports. *Footwear Sci.* (2019) 11:S37–8. doi: 10.1080/19424280.2019.1606065

29. Simon S, Heine A, Dully J, Dindorf C, Ludwig O, Fröhlich M, et al. Pain reduction in patellofemoral knee patients during three-month intervention with biomechanical and sensorimotor foot orthoses: a randomized controlled clinical study. *Preprints.* (2024) 13(1):38. doi: 10.20944/preprints202412.0157.v1

30. Faul F, Erdfelder E, Buchner A, Lang A-G. Statistical power analyses using G* power 3.1: tests for correlation and regression analyses. *Behav Res Methods.* (2009) 41:1149–60. doi: 10.3758/BRM.41.4.1149

31. Schiphof D, Boers M, Bierma-Zeinstra SM. Differences in descriptions of Kellgren and Lawrence grades of knee osteoarthritis. *Ann Rheum Dis.* (2008) 67:1034–6. doi: 10.1136/ard.2007.079020

32. Al-Amri M, Nicholas K, Button K, Sparkes V, Sheeran L, Davies JL. Inertial measurement units for clinical movement analysis: reliability and concurrent validity. *Sensors.* (2018) 18:719. doi: 10.3390/s18030719

33. Nijmeijer EM, Heuvelmans P, Bolt R, Gokeler A, Otten E, Benjaminse A. Concurrent validation of the Xsens IMU system of lower-body kinematics in jump-landing and change-of-direction tasks. *J Biomech.* (2023) 154:111637. doi: 10.1016/j.jbiomech.2023.111637

34. Hortobágyi T, Lesinski M, Gäbler M, VanSwearingen JM, Malatesta D, Granacher U. Effects of three types of exercise interventions on healthy old adults' gait speed: a systematic review and meta-analysis. *Sports Med.* (2015) 45:1627–43. doi: 10.1007/s40279-015-0371-2

35. Pataky TC, Vanrenterghem J, Robinson MA. Zero-vs. one-dimensional, parametric vs. non-parametric, and confidence interval vs. hypothesis testing procedures in one-dimensional biomechanical trajectory analysis. *J Biomech.* (2015) 48:1277–85. doi: 10.1016/j.jbiomech.2015.02.051

36. Laštovička O, Klein T, Abrantes J, Janura M. Immediate effect of individual bars of insoles and their combination on gait parameters in asymptomatic healthy adults. *Somatosens Mot Res.* (2020) 37:125–31. doi: 10.1080/08990220.2020.1753686

37. Becker S, Ludwig O, Woltring S, Simon S, Fröhlich M. Wissenschaftliche studien. In: *Sensomotorische Einlagen: Grundlagen und Funktionen*. Berlin, Heidelberg: Springer-Verlag (2024). p. 33–40.

38. Moisan G, Robb K, Mainville C, Blanchette V. Effects of foot orthoses on the biomechanics of the lower extremities in adults with and without musculoskeletal disorders during functional tasks: a systematic review. *Clin Biomech (Bristol).* (2022) 95:105641. doi: 10.1016/j.clinbiomech.2022.105641

39. Hsu W-C, Jhong Y-C, Chen H-L, Lin Y-J, Chen L-F, Hsieh L-F. Immediate and long-term efficacy of laterally-wedged insoles on persons with bilateral medial knee osteoarthritis during walking. *Biomed Eng Online.* (2015) 14:1–16. doi: 10.1186/s12938-015-0040-6

40. Radwan NL, Ibrahim MM, Eid MA, Aly SM. The long-term effect of foot insoles on kinetic gait parameters in female children with flexible flat foot. *IMJ.* (2020):485–94.

41. Leung J, Moseley A. Impact of ankle-foot orthoses on gait and leg muscle activity in adults with hemiplegia: systematic literature review. *Physiotherapy.* (2003) 89:39–55. doi: doi: 10.1016/S0031-9406(05)60668-2

42. Hoglund LT, Hulcher TA, Amabile AH. Males with patellofemoral pain have altered movements during step-down and single-leg squatting tasks compared to asymptomatic males: a cross-sectional study. *Health Sci Rep.* (2024) 7:e2193. doi: 10.1002/hsr2.2193

43. Takayanagi N, Sudo M, Yamashiro Y, Lee S, Kobayashi Y, Niki Y, et al. Relationship between daily and in-laboratory gait speed among healthy community-dwelling older adults. *Sci Rep.* (2019) 9:3496. doi: 10.1038/s41598-019-39695-0

44. Fukuchi CA, Fukuchi RK, Duarte M. Effects of walking speed on gait biomechanics in healthy participants: a systematic review and meta-analysis. *Syst Rev.* (2019) 8:1–11. doi: 10.1186/s13643-019-1063-z

45. Fulkerson JP. Diagnosis and treatment of patients with patellofemoral pain. *Am J Sports Med*. (2002) 30:447–56. doi: 10.1177/03635465020300032501

46. Salsich GB, Perman WH. Tibiofemoral and patellofemoral mechanics are altered at small knee flexion angles in people with patellofemoral pain. *J Sci Med Sport*. (2013) 16:13–7. doi: 10.1016/j.jsams.2012.04.003

47. Xu D, Zhou H, Quan W, Ma X, Chon T-E, Fernandez J, et al. New insights optimize landing strategies to reduce lower limb injury risk. *Cyborg and Bionic Systems*. (2024) 5:0126. doi: 10.34133/cbsystems.0126

48. Xu D, Quan W, Zhou H, Sun D, Baker JS, Gu Y. Explaining the differences of gait patterns between high and low-mileage runners with machine learning. *Sci Rep*. (2022) 12:2981. doi: 10.1038/s41598-022-07054-1

49. Ludwig O, Becker S, Fröhlich M. Einführung in die ganganalyse. In: *Einführung in die Ganganalyse: Grundlagen, Anwendungsgebiete, Messmethoden*. Berlin, Heidelberg: Springer (2022). p. 1–20. Available at: https://link.springer.com/chapter/10.1007/978-3-662-66235-9_1

50. Bauer L, Hamberger MA, Böcker W, Polzer H, Baumbach SF. Development of an IMU based 2-segment foot model for an applicable medical gait analysis. *BMC Musculoskelet Disord*. (2024) 25:606. doi: 10.1186/s12891-024-07719-0

51. Zeng Z, Liu Y, Li P, Wang L. Validity and reliability of inertial measurement units measurements for running kinematics in different foot strike pattern runners. *Front Bioeng Biotechnol*. (2022) 10:1005496. doi: 10.3389/fbioe.2022.1005496

52. Kobsar D, Charlton JM, Tse CT, Esculier J-F, Graffos A, Krowchuk NM, et al. Validity and reliability of wearable inertial sensors in healthy adult walking: a systematic review and meta-analysis. *J Neuroeng Rehabil*. (2020) 17:1–21. doi: 10.1186/s12984-020-00685-3

53. Xu D, Zhou H, Quan W, Jiang X, Liang M, Li S, et al. A new method proposed for realizing human gait pattern recognition: inspirations for the application of sports and clinical gait analysis. *Gait Posture*. (2024) 107:293–305. doi: 10.1016/j.gaitpost.2023.10.019

54. Uhlrich SD, Uchida TK, Lee MR, Delp SL. Ten steps to becoming a musculoskeletal simulation expert: a half-century of progress and outlook for the future. *J Biomech*. (2023) 154:111623. doi: 10.1016/j.jbiomech.2023.111623

55. Roth S, Roth A, Jotanovic Z, Madarevic T. Navicular index for differentiation of flatfoot from normal foot. *Int Orthop*. (2013) 37:1107–12. doi: 10.1007/s00264-013-1885-6

56. Cavanagh PR, Rodgers MM. The arch index: a useful measure from footprints. *J Biomech*. (1987) 20:547–51. doi: 10.1016/0021-9290(87)90255-7

frontiers | Frontiers in Sports and Active Living

TYPE Original Research
PUBLISHED 09 May 2025
DOI 10.3389/fspor.2025.1546297

OPEN ACCESS

EDITED BY
Lida Mademli,
Aristotle University of Thessaloniki, Greece

REVIEWED BY
Datao Xu,
Ningbo University, China
Ting Long,
The University of Auckland, New Zealand

*CORRESPONDENCE
Kate Pfile
✉ pfilekr@cofc.edu

RECEIVED 16 December 2024
ACCEPTED 23 April 2025
PUBLISHED 09 May 2025

CITATION
Pfile K, Prosser B, Slone H, McLeod M, Gregory C and Hunnicutt J (2025) Walking gait biomechanics in individuals with quadriceps tendon autograft anterior cruciate ligament reconstruction.
Front. Sports Act. Living 7:1546297.
doi: 10.3389/fspor.2025.1546297

Walking gait biomechanics in individuals with quadriceps tendon autograft anterior cruciate ligament reconstruction

Kate Pfile[1]*, Bennett Prosser[1], Harris Slone[2], Michelle McLeod[3], Chris Gregory[4] and Jennifer Hunnicutt[5]

[1]Department of Health and Human Performance, College of Charleston, Charleston, SC, United States, [2]Orthopedics and Physical Medicine & Rehabilitation, Medical University of South Carolina, Charleston, SC, United States, [3]Arthritis Foundation, Atlanta, GA, United States, [4]Health Sciences and Research, Medical University of South Carolina, Charleston, SC, United States, [5]Hunnicutt Writing and Consulting, LLC, Panama City, FL, United States

Introduction: Walking is a vital movement, corresponding to physical activity, health, and independent living. Persistent abnormal lower extremity kinetics and kinematics during walking may influence long-term joint health. Anterior cruciate ligament (ACL) injuries are common sport-related knee joint injuries resulting in short- and long-term dysfunctional movement patterns. Re-establishing normal gait biomechanical patterns following ACL reconstruction (ACLR) is a universal long-term rehabilitative goal and indicator of restored function. The use of the quadriceps tendon (QT) graft technique by orthopedic surgeons is increasing and growing evidence suggests it's viable for ACLR. However, no information is available examining walking gait biomechanics in QT-ACLR patients. Our study evaluated three-dimensional hip and knee joint biomechanics during the stance phase of walking gait in patients with QT-ACLR by comparing the ACLR and nonsurgical limbs. We hypothesized hip and knee joint biomechanics will differ between the QT-ACLR and nonsurgical limbs during the stance phase of gait.
Methods: We recruited a convenience sample of 14 patients with unilateral QT-ACLR ∼11 months post-surgery from an orthopedic surgery clinic. Three-dimensional hip and knee kinematics and kinetics and vertical ground reaction force were assessed while participants walked at self-selected speeds. Data were time-normalized from 0%–100% (% stance phase), and ACLR and nonsurgical limbs were compared using curve analyses with 95% confidence intervals. Cohen's d effect sizes identified clinical differences between limbs.
Results: The ACLR limb was significantly different from the nonsurgical limb for knee flexion angle (1%–8% and 58%–85%), knee flexion moment (14%–23%), hip flexion moment (60%–67%), knee adduction angle (9%–32%, 92%–100%), knee adduction moment (53%–81%), hip frontal plane angle (0%–100%), hip abduction moment (31%–35% and 71%–76%), knee external rotation angle (0%–100%), knee internal rotation moment (55%–84%), hip transverse plane angle (20%–39% and 88%–100%), and hip internal rotation moment (56%–88%). All significant findings had large effect sizes ($d > 0.8$).

Discussion: Three-dimensional biomechanical gait alterations are present at the knee and hip following QT-ACLR when comparing between limbs. This pattern is consistent with other ACLR graft types. Participants demonstrated gait patterns associated with quadriceps avoidance and reduced proximal forces during the loading response and terminal stance phases. Rehabilitation and functional movement programs should target these deficits.

KEYWORDS

kinematics, kinetics, vertical ground reaction force, anterior cruciate ligament reconstruction (ACLR), gait biomechanics

1 Introduction

Injury to the anterior cruciate ligament (ACL) is one of the most common (1, 2) and costly (3, 4) sport-related knee joint injuries. ACL reconstruction (ACLR) attempts to restore stability and promote a return to normal kinematic and kinetic function. Autograft ACLR most commonly includes harvesting a graft from either the patellar, hamstring, or quadriceps tendons. Bone-patellar tendon-bone (PT) and hamstring tendon (HT) grafts are most popular. Use of PT autografts can lead to issues following surgery such as reduced instrumental laxity, anterior knee pain, and knee extensor strength deficits, while the HT graft can result in poorer patient reported outcome measures and knee flexor strength deficits (5, 6). The quadriceps tendon (QT) graft was introduced by Marshall et al. in 1979 (7) and has since gained traction with approximately 10% of orthopedic surgeons in 2020 reportedly using it for ACLR due to positive patient-reported outcomes, increased knee stability, and decreased risk of re-injury (8). A recent systematic review showed that anterior knee pain outcomes are similar between HT and QT grafts with higher pain scores for patellar tendon grafts. Further, QT grafts had similar functional outcomes and revision rates compared to PT and HT autografts (5).

Walking is one of the most essential human movements, corresponding to an active lifestyle, better overall health, and longer independent living (9, 10). Walking requires coordination across different systems and joints. There is consistent evidence that walking biomechanics differ following ACLR, specifically comparing between limb knee kinematics (11), kinetics (11–15), and vertical ground reaction force (vGRF) (16, 17). However, due to the relative novelty of the QT graft option, the literature is lacking a description of knee biomechanics following QT-ACLR during walking gait. As QT autografts become a more prevalent surgical option, information regarding repetitive, functional movements like walking gait are needed to determine if current rehabilitation and functional return considerations are germane following this surgery. Further, examining hip joint biomechanics may identify compensatory movement and loading patterns that contribute to lower extremity dysfunction (18). Abnormal gait, a deviation in temporal-spatial, kinematic, kinetic, or muscle activation patterns from an expected pattern (19), can add to the degeneration of articular cartilage especially in the knee joint, further reducing vGRF absorption and accelerating the onset of posttraumatic osteoarthritis (20, 21). It is reported that ten years following ACL injury approximately one-third of ACLR knees developed posttraumatic knee osteoarthritis, however no patients with QT-ACLR were examined (22).

Curve analysis is beneficial when evaluating gait as it allows the observation of the overall movement pattern rather than conducting analyses at discrete points such as the peak angle or force within a subphase of gait (23). Specifically, curve analysis can reveal differences at key subphases during stance such as loading response, midstance, terminal stance, and preswing. This allows practitioners to better recognize deficiencies and develop appropriate interventions (24). The purpose of our study was to evaluate three-dimensional hip and knee joint biomechanics during the stance phase of walking gait in patients with QT-ACLR by comparing the ACLR and nonsurgical limbs. We hypothesized that hip and knee joint biomechanics will differ between the QT-ACLR and nonsurgical limbs during the stance phase of gait.

2 Materials and methods

2.1 Study design

This investigation was part of a larger study examining functional outcomes of ACLR (31) that received IRB approval from Medical University of South Carolina and all participants provided informed consent to complete patient reported outcomes and a walking gait analysis in the biomechanics lab. Prior to their enrollment in the research study, all participants underwent primary ACLR with QT autografts performed by a single orthopedic surgeon. Grafts were harvested with a minimally invasive, all inside technique preventing the use of bone plugs. Suspensory fixation was performed on the femoral tibial side (25).

2.2 Participants

Participants were recruited from an orthopedic surgeon's clinic as a sample of convenience. Study staff reviewed the health records of patients with prior ACLR for the inclusion/exclusion criteria. Study staff called patients who met the following criteria to request their interest in participating in this research study. The inclusion criteria consisted of the following: 14–55 years of age, history of unilateral, isolated ACLR (with or without concomitant meniscus pathology) between six months to two years post reconstructive procedure using ipsilateral autografts harvested from the QT, and ACLR performed by a fellowship-trained orthopedic surgeon (SH). Exclusion criteria included history of lower extremity injury or surgery, including ACL retears and revisions within the past 6 months, multi-ligament reconstructions, an inability to walk without assistance from an orthotic, knee brace, or another person, and self-reported knee arthritis that would limit range of motion at the knee.

2.3 Procedures

Demographic measures including anthropometrics, the Tegner Activity Scale, time since surgery and presence of concomitant meniscus surgery and patient reported outcome measures including the Lysholm Score, International Knee Documentation Committee Subjective Knee Form (IKDC) and Knee Injury and Osteoarthritis Outcome Score (KOOS) were collected for each patient. A lower extremity biomechanical assessment was completed using an active marker set (PhaseSpace Motion Capture, Phase Space, Inc. San Leandro, CA). A single experienced Certified Athletic Trainer (JH) performed all testing, placing markers on the pelvis and lower extremities (ASIS, greater trochanters, medial and lateral epicondyles, medial and lateral malleoli, 1st and 5th metatarsals and distal 2nd toe) with marker clusters on the sacrum, anterior mid-femur, lateral mid-shanks, and dorsal mid-feet. Participants walked in their preferred walking shoes on a split-belt instrumented treadmill (Bertec, Columbus, OH) at a self-selected walking speed. Participants were given time prior to collecting the data to familiarize themselves with treadmill walking. Once familiar, data capture trials began where participants walked for at least 10 s to reach a steady state, followed by a 30 s period for data collection. Three successful 30 s trials were collected, and patients were given at least 30 s of rest between each trial.

2.4 Data processing

Three-dimensional kinematics were recorded at 120 Hz with a 16-camera motion capture system. Coordinates were interpolated over gaps smaller than 20 samples and resampled at 100 Hz. Bilateral GRFs were recorded at 2000 Hz. Data were then filtered using a 4th order Savitzky-Golay filter acting on 21 data points, resampled at 100 Hz and used to identify gait events during treadmill walking. The stance phase was defined as the period between initial contact and toe off, with thresholds defined as vGRF >20N and <20N, respectively. Kinetic and kinematic variables were normalized for time and reduced to 100 data points depicting 1%–100% of the stance phase of the gait cycle. Kinetic data were calculated using inverse dynamics of a six degree of freedom segment model. The six degrees of freedom refers to the three planes of motion (sagittal, frontal, and transverse planes) at both the hip and knee joints. Vertical ground reaction forces were normalized to body mass while joint moments were normalized by body mass and height and reported as external moments. Joint angles were reported in degrees. All data were processed and analyzed using custom LabVIEW (National Instruments, Austin, USA) programs.

2.5 Statistical analysis

A curve analysis was performed to identify between-limb differences across the stance phase of gait (23). The phases of the gait cycle were interpreted as 1%–16.6% loading response, 16.7%–50% midstance, 50.1%–83.2% terminal stance, 83.3%–100% preswing (26). For each kinetic and kinematic variable, the mean and 95% confidence interval (CI) was calculated and graphed for the nonsurgical and QT-ACLR limbs to identify between limb differences. No overlap in the between-limb confidence intervals for three consecutive percentages during the stance phase indicated a statistically significant difference ($P < 0.05$). All graphs were created using Microsoft Excel (Microsoft Corporation, Redmond, WA). Cohen's d effect sizes with pooled standard deviations were calculated using the average mean and standard deviation during the significantly different window for each statistically significant variable. Effect sizes were interpreted as small ($d < 0.2$), medium ($d = 0.5$), or large ($d \geq 0.8$) (27).

3 Results

3.1 Demographics

The study participants included eleven males and three females, as shown in Table 1. The average time between the data collection period and surgery was 10.8 ± 5.8 months. The reconstructions were performed on six left knees and eight right knees, and all participants were right leg dominant. Averages of self-reported outcome measures are reported in Table 1.

3.2 Overview of main findings

There were significant differences in kinematics between the QT-ACLR limb and nonsurgical limb and each difference had a large effect size (Table 2). There were significant differences with large effect sizes in external joint moments between limbs for all three planes at both the hip and knee. There was no significant difference between limbs for vGRF.

3.2.1 Sagittal plane

At the knee (Figure 1), there was an increase in flexion angle in the ACLR limb compared to the nonsurgical limb from 1%–8%

TABLE 1 Demographic data.

Demographic	Participant (Avg ± SD)
Sex (M/F)	11M 3F
Age (years)	25.9 ± 9.8
Mass (kg)	83.2 ± 16.3
Height (cm)	177.0 ± 11.0
BMI (kg/m^2)	25.0 ± 4.5
Months since surgery	10.8 ± 5.8
Tegner (post-ACLR at time of data collection)	6.6 ± 1.7
Self-selected walking speed (m/s)	0.82 ± 0.22
Lysholm	86.5 ± 7.5
KOOS- Sport	77.0 ± 15.6

TABLE 2 Differences in Hip and knee biomechanics between ACLR and nonsurgical limbs.

Biomechanical Variable	% Stance Phase	ACLR Limb Mean ± SD	Nonsurgical Limb Mean ± SD	Effect Size (95% CI)*
Kinematics (degrees)				
Knee flexion angle	1–8	6.6° ± 0.95	4.0° ± 1.4	2.2 (1.2, 3.1)
	58–85	6.6° ± 1.2	4.0° ± 1.5	1.9 (1.0, 2.8)
Knee adduction angle	9–32	6.4° ± 0.12	7.6° ± 0.25	−6.0 (−7.7, −4.2)
	58–81	4.6° ± 0.16	3.6° ± 0.27	4.7 (3.3, 6.2)
	92–100	6.5° ± 0.10	8.1° ± 0.70	−3.4 (−4.5, −2.2)
Knee transverse plane angle	0–100	14.2° ± 1.9	7.2° ± 2.0	−3.6 (−4.7, −2.4)
Hip frontal plane angle	0–100	1.1° ± 2.2	3.5° ± 2.0	1.2 (0.4, 2.0)
Hip ER angle	20–39	0.37° ± 1.1	2.0° ± 1.0	1.57 (0.7, 2.4)
Hip IR angle	88–100	5.1° ± 0.23	3.2° ± 0.13	−9.7 (−12.4, −7.1)
Kinetics (Nm/kg*m)				
Knee flexion moment	14–23	0.19 ± 0.08	0.28 ± 0.08	−1.2 (−1.9, −0.4)
Knee adduction moment	53–81	0.17 ± 0.01	0.23 ± 0.02	4.0 (2.7, 5.3)
Knee IR moment	55–84	0.06 ± 0.01	0.09 ± 0.01	2.1 (1.2, 3.0)
Hip flexion moment	60–67	0.26 ± 0.03	0.36 ± 0.03	−3.3 (−4.5, −2.2)
Hip adduction moment	31–35	0.71 ± 0.005	0.78 ± 0.005	13.5 (9.9, 17.0)
	71–76	0.62 ± 0.01	0.68 ± 0.01	6.3 (4.5, 8.0)
Hip ER moment	56–88	0.05 ± 0.01	0.08 ± 0.02	1.7 (0.8, 2.6)

Negative indicates a decreased moment or angle in the ACLR limb compared to the nonsurgical limb. Only effect sizes for statistically significant differences are reported. ER, external rotation; IR, Internal rotation.

and 58%–85% of the stance phase as well as a smaller knee flexion moment from 14%–23% of stance. Additionally, the ACLR limb had a reduced hip flexion moment from 60%–67% compared to the nonsurgical limb.

3.2.2 Frontal plane

The ACLR limb had a decreased knee adduction angle from 9%–32% and 92%–100% of the stance phase and an increased knee adduction angle from 58%–81% of the stance phase compared to the nonsurgical limb. There were decreases in knee adduction moment from 53%–81% compared to the nonsurgical limb. At the hip (Figure 2), in comparison to the nonsurgical limb, the ACLR limb was shifted toward more abduction and less adduction for the entirety of stance and showed a reduced hip abduction moment from 31%–35% and 74%–76%.

3.2.3 Transverse plane

There was a reduction in knee external rotation angle for the ACLR limb throughout all of stance compared to the nonsurgical limb and a decreased knee internal rotation moment from 55%–84% in the ACLR limb compared to the nonsurgical limb (Figure 1). The ACLR limb also had a decreased hip external rotation angle from 20%–39% of stance phase and an increased hip internal rotation angle at 88%–100% of stance phase compared to the nonsurgical limb. The ACLR limb had a decreased hip internal rotation moment from 56%–88% compared to the nonsurgical limb (Figure 2).

4 Discussion

The purpose of our study was to evaluate three-dimensional hip and knee joint biomechanics during the stance phase of walking gait in patients with QT-ACLR by comparing the ACLR and nonsurgical limbs. Consistent with our hypothesis, the QT-ACLR limb displayed abnormal biomechanics in both kinematic and kinetic variables during stance compared to the nonsurgical limb; where most differences took place during the force absorption and propulsive periods of gait. However, unlike previous research on patients with ACLR, there were no significant differences between limbs for vGRF (16, 17, 28).

4.1 Loading response

Previous studies investigating walking biomechanics in other ACLR populations identified between-limb differences in both hip and knee joint kinematics (11, 28). A meta-analysis shows that stance phase peak knee flexion angle is significantly lower in the ACLR limb compared to the nonsurgical limb during gait (11). However, it is not feasible to determine exactly where within the stance phase the peak angle occurred due to the nature of the data extraction for this meta-analysis. Participants in our study displayed abnormal gait through significantly more knee flexion (terminal extension avoidance) compared to their nonsurgical limb during the loading response phase to allow for weight acceptance (Figure 1). Further examination revealed the participants' ACLR limbs displayed less knee flexion excursion (the difference between the maximum and minimum recorded knee flexion angles) with an average of 29° compared to 33° in the nonsurgical limb, exceeding the minimally clinically important difference of 3° (29). This "stiffened-knee strategy" where participants displayed less knee flexion excursion was also observed in patients with PT-ACLR (16) and ACL-deficiency (ACLD) (29). Participants may refrain from loading the joint in terminal extension due to on-going apprehension, a quadriceps strength deficit, or inadequate neuromuscular control (30). However, when compared to patients with PT-ACLR, those with

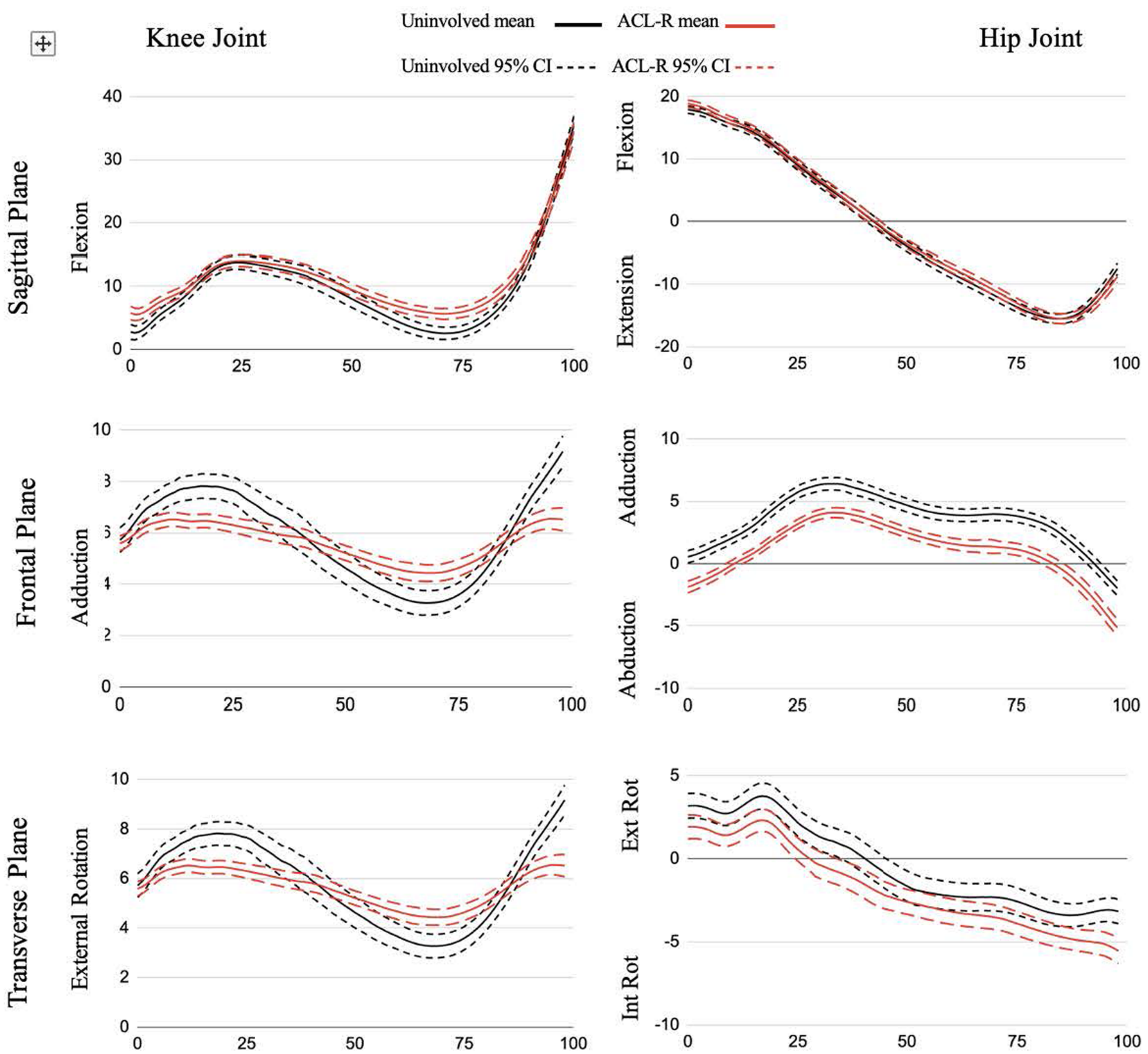

FIGURE 1
Joint angle ensemble curves for stance phase of gait. All data are presented in degrees on the y-axis across % stance phase on the x-axis.

QT-ACLR have demonstrated similar quadriceps cross-sectional area, isokinetic strength, and muscle activation, as well as functional hop test outcomes (31). Continued research of QT-ACLR functional outcomes beyond subjective questionnaires are needed in larger samples to better explain these findings. The literature shows that time since surgery can impact knee flexion angle in those who underwent ACLR (16, 28). Our study did not control for the specific time since surgery in which the gait analysis was conducted, thus it is possible that this variability (ranging from 6–23 months) may be a factor.

The ACLR limb had a decreased external knee flexion moment during the late loading response and early midstance phases when compared to the nonsurgical limb, as shown in Figure 1. This is the period when body weight is transferred to the lead stance leg and the head, arms, and trunk begin to align over the stance leg. Our finding is consistent with the existing gait literature examining peak knee flexion moment in patients with ACLR (11). Additionally, quadriceps weakness persists following ACLR (32) and previous research classifying ACLR patients (with a mixture of graft types) as weak or strong based on isometric quadriceps strength found those in the weak group displayed a reduced internal knee extension moment from 6%–72% of the stance phase (33). This suggests that diminished function of the quadriceps could also be the cause of reduced external knee flexion moment following QT-ACLR. In contrast, ACLR patients who completed criterion-based rehabilitation (including quadriceps strength symmetry >80%) showed no correlation between quadriceps strength asymmetry and internal knee extensor moment asymmetry (34). A recent meta-analysis (35) found quadriceps isokinetic strength limb symmetry was below 90% until 24 months post QT-ACLR. Further, there were no differences in strength between QT-ACLR and PT-ACLR populations serving as an explanation for why other studies including alternative graft types may also show an altered external knee flexion moment. Previously published research reported on this QT-ACLR patient sample identified asymmetry

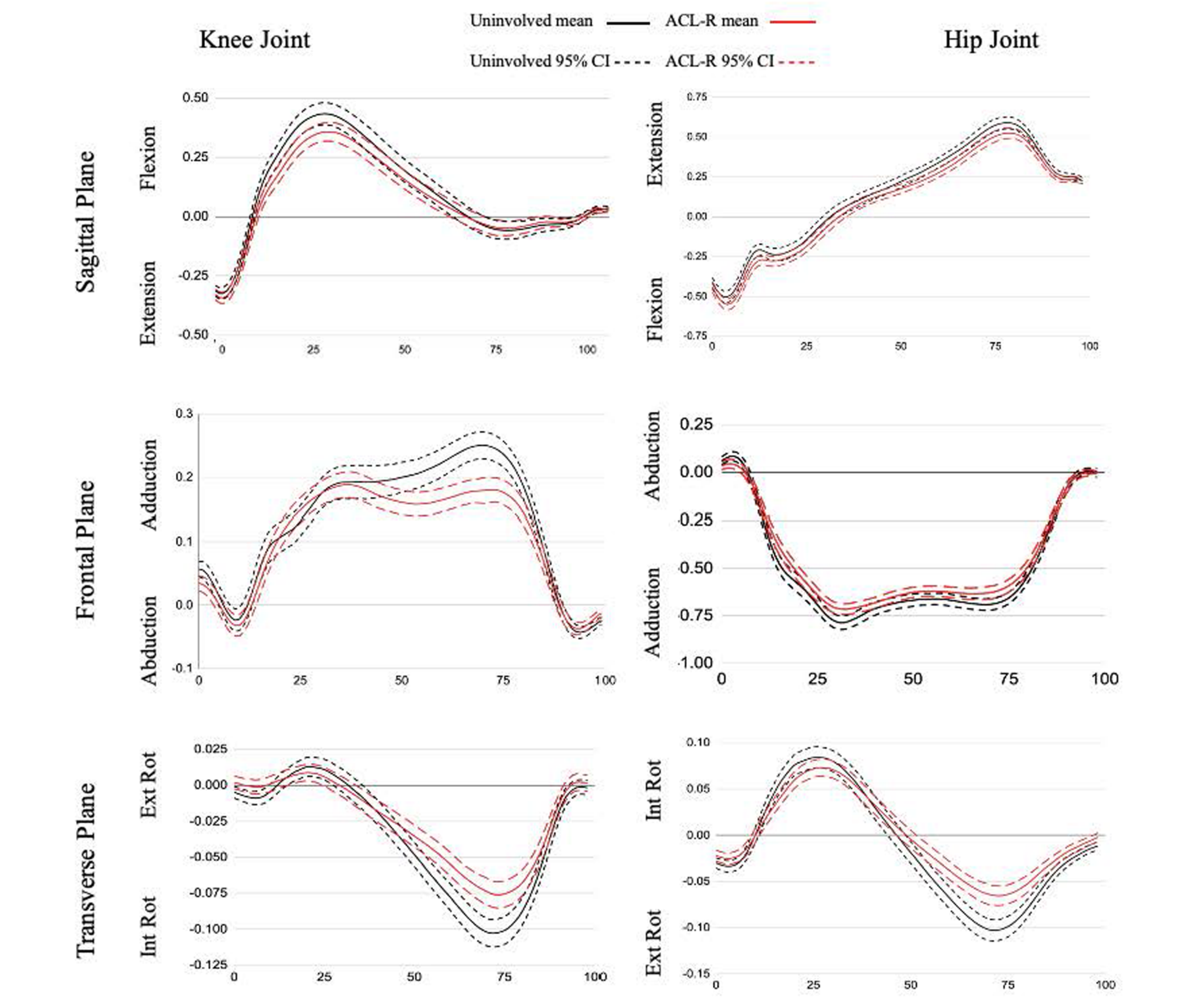

FIGURE 2
Joint moment ensemble curves for stance phase of gait. All data are presented as external moments (Nm/kg*m) on the y-axis across % stance phase on the x-axis.

in quadriceps muscle strength and cross-sectional area outcomes, despite a high limb symmetry index for muscle activation (blinded citation). QT-ACLR rehabilitation should continue to emphasize both eccentric and concentric quadriceps strengthening along with gait training to promote quadriceps loading through terminal knee extension as a foundation to restoring function. Further research is needed to conclude whether quadriceps strength deficits cause aberrant knee biomechanics during walking and how these adaptations may affect long-term joint health.

4.2 Terminal stance

Participants in our study displayed significantly more knee flexion (terminal extension avoidance) compared to their nonsurgical limb during the loading response phase to allow for weight acceptance and the terminal stance phase to initiate propulsion (Figure 1). This latter phase difference is a similar finding to the early ACLR group (defined as 6 months-24 months post-ACLR) described by Goetschius et al. (28), and Gao et al. (36) who included a combination of ACLR auto- and allograft patients, and the weak quadriceps group in Pietrosimone et al. (33).

We also found decreased external knee adduction and internal rotation moments in the ACLR limb during the terminal stance phase. The reduced knee adduction moment occurs at the same time as the flattening of the knee adduction angle curve suggesting the kinematic pattern may be driving the kinetic pattern where a less adducted joint results in a lower joint torque when there is no difference in the external vGRF between limbs (37). Overall, the reduced joint moment peaks suggest an adaptive pattern to unload the joint and indicates a less dynamic strategy for frontal and transverse plane force management in the ACLR limb (Figure 2) (28). The ACL is loaded under an internal tibial rotation torque coupled with small (0–30) knee flexion angles (38). The significant transverse plane difference occurred between 55%–84% of the stance phase, corresponding to a gait phase with low knee flexion angles. Therefore, the reduction in transverse plane moment may be an effort to protect the ACL from high torques.

There are fewer studies reporting hip joint moments during gait in patients with ACLR. A meta-analysis synthesized 27 total studies (39), and only one (14) reported hip sagittal plane kinetics in comparison to a control group. Further, there were no results for frontal or transverse plane moments. More recently, Goetschius et al. (28) reported no differences in hip joint moments between ACLR and nonsurgical limbs. However, we found differences between limbs in all three planes during the terminal stance phase. Terminal stance is designated by the proximal leg advancing forward over the foot while the trunk also moves ahead of the support leg (26). Our results showed the QT-ACLR limb had reduced external hip extension and external rotation moments and a flattening of the hip adduction moment in both peaks during loading response and terminal stance. The reduced hip external rotation moment during the terminal stance phase may represent a neuromuscular adaptation to stabilize the proximal joint (40). As a whole, these kinetic differences may indicate a lack of proximal strength and/or neuromuscular control restoration during a relatively low demand functional task in patients 6–23 months post ACLR. A study comparing lunge biomechanics between pre- and post-ACLR time points in a small, mixed ACLR patient group (without QT patients) found a 15% increase in internal hip extensor moment. This points to a compensatory strategy with higher proximal force contribution following reconstruction that contrasts with our results (18). These are different patients, graft types, and tasks but both studies depict initial investigations into a complex question around lower extremity biomechanical patterns following ACLR. Gluteal strength is a common emphasis within ACLR rehabilitation programs, and a recent meta-analysis found no differences in hip strength outcomes between the ACLR limb and nonsurgical limb or healthy control (41), contrary to evidence of sustained thigh muscle weakness. Therefore, additional research is needed to better understand the driving force of the interlimb kinetic differences at the hip and influence on distal joint biomechanics and function. External hip joint moment magnitude is influenced by the position of the center of mass, so another future direction may be to examine trunk positioning during gait to further elucidate this finding (42).

4.3 Sagittal plane

There was no significant difference between limbs for the hip sagittal plane angle, which is consistent with the limited literature (28). However, without a control group we cannot specify whether the surgical limb returned to normal motion or if the nonsurgical limb adapted to maintain symmetry with the ACLR limb. Slater et al. (39) reported a peak mean hip flexion angle of 25° in the healthy control group and just under that for the ACLR group nonsurgical limb while our participants' peak flexion angles were 18° and 19° for the nonsurgical and ACLR limbs, respectively. Further, the meta-analysis shows that time post-surgery may influence this movement pattern because participants at 9 and 11 months post-ACLR had significantly more hip flexion with 37° and 33°, respectively (39). The average time since surgery for our participants fell within this time frame but they displayed only 50%–60% of the peak hip flexion. This discrepancy may be due to differences in graft type as this was not specified in the meta-analysis results. To our knowledge, this is the first study describing gait biomechanics for patients with QT-ACLR so this population would not have been included in the earlier meta-analysis. Grafting the central portion of the QT may influence function of the rectus femoris as a hip flexor and future investigation should examine hip flexor strength and neuromuscular control following QT-ACLR.

4.4 Frontal and transverse plane

A pattern of reduced excursion in the ACLR limb was present in the knee adduction angle (Figure 1). We found 2° of excursion in the ACLR limb compared to 6° in the nonsurgical limb. This represents less dynamic frontal plane motion, with less adduction during the loading response and preswing stance phases and more adduction during the midstance and terminal stance phases of gait. Fewer studies have reported frontal plane knee angle kinematics during walking and of those limited studies, the results are mixed (11) with some showing no difference between limbs (28), while others report a decreased peak knee adduction angle (14) or an increased peak adduction angle when compared to a healthy control group (43, 44). Participants may be limiting frontal plane range of motion due to changes in postural control post-ACLR (45) in an effort to stabilize the joint by limiting variability. Staying within a smaller envelope of motion throughout the stance phase of gait decreases their adaptability to various conditions and perturbations (46). The reduction in variability within our participants can be visualized (Figure 1) by the narrower confidence interval (driven by a smaller standard deviation) for the ACLR limb compared to the nonsurgical limb.

Research is inconclusive on frontal plane loading following ACLR. One study found an increase in peak internal knee abduction moment (corresponding to an increased external knee adduction moment) in patients with ACLR compared to a matched healthy control (43). Another showed no difference for peak external knee adduction moment when comparing PT-ACLR and HT-ACLR male patients however both groups displayed smaller peak moments when compared to a control group (14). While a third reported a decreased knee abduction moment compared to healthy matched controls (47). Frontal plane loading is significant as it influences the distribution of forces between the medial and lateral tibiofemoral joint compartments (14, 43) and corresponds to changes in cartilage composition measures (48). More specifically, in healthy cartilage a larger knee adduction moment improves medial cartilage thickness whereas in those with osteoarthritis a higher adduction moment is associated with reduced cartilage thickness (49). These differences in frontal plane biomechanics for the ACLR limb indicate abnormal biomechanics and the potential for changes to cartilage loading and homeostasis (49). Literature suggests that frontal plane loading strategies shift with time post-surgery where during the initial postsurgical phase (<2 years) external adduction moments are less than the nonsurgical limb but a reversed pattern with longer recovery time (>5 years) (28). Underloading, an adaptation to impairments such as pain, swelling, decreased range of motion, and muscle weakness immediately following ACLR, creates a "learned nonuse" phenomenon (50) and has been shown to be a contributor of poor cartilage health and potentially osteoarthritis (48).

The transverse plane knee angle showed a large and consistent effect between limbs where the ACLR limb had less tibial external rotation compared to the nonsurgical limb throughout the entirety of stance. A similar pattern was described in other ACLR (36) and ACLD patients (51). The difference in external rotation may stem from a loss of external rotation during the swing phase due to an insufficient screw-home mechanism. The screw-home mechanism occurs when the tibiofemoral joint approaches terminal extension and the tibia externally rotates approximately 15 degrees during the last 20 degrees of extension. From initial contact through the loading response phase the knee flexes and begins to reverse the screw-home mechanism moving towards internal rotation (52). While we did not report on swing phase kinematics, our participants displayed a more flexed knee at initial contact suggesting they may have never reached terminal extension and achieved the maximum amount of external rotation that couples with end-range extension. This gait pattern may be an unconscious strategy to limit anterior tibial shear force that occurs with open kinetic chain knee extension (38). Future research should include an analysis of the full gait cycle to better understand how kinematics during the non-weight bearing swing phase influence limb and joint positioning during the stance phase. Overall, these differences in tibiofemoral kinematics may alter contact patterns and affect load distribution to the meniscus and cartilage within the joint (16, 51). A longitudinal study found the ACLR cohort had changes in tibial rotation kinematics at six months that corresponded to cartilage matrix changes at the one year follow up (53). Therefore, over time, and millions of steps, altered kinematics may be a significant contributor to the development of posttraumatic osteoarthritis.

The curve analysis revealed a pattern at the hip of decreased adduction/increased abduction and decreased external rotation/increased internal rotation for the surgical limb. The QT-ACLR limb showed a significant and consistent ~2.5° reduction in adduction throughout the entirety of stance. There was a ~1–2.5° difference in transverse plane hip motion however this wasn't significant except for the preswing phase, where there was a trend toward more internal rotation (Figure 2). Evidence reporting frontal plane hip angle is limited (28, 39, 43) and absent for transverse plane within ACLR populations. The early ACLR group in Goetschius et al. (28) had more hip adduction during the swing phase but no inter-limb differences during stance. Our analysis did not include the swing phase, but future research should be conducted to better understand how the limb is positioned prior to loading. Another study (43) reported a peak hip adduction angle of 8.8° for patients with ACLR and 9.2° for healthy controls along with a hip adduction excursion of 9°. This peak is twice as high compared to our ACLR limb finding but in line with the total frontal plane excursion of 9°. The average time since surgery for the previously studied group was 5.4 ± 4.4 years, representing a wide range. Previous studies have shown different kinematic patterns over time when examining ACLR populations from a cross-sectional approach (28, 39) and a change in gait biomechanics between 6 and 12 months (16). Therefore, hip adduction angle may shift and "normalize" over time (28) but maintain a consistent motion excursion within the frontal plane.

4.5 Vertical ground reaction force

We hypothesized differences in vGRF between the ACLR and nonsurgical limbs; however, we did not find any statistically significant differences. The lack of a true control group hinders our ability to fully interpret this finding and determine whether force absorption patterns were restored in patients following QT-ACLR or if the symmetry is driven by adaptation within the nonsurgical limb. A study examining walking biomechanics in patients with PT-ACLR at 6- and 12-months post-surgery showed differences in vGRF patterns for both the ACLR and nonsurgical limbs compared to healthy controls suggesting that even the nonsurgical limb accommodated over time (16). There is a complex relationship between external loads and internal forces (from muscles) that may produce similar vGRF patterns between limbs, as we observed in our study, and potentially correspond to altered articular cartilage loading patterns.

A study investigating the effect of walking speed on vGRF symmetry in ACLR individuals found more interlimb symmetry at a slower walking speed compared to faster speeds. However, this finding was not apparent for healthy controls (54). Specifically, these participants determined their self-selected pace overground and then performed the walking trials on a split-belt instrumented treadmill. Our participants determined their self-selected pace on the split-belt instrumented treadmill and this likely influenced their walking speed as it is considerably slower (0.82 ± .22 m/s) than other reported studies with ACLR patients (16, 33, 54, 55).

4.6 Limitations

We note several limitations including a convenience sample of only 14 participants 6-months to 2-years post-surgery with an unequal distribution between males and females resulting in heterogeneity for both participant age and months since surgery, creating potentially more variability in the findings. However, all participants had returned to unrestricted physical activity and performed their gait analysis within this time frame, similar to the "early ACLR" group as defined in the Goetchius et al. (28)

study allowing for comparison with their findings. Further, the case series study design did not include a control group limiting our ability to determine whether the nonsurgical limb compensated to maintain symmetry. Previous research appears to be mixed on whether there are biomechanical differences between the nonsurgical limb and a healthy control group, influenced by time since surgery (28). Emerging research points to limb dominance potentially influencing joint loading in ACL-R populations, but this was not accounted for in our analysis (56).

Participants had a slow average self-selected walking speed. Gait biomechanics are speed-dependent where lower walking speeds may result in reduced kinetics and joint ROM excursion. The lower self-selected walking speed for our participants may have affected the magnitude and symmetry of forces generated during the stance phase of the gait cycle and comparability across populations. We examined vGRF as a measure of limb loading but it does not fully represent joint loading that is also influenced by muscle forces and co-contraction that are not captured by external forces. Future study could include musculoskeletal modeling to better estimate joint contact forces. We didn't capture trunk kinematics which can influence distal joint moments (57) and this is an area for future researchers to examine as well as implementing statistical parametric mapping for a more comprehensive analysis of time-series data. However, the results remain valuable as they represent the function of a specific, novel research population.

4.7 Conclusion

This study is the first investigation into QT-ACLR walking biomechanics. Three-dimensional kinematic and kinetic gait alterations are present at the knee and hip in patients with QT-ACLR in the ACLR limb when compared to the nonsurgical limb, a pattern consistent with other ACLR patient graft types. Most of these differences occur in the periods of gait associated with higher forces, despite not finding significant differences in vGRF between limbs. In comparison to patient reported outcome measures, data describing gait biomechanics following QT-ACLR are extremely limited. Future studies should aim to test larger and more homogenous populations to better assess walking gait at specific recovery time points and examine multiple joints within the kinetic chain to evaluate for potential compensatory patterns. Researchers should work to compare gait biomechanics between patients with QT-ACLR and a true control group, knowing that nonsurgical limbs may also demonstrate biomechanical deficits following injury and ACLR. In line with ACLR rehabilitation protocols, the results of this study support emphasizing restoration of full knee extension, neuromuscular activation and strengthening of the quadriceps and proximal hip musculature to improve knee joint biomechanics and functional movement patterns.

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 humans were approved by The Institutional Review Board for Human Research at the Medical University of South Carolina. The studies were conducted in accordance with the local legislation and institutional requirements. Written informed consent for participation in this study was provided by the participants' legal guardians/next of kin.

Author contributions

KP: Conceptualization, Formal analysis, Visualization, Writing – original draft, Writing – review & editing. BP: Data curation, Formal analysis, Visualization, Writing – original draft. HS: Conceptualization, Methodology, Resources, Writing – review & editing. MM: Conceptualization, Methodology, Writing – review & editing. CG: Conceptualization, Methodology, Resources, Supervision, Writing – review & editing. JH: Conceptualization, Data curation, Investigation, Methodology, Project administration, Writing – review & editing.

Funding

The author(s) declare that no financial support was received for the research and/or publication of this article.

Conflict of interest

JH was employed by Hunnicutt Writing and Consulting, LLC.

The remaining 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.

Generative AI statement

The author(s) declare that no Generative AI was used in the creation of this manuscript.

Publisher's note

References

1. Chia L, De Oliveira Silva D, Whalan M, McKay MJ, Sullivan J, Fuller CW, et al. Non-contact anterior cruciate ligament injury epidemiology in team-ball sports: a systematic review with meta-analysis by sex, age, sport, participation level, and exposure type. *Sports Med.* (2022) 52(10):2447–67. doi: 10.1007/s40279-022-01697-w

2. Joseph AM, Collins CL, Henke NM, Yard EE, Fields SK, Comstock RD. A multisport epidemiologic comparison of anterior cruciate ligament injuries in high school athletics. *J Athl Train.* (2013) 48(6):810–7. doi: 10.4085/1062-6050-48.6.03

3. Herzog MM, Marshall SW, Lund JL, Pate V, Spang JT. Cost of outpatient arthroscopic anterior cruciate ligament reconstruction among commercially insured patients in the United States, 2005–2013. *Orthop J Sports Med.* (2017) 5(1):2325967116684776. doi: 10.1177/2325967116684776

4. Ross AG, Agresta B, McKay M, Pappas E, Cheng T, Peek K. Financial burden of anterior cruciate ligament reconstructions in football (soccer) players: an Australian cost of injury study. *Inj Prev.* (2023) 29(6):474–81. doi: 10.1136/ip-2023-044885

5. Ajrawat P, Dwyer T, Whelan D, Theodoropoulos J, Murnaghan L, Bhargava M, et al. A comparison of quadriceps tendon autograft with bone-patellar tendon-bone autograft and hamstring tendon autograft for primary anterior cruciate ligament reconstruction: a systematic review and quantitative synthesis. *Clin J Sport Med.* (2021) 31(4):392–9. doi: 10.1097/JSM.0000000000000765

6. Migliorini F, Torsiello E, Trivellas A, Eschweiler J, Hildebrand F, Maffulli N. Bone-patellar tendon-bone versus two- and four-strand hamstring tendon autografts for ACL reconstruction in young adults: a Bayesian network meta-analysis. *Sci Rep.* (2023) 13(1):6883. doi: 10.1038/s41598-023-33899-1

7. Marshall JL, Warren RF, Wickiewicz TL, Reider B. The anterior cruciate ligament: a technique of repair and reconstruction. *Clin Orthop.* (1979) 143:97–106.

8. Arnold MP, Calcei JG, Vogel N, Magnussen RA, Clatworthy M, Spalding T, et al. ACL study group survey reveals the evolution of anterior cruciate ligament reconstruction graft choice over the past three decades. *Knee Surg Sports Traumatol Arthrosc.* (2021) 29(11):3871–6. doi: 10.1007/s00167-021-06443-9

9. Lee IM, Buchner DM. The importance of walking to public health. *Med Sci Sports Exerc.* (2008) 40(7):S512–8. doi: 10.1249/MSS.0b013e31817c65d0

10. Weber D. Differences in physical aging measured by walking speed: evidence from the English longitudinal study of ageing. *BMC Geriatr.* (2016) 16(1):31. doi: 10.1186/s12877-016-0201-x

11. Kaur M, Ribeiro DC, Theis JC, Webster KE, Sole G. Movement patterns of the knee during gait following ACL reconstruction: a systematic review and meta-analysis. *Sports Med.* (2016) 46(12):1869–95. doi: 10.1007/s40279-016-0510-4

12. Erhart-Hledik JC, Chu CR, Asay JL, Andriacchi TP. Longitudinal changes in knee gait mechanics between 2 and 8 years after anterior cruciate ligament reconstruction. *J Orthop Res.* (2018) 36(5):1478–86. doi: 10.1002/jor.23770

13. Levinger P, Webster KE, Feller J. Asymmetric knee loading at heel contact during walking in patients with unilateral knee replacement. *Knee.* (2008) 15(6):456–60. doi: 10.1016/j.knee.2008.06.002

14. Webster KE, Feller JA, Wittwer JE. Longitudinal changes in knee joint biomechanics during level walking following anterior cruciate ligament reconstruction surgery. *Gait Posture.* (2012) 36(2):167–71. doi: 10.1016/j.gaitpost.2012.02.004

15. Wellsandt E, Khandha A, Capin J, Buchanan TS, Snyder-Mackler L. Operative and nonoperative management of anterior cruciate ligament injury: differences in gait biomechanics at 5 years. *J Orthop Res.* (2020) 38(12):2675–84. doi: 10.1002/jor.24652

16. Davis-Wilson HC, Pfeiffer SJ, Johnston CD, Seeley MK, Harkey MS, Blackburn JT, et al. Bilateral gait 6 and 12 months post–anterior cruciate ligament reconstruction compared with controls. *Med Sci Sports Exerc.* (2020) 52(4):785–94. doi: 10.1249/MSS.0000000000002208

17. Pietrosimone B, Seeley MK, Johnston C, Pfeiffer SJ, Spang JT, Blackburn JT. Walking ground reaction force post-ACL reconstruction: analysis of time and symptoms. *Med Sci Sports Exerc.* (2019) 51(2):246–54. doi: 10.1249/MSS.0000000000001776

18. Stenroth L, Bartholdy C, Schwarz Larsen J, Sørensen MS, Smale KB, Flaxman TE, et al. Altered movement strategy during functional movement after an ACL injury, despite ACL reconstruction. *Front Sports Act Living.* (2022) 4:994139. doi: 10.3389/fspor.2022.994139

19. Perry J, Burnfield JM. *Gait Analysis: Normal and Pathological Function. 2nd ed.* Thorofare, NJ: SLACK (2010). p. 551.

20. Bjornsen E, Schwartz TA, Lisee C, Blackburn T, Lalush D, Nissman D, et al. Loading during midstance of gait is associated with magnetic resonance imaging of cartilage composition following anterior cruciate ligament reconstruction. *Cartilage.* (2022) 13(1):194760352110722. doi: 10.1177/19476035211072220

21. O'Connell D, Golightly Y, Lisee C, Pietrosimone B. Interlimb differences in T1ρ MRI relaxation times linked with symptomatic knee osteoarthritis following anterior cruciate ligament reconstruction. *Knee.* (2023) 41:353–9. doi: 10.1016/j.knee.2023.02.006

22. The MOON Group, Everhart JS, Jones MH, Yalcin S, Reinke EK, Huston LJ, et al. The clinical radiographic incidence of posttraumatic osteoarthritis 10 years after anterior cruciate ligament reconstruction: data from the MOON nested cohort. *Am J Sports Med.* (2021) 49(5):1251–61. doi: 10.1177/0363546521995182

23. Duhamel A, Bourriez JL, Devos P, Krystkowiak P, Destée A, Derambure P, et al. Statistical tools for clinical gait analysis. *Gait Posture.* (2004) 20(2):204–12. doi: 10.1016/j.gaitpost.2003.09.010

24. Park J, Seeley MK, Francom D, Reese CS, Hopkins JT. Functional vs. traditional analysis in biomechanical gait data: an alternative statistical approach. *J Hum Kinet.* (2017) 60:39–49. doi: 10.1515/hukin-2017-0114

25. Slone HS, Ashford WB, Xerogeanes JW. Minimally invasive quadriceps tendon harvest and graft preparation for all-inside anterior cruciate ligament reconstruction. *Arthrosc Tech.* (2016) 5(5):e1049–56. doi: 10.1016/j.eats.2016.05.012

26. Houglum PA, Bertoti D, Brunnstrom S. *Brunnstrom's Clinical Kinesiology. 6th ed.* Philadelphia: F.A. Davis (2012). p. 704.

27. Sullivan GM, Feinn R. Using effect size-or why the *P* value is not enough. *J Grad Med Educ.* (2012) 4(3):279–82. doi: 10.4300/JGME-D-12-00156.1

28. Goetschius J, Hertel J, Saliba SA, Brockmeier SF, Hart JM. Gait biomechanics in anterior cruciate ligament–reconstructed knees at different time frames postsurgery. *Med Sci Sports Exerc.* (2018) 50(11):2209–16. doi: 10.1249/MSS.0000000000001693

29. Di Stasi SL, Snyder-Mackler L. The effects of neuromuscular training on the gait patterns of ACL-deficient men and women. *Clin Biomech.* (2012) 27(4):360–5. doi: 10.1016/j.clinbiomech.2011.10.008

30. Pietrosimone B, Lepley AS, Kuenze C, Harkey MS, Hart JM, Blackburn JT, et al. Arthrogenic muscle inhibition following anterior cruciate ligament injury. *J Sport Rehabil.* (2022) 31(6):694–706. doi: 10.1123/jsr.2021-0128

31. Hunnicutt JL, Gregory CM, McLeod MM, Woolf SK, Chapin RW, Slone HS. Quadriceps recovery after anterior cruciate ligament reconstruction with quadriceps tendon versus patellar tendon autografts. *Orthop J Sports Med.* (2019) 7(4):232596711983978. doi: 10.1177/2325967119839786

32. Johnston PT, Feller JA, McClelland JA, Webster KE. Strength deficits and flexion range of motion following primary anterior cruciate ligament reconstruction differ between quadriceps and hamstring autografts. *J ISAKOS Jt Disord Orthop Sports Med.* (2021) 6(2):88–93. doi: 10.1136/jisakos-2020-000481

33. Pietrosimone B, Davis-Wilson HC, Seeley MK, Johnston C, Spang JT, Creighton RA, et al. Gait biomechanics in individuals meeting sufficient quadriceps strength cutoffs after anterior cruciate ligament reconstruction. *J Athl Train.* (2021) 56(9):960–6. doi: 10.4085/425-20

34. Arhos EK, Capin JJ, Buchanan TS, Snyder-Mackler L. Quadriceps strength symmetry does not modify gait mechanics after anterior cruciate ligament reconstruction, rehabilitation, and return-to-sport training. *Am J Sports Med.* (2021) 49(2):417–25. doi: 10.1177/0363546520980079

35. Johnston PT, McClelland JA, Feller JA, Webster KE. Knee muscle strength after quadriceps tendon autograft anterior cruciate ligament reconstruction: systematic review and meta-analysis. *Knee Surg Sports Traumatol Arthrosc.* (2021) 29(9):2918–33. doi: 10.1007/s00167-020-06311-y

36. Gao B, Zheng NN. Alterations in three-dimensional joint kinematics of anterior cruciate ligament-deficient and -reconstructed knees during walking. *Clin Biomech.* (2010) 25(3):222–9. doi: 10.1016/j.clinbiomech.2009.11.006

37. Holder J, Trinler U, Meurer A, Stief F. A systematic review of the associations between inverse dynamics and musculoskeletal modeling to investigate joint loading in a clinical environment. *Front Bioeng Biotechnol.* (2020) 8:603907. doi: 10.3389/fbioe.2020.603907

38. Beaulieu ML, Ashton-Miller JA, Wojtys EM. Loading mechanisms of the anterior cruciate ligament. *Sports Biomech.* (2023) 22(1):1–29. doi: 10.1080/14763141.2021.1916578

39. Slater LV, Hart JM, Kelly AR, Kuenze CM. Progressive changes in walking kinematics and kinetics after anterior cruciate ligament injury and reconstruction: a review and meta-analysis. *J Athl Train.* (2017) 52(9):847–60. doi: 10.4085/1062-6050-52.6.06

40. Liao TC, Samaan MA, Popovic T, Neumann J, Zhang AL, Link TM, et al. Abnormal joint loading during gait in persons with hip osteoarthritis is associated with symptoms and cartilage lesions. *J Orthop Sports Phys Ther.* (2019) 49(12):917–24. doi: 10.2519/jospt.2019.8945

41. Girdwood M, Culvenor AG, Patterson B, Haberfield M, Rio EK, Hedger M, et al. No sign of weakness: a systematic review and meta-analysis of hip and calf muscle strength after anterior cruciate ligament injury. *Br J Sports Med.* (2024) 58(9):500–10. doi: 10.1136/bjsports-2023-107536

42. Diamond LE, Allison K, Dobson F, Hall M. Hip joint moments during walking in people with hip osteoarthritis: a systematic review and meta-analysis. *Osteoarthritis Cartilage.* (2018) 26(11):1415–24. doi: 10.1016/j.joca.2018.03.011

43. Butler RJ, Minick KI, Ferber R, Underwood F. Gait mechanics after ACL reconstruction: implications for the early onset of knee osteoarthritis. *Br J Sports Med.* (2009) 43(5):366–70. doi: 10.1136/bjsm.2008.052522

44. Georgoulis AD, Papadonikolakis A, Papageorgiou CD, Mitsou A, Stergiou N. Three-dimensional tibiofemoral kinematics of the anterior cruciate ligament-deficient and reconstructed knee during walking. *Am J Sports Med.* (2003) 31(1):75–9. doi: 10.1177/03635465030310012401

45. Monfort SM, Simon JE, Miko SC, Grooms DR. Effects of cognitive- and motor-dual tasks on postural control regularity following anterior cruciate ligament reconstruction. *Gait Posture*. (2022) 97:109–14. doi: 10.1016/j.gaitpost.2022.07.246

46. Decker LM, Moraiti C, Stergiou N, Georgoulis AD. New insights into anterior cruciate ligament deficiency and reconstruction through the assessment of knee kinematic variability in terms of nonlinear dynamics. *Knee Surg Sports Traumatol Arthrosc*. (2011) 19(10):1620–33. doi: 10.1007/s00167-011-1484-2

47. Patterson MR, Delahunt E, Caulfield B. Peak knee adduction moment during gait in anterior cruciate ligament reconstructed females. *Clin Biomech*. (2014) 29(2):138–42. doi: 10.1016/j.clinbiomech.2013.11.021

48. Evans-Pickett A, Lisee C, Horton WZ, Lalush D, Nissman D, Blackburn JT, et al. Worse tibiofemoral cartilage composition is associated with insufficient gait kinetics after ACL reconstruction. *Med Sci Sports Exerc*. (2022) 54(10):1771–81. doi: 10.1249/MSS.0000000000002969

49. Andriacchi TP, Koo S, Scanlan SF. Gait mechanics influence healthy cartilage morphology and osteoarthritis of the knee. *J Bone Jt Surg*. (2009) 91(Supplement_1):95–101. doi: 10.2106/JBJS.H.01408

50. Chan M, Sigward SM. Individuals following anterior cruciate ligament reconstruction practice underloading strategies during daily activity. *J Orthop Res*. (2022) 40(3):565–72. doi: 10.1002/jor.25070

51. Andriacchi TP, Dyrby CO. Interactions between kinematics and loading during walking for the normal and ACL deficient knee. *J Biomech*. (2005) 38(2):293–8. doi: 10.1016/j.jbiomech.2004.02.010

52. Kim HY, Kim KJ, Yang DS, Jeung SW, Choi HG, Choy WS. Screw-home movement of the tibiofemoral joint during normal gait: three-dimensional analysis. *Clin Orthop Surg*. (2015) 7(3):303. doi: 10.4055/cios.2015.7.3.303

53. Amano K, Pedoia V, Su F, Souza RB, Li X, Ma CB. Persistent biomechanical alterations after ACL reconstruction are associated with early cartilage matrix changes detected by quantitative MR. *Orthop J Sports Med*. (2016) 4(4):2325967116644421. doi: 10.1177/2325967116644421

54. Garcia SA, Brown SR, Koje M, Krishnan C, Palmieri-Smith RM. Gait asymmetries are exacerbated at faster walking speeds in individuals with acute anterior cruciate ligament reconstruction. *J Orthop Res*. (2022) 40(1):219–30. doi: 10.1002/jor.25117

55. Garcia SA, Johnson AK, Brown SR, Washabaugh EP, Krishnan C, Palmieri-Smith RM. Dynamic knee stiffness during walking is increased in individuals with anterior cruciate ligament reconstruction. *J Biomech*. (2023) 146:111400. doi: 10.1016/j.jbiomech.2022.111400

56. Goto S, Garrison JC, Singleton SB, Dietrich LN, Hannon JP. Effects of limb dominance on patellofemoral joint loading during gait at 12 weeks after anterior cruciate ligament reconstruction. *Orthop J Sports Med*. (2022) 10(7):23259671221088316. doi: 10.1177/23259671221088316

57. Byrnes SK, Holder J, Stief F, Wearing S, Böhm H, Dussa CU, et al. Frontal plane knee moment in clinical gait analysis: a systematic review on the effect of kinematic gait changes. *Gait Posture*. (2022) 98:39–48. doi: 10.1016/j.gaitpost.2022.07.258