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Department of Rehabilitation Sciences

Musculoskeletal Rehabilitation Research Group

From ankle sprain to chronic ankle instability:

**Decoding pathomechanical
foot joint loading patterns during sporting tasks
and intrinsic foot muscle properties and functions**

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Supervisors: Prof. Christine Detrembleur,
Prof. Kevin Deschamps and Dr Sébastien Lobet

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‘Ankle after a sprain’: Alexis Berg generously granted his permission to use this picture.

‘Athlete performing a change of direction’ and ‘Foot muscle ultrasound’: GPT 5.2.

‘Forest plot’ and ‘Reflective marker’: J.-L. Peters-Dickie.

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« [...] »

À l'enfant qui viendra

Qui nous ressemblera

Qui sera à la fois toi et moi »

Joe Dassin, « à toi » (1976).

Context of this PhD project

Environment

The Neuromusculoskeletal laboratory (NSMK) of the UCLouvain aims to conduct studies among individuals affected with various neurological and musculoskeletal disorders. This broad range of research topics, together with the complementary backgrounds of researchers affiliated at NMSK laboratory, encourages multidisciplinary approaches to investigate tissular integrity, altered function, and restrictions in participation. Prof. Christine Detrembleur, head of NMSK laboratory from 2015 to 2022, has been an indispensable figure in the development of the laboratory, supporting countless doctoral researchers, as a dedicated supervisor or collaborator.

The clinical motion analysis laboratory (CMAL) of the university hospital Leuven lies a few kilometers from Katholieke Universiteit Leuven. Despite its calm environment in the middle of fields and meadows for some sheeps, CMAL is a place of cutting-edge biomechanical research among children with cerebral palsy and adults with foot-related disorders. Prof. Kevin Deschamps has been operating the advanced gait analysis facilities for more than a decade. His various studies aimed to better understand the development pathways of ankle and foot-related disorders, improve medical and paramedical care, and ultimately promote health status. A few years ago, Dr Sébastien Lobet completed his PhD at UCLouvain and subsequently carried out postdoctoral research at CMAL, focusing on the characterization of ankle and foot biomechanics in individuals with haemophilia.

Timeline

Since graduating with my first Master's degree in physiotherapy and rehabilitation sciences at UCLouvain, I have felt a strong intuition to pursue a PhD. It offers a unique opportunity to keep learning while contributing new knowledge through research. This journey also opens the door to unexpected collaborations.

To meet the requirements for admission to the PhD programme, I completed a second Master's degree in Physiotherapy. I then approached Prof. Detrembleur,

bringing little more than my motivation and the desire to embark on a doctoral journey. This PhD project was born from the emerging connection between Prof. Detrembleur and Prof. Deschamps, who were probably guided by the feeling that their complementary profiles would result in a fruitful and pleasant collaboration. Dr Lobet immediately accepted to join the adventure, offering his expertise in multi-segment foot models and guidance on writing grant proposals.

During my first year of PhD, I became familiar with multi-segment foot models by assisting with gait analyses conducted by Prof. Deschamps in CMAL. I also wrote the protocol of the studies that would be conducted in that laboratory. During my second and third years of PhD, the recruitment of CMAL study participants progressed and we conducted an ultrasound reliability study with Nicolas Haelewyn (KU Leuven, Bruges), Valentien Spanhove (UGent), and Roel de Ridder (UGent). For a few months, there was a failure with CMAL's equipment, preventing data acquisition. During that time, I conducted a study comparing two protocols to generate foot muscle fatigue at NMSK laboratory, collaborating with Dr Anh Phong Nguyen and Dr Virginie Otlet. We also collaborated with Marina Guallar-Bouloc (Spain) and Dr Benjamin Hidalgo (UCLouvain) to conduct a meta-analysis on foot muscle strengthening. The last year of PhD offered me the opportunity to present my work at two international conferences and to step up the pace of article writing.

This PhD has also given me the opportunity to collaborate with inspiring researchers on topics that are particularly meaningful to me. At the outset of these four years, I contributed to Nicolas' systematic review on ultrasound imaging of the foot muscles. Later, together with Dr Anh Phong Nguyen and Thomas Hansoulle, we conducted a meta-analysis identifying ankle sprain as a common injury among runners. More recently, Anh Phong involved me in a study assessing running ground reaction forces across various clinical categories related to chronic ankle instability. In addition, I had the chance to collaborate with several researchers in supervising Master's theses, some of which may lead to future publications.

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Summary

Chronic ankle instability (CAI) is a frequent consequence of lateral ankle sprains, characterized by persistent symptoms and functional deficits, whereas *copers* recover with minimal impairments. Altered movement patterns, muscle strength, and muscle sizes are thought to contribute to this continuum, yet ankle and foot joint biomechanics during change-of-direction tasks and the role of intrinsic foot muscles (IFMs) remain poorly understood.

This PhD aimed to compare ankle and foot joint dynamics among individuals with CAI, copers, and healthy controls during side-step cutting, and to investigate IFM sizes, strength, and fatigue-related effects. Methodologically, it evaluated IFM strengthening interventions, the reliability of low-end ultrasound imaging, and different foot muscle fatigue protocols.

Meta-analyses showed that IFM exercises and minimalist footwear may induce trivial to moderate increases in IFM size and foot strength, albeit with very low certainty of evidence. Exercises reduced medial longitudinal foot mobility, whereas minimalist footwear promoted a more forefoot-oriented running pattern. Experimentally, concentric toe flexor contractions induced twenty percent hallux flexion strength decline, indicating fatigue, whereas isometric contractions combined with electrical stimulation did not. Both interventions acutely modified running ground reaction forces. Ultrasound imaging with a low-end device demonstrated moderate to good test-retest and inter-rater reliability when measuring identical images, but poor reliability when image acquisition was included.

During anticipated 45° side-step cutting, foot joint kinematics were largely similar across groups, although individuals with CAI showed trends toward altered ankle, midfoot, and hallux kinetics, as well as reduced posterior ground reaction forces. IFM sizes did not differ between groups, but hallux flexion strength was lower in CAI and copers compared to controls. Following isometric contractions plus electrical stimulations, individuals with CAI exhibited a moderate strength decline and changes in ankle and foot joint dynamics.

Overall, this PhD highlights the multifactorial and heterogeneous nature of CAI, in which subtle kinetic, neuromuscular, and sensory-perceptual alterations coexist. These findings support investigating CAI using task-specific, integrative, and longitudinal approaches that account for symptom severity, functional subgroups, and neurocognitive demands. Advancing foot assessment methods, refining conceptual models, and evaluating targeted treatment modalities are essential to better understand the mechanisms underlying CAI and to optimize individualized rehabilitation.

Résumé

L'instabilité chronique de la cheville (ICC) est une conséquence fréquente des entorses latérales de la cheville, caractérisée par des symptômes persistants et des déficits fonctionnels, tandis que les *copers* récupèrent leur fonction. Ce continuum clinique semble influencé par des altérations des mouvements, de la force et de la taille des muscles. En revanche, trop peu d'études ont évalué la biomécanique du pied lors de changements de direction, ainsi que la fonction des muscles intrinsèques du pied (MIP).

Cette thèse avait pour objectif principal de comparer la dynamique articulaire de la cheville et du pied chez des personnes avec ICC, des *copers* et des sujets contrôles sains lors de changements de direction, ainsi que d'examiner la taille, la force et les effets liés à la fatigue des MIP. Sur le plan méthodologique, elle a évalué l'efficacité du renforcement des MIP, la fiabilité d'appareils d'échographie d'entrée de gamme, ainsi que différents protocoles de fatigue.

Les méta-analyses ont montré que les exercices ciblant les MIP et le port de chaussures minimalistes pouvaient induire des augmentations triviales à modérées de la taille des MIP et de la force du pied, avec un niveau de certitude des preuves très faible. Les exercices ont réduit la mobilité de l'arche longitudinale médiale du pied, tandis que les chaussures minimalistes ont favorisé une pose du pied davantage vers l'avant-pied lors de la course. Sur le plan expérimental, les contractions concentriques des fléchisseurs des orteils ont réduit de 20% la force de flexion de l'hallux, contrairement aux contractions isométriques associées à de l'électrostimulation. Les deux interventions ont modifié de manière aiguë les forces de réaction du sol lors de la course. L'échographie a montré une fiabilité modérée à bonne en test-retest et inter-évaluateurs lors de la mesure d'images identiques, mais une fiabilité faible lorsque l'acquisition des images était incluse.

Lors de changements de direction anticipés à 45°, la cinématique articulaire du pied étaient globalement similaires entre les groupes. En revanche, les individus avec ICC ont montré des tendances vers des altérations de la cinétiques de la cheville, du médio-pied et de l'hallux, ainsi qu'une réduction de la composante postérieure des forces de réaction du sol. La taille des MIP ne différait pas entre les groupes, mais la force de flexion de l'hallux était réduite chez les individus

présentant une ICC et chez les *copers* par rapport aux sujets contrôles. À la suite de contractions isométriques plus électrostimulation, les individus avec ICC ont présenté une diminution modérée de la force ainsi que des modifications de la dynamique articulaire de la cheville et du pied.

Dans l'ensemble, cette thèse met en évidence la nature multifactorielle et hétérogène de l'ICC, dans laquelle coexistent des altérations subtiles d'ordre cinétique, neuromusculaire, sensorielles et perceptuelles. Ces résultats soutiennent l'intérêt d'étudier l'ICC à l'aide d'approches intégratives, tâche-spécifiques et longitudinales, tenant compte de la sévérité des symptômes, de sous-groupes fonctionnels et des exigences neurocognitives. Le développement d'outils d'évaluation du pied, l'affinement des modèles conceptuels et l'évaluation de modalités thérapeutiques sont essentiels pour mieux comprendre les mécanismes sous-jacents de l'ICC et individualiser la rééducation.

Douter de tout et tout croire sont deux solutions également commodes qui l'une et l'autre nous dispensent de réfléchir.

Henri Poincaré (1854 – 1912)

General introduction

I. The ankle and foot complex

Phylogenetic evolution

Bipedalism has emerged 6-7 million years ago in early biped apes.¹ Their ankle and foot complex was characterized by an abducted and opposable hallux, long and curves toes, important midfoot flexibility, a flat or low longitudinal arch, a small calcaneus with limited joint contact area, and a short Achilles tendon. These features allowed efficient grasping and tree climbing, but limited efficiency for terrestrial locomotion.¹⁻⁴ In the million years that followed, the foot evolution was a gradual and mosaic process, in which different anatomical components evolved at varying rates in response to shifting selective pressures associated first with terrestrial bipedal walking, then with residual climbing behaviours, and eventually with the demands of long-distance endurance running (Figure 1).³

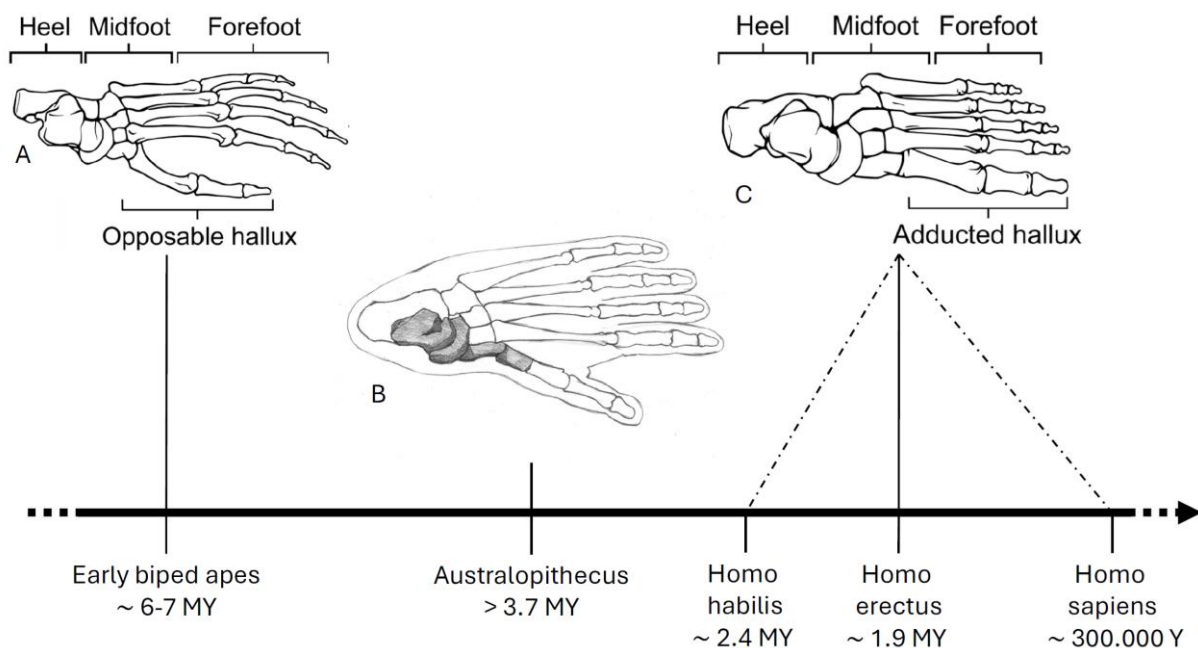


Figure 1. Evolution of foot from biped apes (A)³ to *Australopithecus afarensis* (B),⁵ followed by genus *Homo* (C),³ with some traits appearing in *Homo habilis*, most traits appearing in *Homo erectus* and latest features appearing in *Homo sapiens*.

Australopithecus adopted a mechanically efficient walking gait pattern at least 3.7 million years ago.⁶ This pre-human specie was also able to run, but running speed was considerably lower, and running cost was considerably higher than in early and modern humans.¹ *Australopithecus* – as for example the well-known “Lucy” – lacked some adaptations for a more efficient running, that emerged in *Homo habilis* (~ 2.4 up to 1.5 million years ago), and expanded in *Homo erectus* (~ 1.9 million years ago).¹

Homo habilis benefited from a permanently adducted hallux and shorter toes than *Australopithecus*, ensuring stability during ankle plantarflexion, along with a closed-packed calcaneocuboid joint, allowing energy storage during bouts of running. Additional adaptations fixated in *Homo erectus* include a longer Achilles tendon, better definition of foot arches, increased gluteus maximus volume and strength, shorter triceps surae muscle fibres and more power capacity from that muscle, longer lower limbs, and reduced upper body mass. A shorter Achilles tendon and higher foot arches ensure energy storage and release during running, longer lower limbs allow longer steps, while reduced upper body mass lowers the moment of inertia and muscular effort in running. Other adaptations affecting the pelvis, lower back, thorax, forearms, shoulders, neck, and head also emerged in *Homo habilis* or *Homo erectus*. During running, these traits reduce musculoskeletal stress, improve trunk and head stabilisation, enable dissociation of trunk and pelvis rotations, and help thermoregulation. *Homo erectus* became a good endurance runner, although he was – just as modern *Homo sapiens* – a poor sprinter. Endurance ability may have played a major role in human evolution, as part of hunting, resource acquisition, and social interactions.¹⁻⁴

Ankle-foot complex in modern humans

Introducing the foot core paradigm

The modern human ankle and foot complex (Figure 2) is shaped by its 28 bones, dozens of ligaments and the plantar fascia,⁷ that arrange in a considerable variety of postures and proportions across humans.⁸ This anatomy forms the medial longitudinal, lateral longitudinal and midtarsal transverse arches of the foot.⁹ Intrinsic and extrinsic foot muscles contribute to maintain this structural integrity and allow active movements, under control of the central nervous system, by regulating foot and arch deformation to ensure functional stability while preserving mobility during posture and locomotion.¹⁰ The central nervous system continuously receives information about ankle and foot deformations and movements from mechanoreceptors located in the skin, ligaments, tendons, joints, and muscles.¹¹

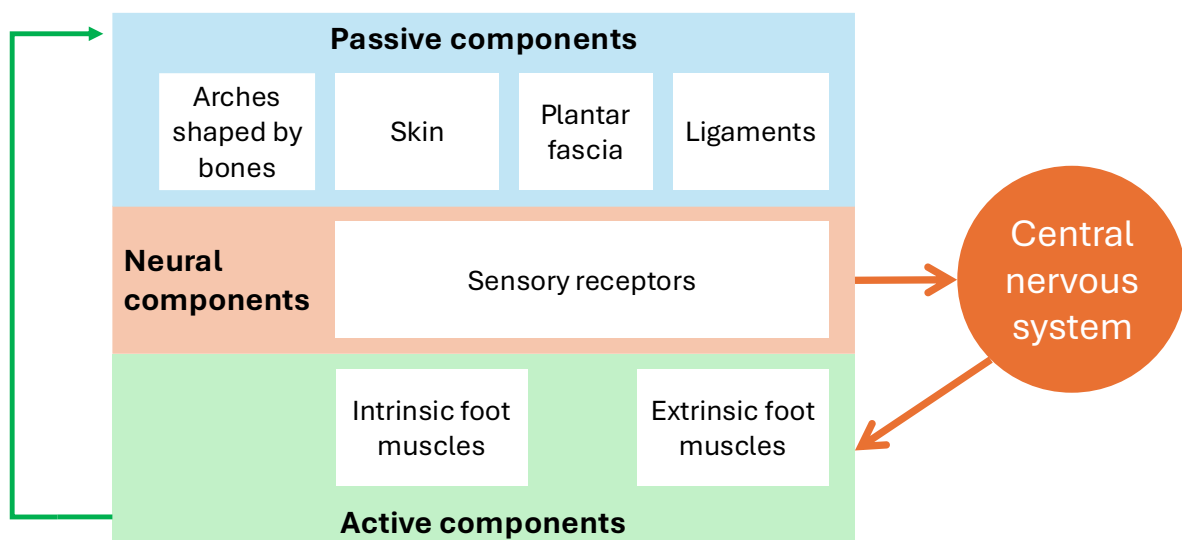


Figure 2. The foot and its connection with the central nervous system. The ankle and foot complex can be divided into three complementary systems. The passive components provide structural integrity, actively regulated by the active components (green arrow). This is done under the control of the central nervous system, that is informed by afferences from the neural components (orange arrows). Neural receptors are located within the passive and active components.

The intrinsic foot muscles (IFMs) have their origins and insertions within the foot (Figure 3),¹² while the extrinsic foot muscles originate outside the foot and present as tendons at the foot level. The superficial layer of IFMs contains the abductor hallucis, flexor digitorum brevis and abductor digiti minimi. Quadratus plantae is located deeper under the flexor digitorum brevis and inserts on the tendon of the flexor digitorum longus. The third layer includes flexor hallucis brevis and the two heads of adductor hallucis. Flexor digiti minimi brevis, lumbricals, interossei, extensor hallucis brevis, and extensor digitorum brevis are also part of the IFMs.¹²⁻¹³

Extrinsic foot muscles can be grouped in three compartments. Laterally, fibularis brevis and longus perform ankle eversion and plantarflexion. The anterior compartment includes muscles mainly involved in ankle dorsiflexion and toe extension. Muscles of the posterior compartment include triceps surae (composed of soleus and the two heads of gastrocnemius muscle) that attaches to the calcaneus through Achilles tendon, tibialis posterior that has superficial insertions on bones of the medial longitudinal arch, flexor hallucis longus and flexor digitorum longus.⁷

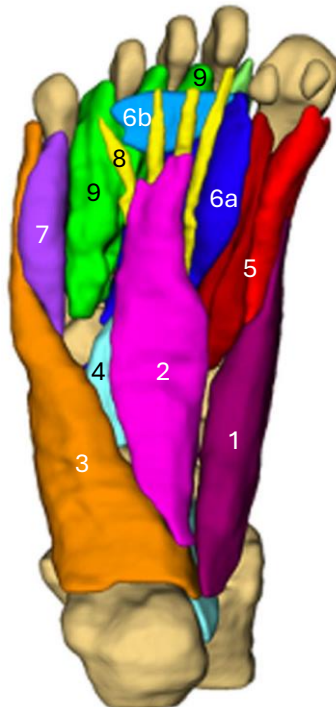


Figure 3. Intrinsic foot muscles. (1) abductor hallucis, (2) flexor digitorum brevis, (3) abductor digiti minimi, (4) quadratus plantae, (5) flexor hallucis brevis (medial and lateral heads), (6) adductor hallucis oblique (a) and transverse (b) heads, (7) flexor digiti minimi brevis, (8) lumbricals, and (9) interossei. Extensor hallucis brevis, and extensor digitorum brevis are located at the dorsum of the foot (not shown). Figure adapted from Smith et al. (2021).¹³

Functions of the ankle and foot complex

The ankle and foot complex contributes to postural control, adapts to constraints, and absorbs and generates energy during locomotion,³ under the control of the central nervous system.⁹ Walking and running are continuous transfers of energy: potential energy results from height of body parts, tissue elasticity (elastic energy), and molecular arrangements (chemical energy used at muscles). Kinetic energy is related to the velocity of body parts. Heat results from energy ‘losses’ during energetic exchanges. The human foot exhibits important joint motions during walking or running, especially at talocrural and midfoot joints.^{14 15}

From foot strike to mid-stance during running, the foot converts kinetic energy into elastic energy, which is subsequently returned in late stance for propulsion.¹⁶ Elastic energy is mostly stored in the plantar fascia,¹⁷ which keeps a relatively constant length in early-stance to mid-stance thanks to metatarsophalangeal plantarflexion (i.e. ‘reverse windlass mechanism’). The plantar fascia then lengthens around mid-stance, once the metatarsophalangeal joints have finished their course.¹⁸ With their short muscle fibre length and long tendons relative to muscle size, the IFMs contribute in a small extent to this energy storage,¹⁹⁻²⁰ through isometric contraction.²¹

During later periods of the stance phase of walking and running, the plantar fascia recoils, meaning elastic energy is released and its length decreases to assist propulsion.¹⁷⁻¹⁹ During that period, the metatarsophalangeal joints dorsiflex, delaying the release of the energy stored in the plantar fascia until very late stance,²² and elastic energy stored in the IFM tendons is also released.²¹ The IFMs help stiffen the metatarsophalangeal joints through active concentric or isometric contraction, aiding push-off.¹⁹ The IFMs could even be the primary source of foot rigidity during push-off.²²

Evidence from jumping tasks suggests that the IFMs modulate the geometry of the ankle-foot complex, by stiffening the medial longitudinal arch and metatarsophalangeal joints, in addition to shifting the point of application of the ground reaction force closer to the ankle joint.²³ This varying foot geometry modulates the torque of the ankle plantar flexors, as shown during walking and during jumping.²²⁻²³ The ankle plantar flexors produce a large amount of positive power during walking and running,²⁴ which is mainly delivered through elastic recoil of the Achilles tendon during propulsion.²⁵ Other extrinsic foot muscles also

assist foot function during locomotion. For example, tibialis posterior tendon stores energy during pronation occurring in early walking stance, and returns that strain energy during propulsion.²⁶

Mechanoreceptors play a key role in proprioception. Proprioception is the sense of body segment positions, movements, loads applied to limbs and forces applied by muscles.¹¹ Specifically, slowly adapting type II fibres located in the sole of the foot and muscle spindles located in the IFMs appear to contribute to postural control during standing.²⁷

In summary, the ankle-foot complex functions as an integrated system with both mechanical and sensorimotor roles, absorbing, storing, and generating energy during locomotion while adapting its geometry and supporting postural control through coordinated interactions among passive tissues, intrinsic and extrinsic muscles, and the central nervous system.

II. Ankle sprain and chronic ankle instability

Epidemiology and societal burden

An ankle sprain is an acute strain or tear of one or more ankle ligaments (Figure 4). Ankle sprains involve the lateral ligament complex in 80-90% of the cases, while the medial ligaments (i.e. deltoid) and the distal tibiofibular syndesmosis are injured in 5-15% and 10% of ankle sprains, respectively.²⁸ Concomitant injuries are also frequent,²⁹ with for example a syndesmotic rupture in addition to a sprain of the lateral ligaments, resulting in delayed healing and return to sport times.³⁰

A lateral ankle sprain (LAS) is defined as an acute traumatic injury to the lateral ankle ligament complex caused by excessive inversion of the hindfoot or a combined plantarflexion and adduction of the foot, resulting in initial deficits of function and disability.³¹⁻³² LAS is the most common lower-limb musculoskeletal injury among the physically active population,³³⁻³⁴ and the most common injury reported in various sports, with incidence rates of 1-7 LAS per 1000 exposures.²⁸ A single LAS is associated with a significant societal and financial burden, with short-term costs approximated at €820 per sprain in the Netherlands, mainly explained by the reduced productivity at work.³³ Athletes have an average time-loss duration of 2 to 4 weeks, which can sometimes be extended to several months.³⁵⁻³⁶

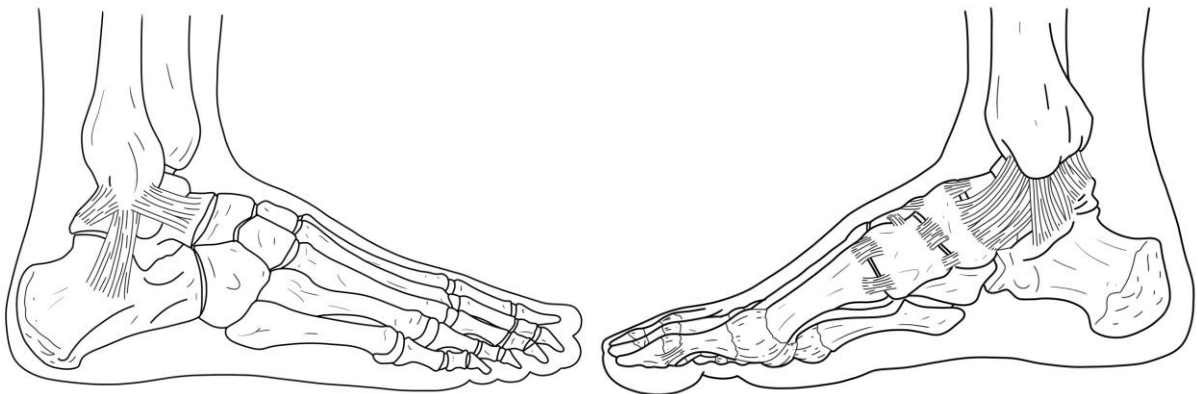


Figure 4. Lateral (left) and medial (right) view of the foot. Lateral ankle ligaments include the anterior talofibular, calcaneofibular and posterior talofibular ligaments (from anterior to posterior), as well as anterior and posterior inferior tibiofibular ligaments (not shown). Medially, the three branches of the deltoid ligament complex are shown. Figures in the public domain.³⁷⁻³⁸

Continuum of dysfunctions

Approximately 40% of individuals who sustained a LAS will experience chronic ankle instability (CAI).³³ CAI is a multifaceted condition defined by recurrent ankle sprains or episodes of the ankle giving way, along with symptoms such as perception of instability, pain, and altered function, that persist for more than one year after the initial LAS.^{31, 39} On the other hand, the remaining 60% of individuals who sustained a LAS will manage to return to their pre-injury level of physical activity without recurrent injury, symptoms, or self-reported function deficits, within the year that followed their more recent LAS. Those individuals are called ‘copers’.³⁹⁻⁴⁰ ‘CAI’ and ‘coper’ are the two extremes of a continuum of clinical presentations. Both groups were exposed to a similar initial injury, on the contrary to healthy participants, which renders the comparison between CAI and copers particularly relevant in research.³⁹⁻⁴⁰

The International Ankle Consortium has published criteria to select participants with CAI in research settings,³¹ that completed existing criteria for copers (Table 1).⁴⁰ Both groups sustained at least one ‘substantial LAS’ that occurred more than one year prior inclusion in the study. A ‘substantial LAS’ results in at least one interrupted day of desired physical activity,³¹ or three days in another relevant source.⁴⁰

Table 1: Selection criteria for individuals with chronic ankle instability, copers and healthy controls used in research

Item	CAI	Coper	Healthy control
First ‘substantial’ LAS	≥ 1 year ago	≥ 1 year ago	None
Last ‘substantial’ LAS	≥ 3 months ago	≥ 1 year ago	None
Ankle giving way in the last 6 months	≥ 2	None	None
Perceived instability	Yes	No	No
Self-reported function	If relevant for the research question	Not altered	Not altered

Legend: CAI (chronic ankle instability), LAS (lateral ankle sprain).

Individuals with CAI had their last substantial LAS at least three months prior enrolment, and they must report perceived instability. This central component of CAI can be estimated by using the Cumberland Ankle Instability Tool (CAIT).³¹ Suggested CAIT cut-off scores have varied from $\leq 27/30$ in the 2010s,⁴¹ to $\leq 25/30$ in 2014,⁴² while the latest model of CAI mentions $\leq 24/30$,³⁹ although it specifically refers to the article that validated $\leq 25/30$.⁴² Furthermore, language-specific versions of the CAIT were validated on heterogeneous populations, resulting in important variation in cut-off scores. For example, the French-version uses ≤ 23 while the Dutch-version uses ≤ 12 ,⁴³⁻⁴⁴ raising concerns about the validity of these thresholds. Self-reported function can also be part of inclusion criteria for CAI, especially if altered function is relevant for the research question.³⁹ Foot and Ankle Ability Measure (FAAM) questionnaire is made of two subscales that assess self-reported function during daily living and sport activities.^{42, 45}

Coper may have sustained more than one LAS as soon as their last sprain occurred more than one year prior study participation.⁴⁰ They should score $\geq 28/30$ on the CAIT,³⁹⁻⁴⁰ and $> 95\%$ on both FAAM-subscales.³⁹ Return to pre-injury level of physical activity could be assessed using Tegner activity-level scale,⁴⁶ but pre-injury data are often missing. ‘Healthy control’ individuals have the same inclusion criteria as copers; except they did not sustain any ankle sprain.

Selection criteria used in research are strict, reducing the risk of false positives in either group. In clinical settings, health care professionals should consider altered function more holistically and address even slight deficits through appropriate rehabilitation.⁴⁷

Development pathways of chronic ankle instability

Evolution of paradigms

Models explaining the high occurrence of CAI have evolved since Freeman speculated in 1965 that ankle instability was caused by functional and/or mechanical instability, referring to complaints such as ankle giving way and measurable deficits such as joint laxity.⁴⁸⁻⁴⁹ Tropp et al. then defined functional instability as a sensorimotor issue in the 1980s, with altered sensory afferences and altered motor behaviour leading to instability.⁵⁰⁻⁵² After a review of ankle instability mechanisms in 2000,⁵³ Hertel framed CAI as a heterogeneous condition through a Venn-diagram that included several impairments within functional and mechanical instability.⁵⁴ Then, Hiller introduced ‘recurrent ankle sprain’ as a distinct component of CAI.⁵⁵

The most recently published comprehensive model distinguishes pathomechanical, motor, behavioural, sensorial, and perceptual impairments. These deficits interact, under the influence of personal and environmental factors, to result in the clinical presentation on the continuum from CAI to coper (Figure 5).³⁹ This model integrates bio-, psycho-, and social constructs,⁵⁶ and can be interpreted in light of the International Classification of Impairments, Disabilities, and Handicaps paradigm.⁵⁷

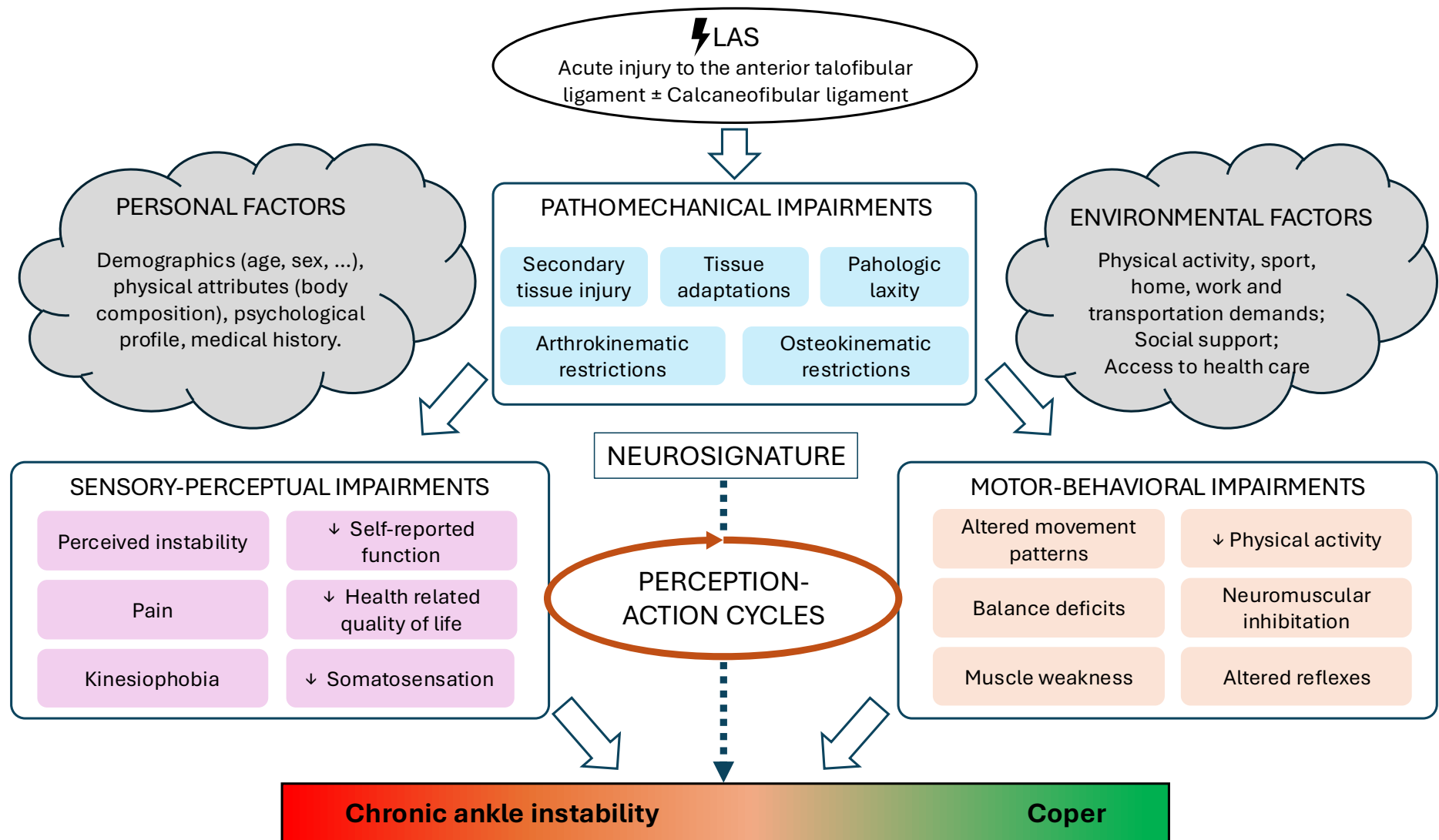


Figure 5. Impairments and factors contributing to the clinical outcome of an individual who sustained a lateral ankle sprain, adapted from Hertel and Corbett.³⁹

Current understanding of chronic ankle instability

In the model of Hertel and Corbett (Figure 5), pathomechanical deficits involve impairments of structural integrity and movement amplitudes. Pathologic ligament laxity,⁵⁸ ligament structural changes,⁵⁹ and cartilage and osteochondral lesions⁶⁰⁻⁶¹ are visible in patients with CAI, especially in ‘secondary tissue injuries’ resulting from recurrent episodes of LAS or ankle giving way.³⁹ Tissue adaptations also involve the muscles of the lower leg, with atrophy,⁶² architectural,⁶³ and composition changes.⁶⁴ However, there has been very little investigation of IFM sizes in CAI,^{62, 65-66} while this would inform rehabilitation. Regarding movement amplitudes, osteokinematic restrictions mainly involve limitations in dorsiflexion,⁶⁷⁻⁶⁸ while arthrokinematics restrictions include a limitation of posterior talar glide,⁶⁹ an internal rotation position of the talus relative to the tibia,⁷⁰ and conflicting changes in fibular position.⁷⁰

Individuals with CAI report perceptual impairments such as perceived instability,⁷¹ pain,⁷² fear of movement (i.e. kinesiophobia), and diminished self-reported function and health-related quality of life.^{71, 73} They also have somatosensory difficulties affecting proprioception,⁷⁴ force sense,⁷⁵ and plantar sensation.⁷⁶ The integration of competing sensory inputs may be compromised, with higher reliance of visual modalities.⁷⁷

CAI is associated with decreased ankle strength,⁷⁸⁻⁷⁹ and modified ankle muscle activation patterns during locomotion.⁸⁰ Knee, hip, and even trunk muscle groups are also affected by decreased strength,^{78, 81} and adapted activation patterns.⁸² However, few research examined strength at the foot level.⁶⁵ Motor-behavioral impairments also include decreased level of physical activity, balance deficits, and altered movement patterns.³⁹ Little studies examined movement patterns of the foot joints,⁸³⁻⁸⁸ despite high demands placed on the foot during locomotion and sporting tasks.

Perception-action cycles represent the continuous mutual interaction between sensory-perceptual and motor-behavioural impairments. The neurosignature, placed in the center of the model, refers to the specific neural activity pattern of each individual that results from the continuous integration of sensory, emotional, cognitive and motor information, influenced by personal and environmental factors.³⁹

Hertel and Corbett's model is mainly based on cross-sectional studies. Therefore, most 'impairments' and 'factors' must be regarded as potentially contributing factors, rather than prospectively-validated risk factors.⁸⁹

More holistically, LAS is trivialized by many athletes, patients and therapists. Many athletes return to play less than 24 hours after their injury,⁹⁰ although natural ligament healing lasts several weeks,⁹¹ and approximately the half of patients do not seek treatment.⁹² Additionally, only 45% of primary care physicians were identified with good knowledge about ankle sprain,⁹³ and 25-45% of physiotherapists properly followed assessment and treatment guidelines for LAS.⁹⁴⁻⁹⁶ Therefore, rehabilitation of LAS and CAI is likely suboptimal, while accessing an early and comprehensive treatment improves long-term functional status of patients with LAS.^{47, 97-98} McCann and Gribble speculated that inappropriate treatments and weak adherence of some patients contribute to the high occurrence of CAI.⁹⁹

CAI is thought to result from the complex interplay of multiple impairments, yet investigating all of them within a single study is not feasible. This PhD project therefore focuses on foot joint movement patterns during sporting tasks, as well as IFM sizes and strength.

III. Ankle and foot joint kinematics and kinetics during side-step cutting in CAI

Definitions and conventions

An analysis of movement refers to the description, analysis and quantification of displacements of body parts. The human body is typically modelled as segments, articulated by joints, enabled to move in different directions (degrees of freedom).¹⁰⁰⁻¹⁰¹ The sagittal plane separates left and right, the frontal plane separates anterior/posterior, and the transverse plane separates proximal/distal.¹⁰⁰⁻¹⁰¹ When it comes to the foot, conventions for planes of movement and ‘neutral’ foot position vary.¹⁰² In this thesis, will use the aforementioned planes and we define ‘neutral’ foot position with 90° relative to the shank in the sagittal plane. Movements of the foot relative to the shank are ‘plantar-/dorsiflexion’ (sagittal plane), ‘inversion/eversion’ (frontal plane), and ‘adduction/abduction’ (transverse plane).¹⁰²

Kinematics is the study of the position, velocity and acceleration of body parts. It requires a system able to track the position and orientation of body segments over time, using for example high-speed cameras tracking reflective markers placed on anatomical landmarks. Each body-model has its own definitions of segments, influencing the kinematic outcomes. At least three markers per segment are required to characterize movement in the three planes.¹⁰⁰

Kinetics is the study of forces and moments causing movements. Force plates are used to collect external forces exerted between the body and its environment. Internal joint moments represent the net result of all active (muscles) and passive (ligaments, intraarticular structures) forces that produce rotation around a joint axis. Inverse dynamics is used to compute net internal joint moments by applying Newton–Euler equations to each segment, from distal to proximal, knowing kinematics and ground reaction forces (Figure 6). Muscle power at a joint represents the rate of energy generation or absorption and is calculated as the product of the net joint moment and the joint angular velocity. Positive values indicate energy generation (net concentric action), whereas negative values indicate energy absorption (net eccentric action).¹⁰⁰

Spatiotemporal parameters refer to several constructs related to time and space, such as walking speed, the time between two ground reaction forces events, and step frequency.

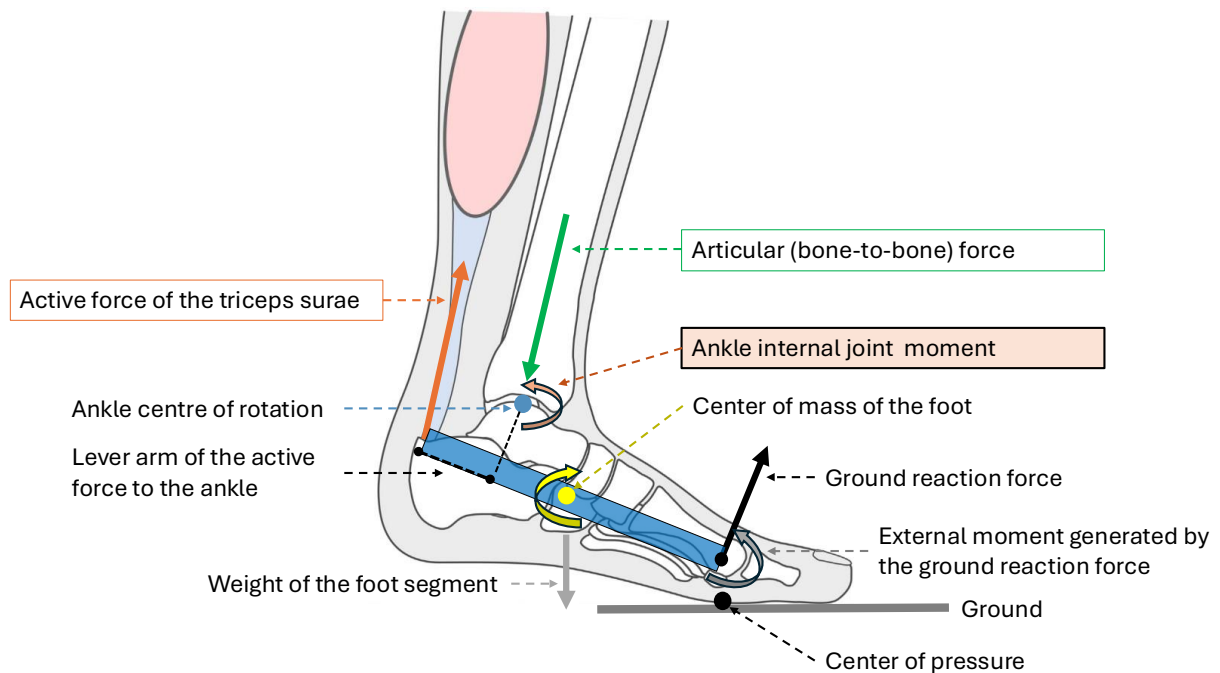


Figure 6. Forces and moments acting on the foot during the push-off phase of running or side-cutting, and the principle of inverse dynamics at the ankle joint. The ground reaction force (black arrow) applies at the foot centre of pressure, and both are measured with the force plate. The horizontal and vertical components of this ground reaction force generate an external moment at the ankle (grey curved arrow), proportional to their lever arm to the ankle joint centre. The position, velocity, and acceleration of the foot centre of mass (yellow) are obtained from motion capture, while segment mass and inertia are estimated from anthropometric data. These allow the calculation of inertial forces and moments acting at the foot centre of mass. Inverse dynamics then determines the net internal joint moment at the ankle (orange curved arrow) as the moment required to balance the sum of the external moments (i.e. ground reaction, gravity of the foot segment, and inertial terms). This internal moment is mainly produced by the active muscle force of the triceps surae, in addition to articular reaction forces. Figure adapted from Iaquinto (2023), according to the description of Winter (2019).¹⁰⁰⁻¹⁰¹

Multi-segment foot models

Most body-models, such as the Plug-in Gait model,¹⁰³ represent the foot as a single rigid segment. Research has shown that this approach overestimates ankle joint kinematics and kinetics, and neglects the contribution of more distal foot segments.¹⁰⁴⁻¹⁰⁵ Therefore, *kinematic* models dividing the foot into several segments were published more than 25 years ago.¹⁰⁶⁻¹⁰⁷ Since then, dozens of multi-segment foot models (MSFMs) have been developed, including 2-26 individual foot segments.¹⁰⁸ Among those, Rizzoli MSFM allows to distinguish the shank (tibia and fibula), hindfoot (talus and calcaneus), midfoot (cuboid, navicular, cuneiforms), forefoot (metatarsals), and hallux (Figure 7). These segments are articulated by the ankle, Chopart, Lisfranc and metatarsophalangeal joints, enabled to rotate with 6 degrees of freedom, except the metatarsophalangeal joint that only moves in flexion/extension. This kinematic MSFM requires 16 reflective markers per side,¹⁰⁹⁻¹¹⁰ and benefits from 1.5 to 5.6° between-session variability in kinematic outcomes, among trained examiners.¹¹¹

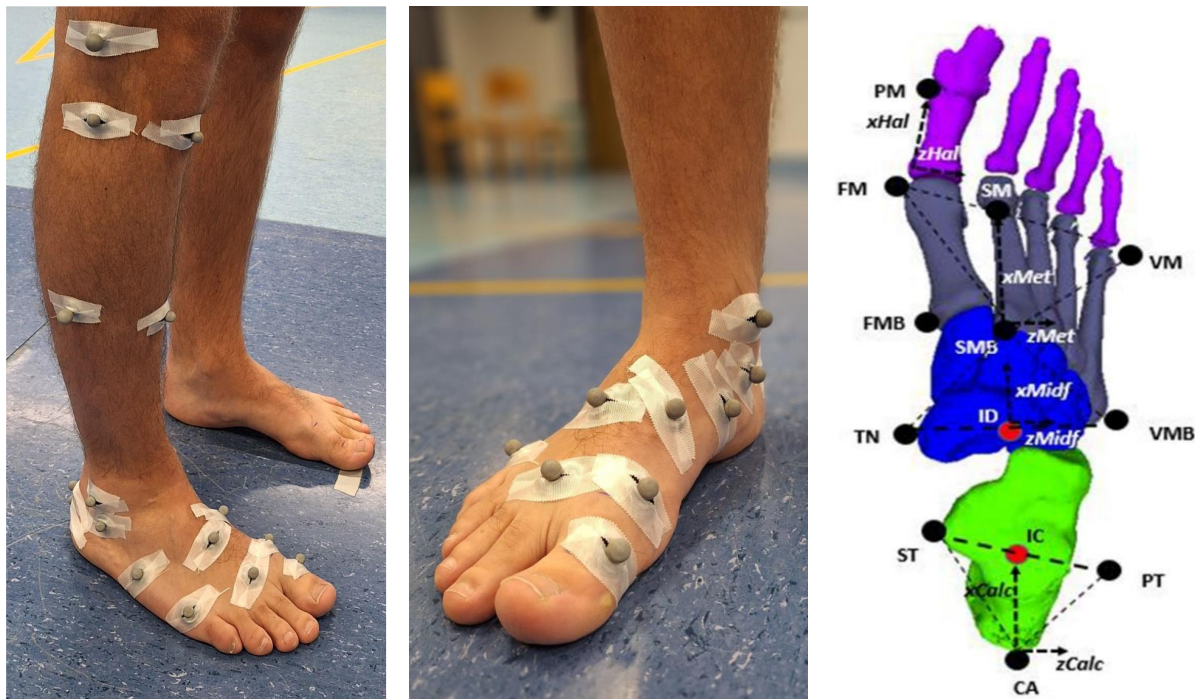


Figure 7. Placement of reflective marker according to the Rizzoli foot model. This model is composed of the shank, hindfoot (green), midfoot (blue), forefoot (grey), and hallux (purple).¹¹²

More recently, the body of knowledge has expanded to *kinetic* MSFMs, resulting in foot joint moment, power, and work.^{109, 113-114} In order to apply inverse dynamics, researchers need – for every time frame and each foot segment – to determine the resulting ground reaction force and the centre of application of that force. Therefore, Bruening et al. asked participants to walk while ensuring that their feet landed between two adjacent force plates,¹¹⁵ while Saraswat et al. recorded force and plantar pressure simultaneously, by placing a pressure plate on top of a force plate.¹¹⁴ Both research groups calculated joint kinetics using three segment foot models.¹¹⁴⁻¹¹⁵ Later, Deschamps et al. expanded this method to a four-segment foot model, which separates the metatarsal bones from the midfoot. In this paradigm, foot segment locations are obtained by vertically projecting marker coordinates onto the pressure matrix (Figure 8), and force point of application is calculated as the segment centre of pressure.¹⁰⁹ The pressure plate records vertical forces, but not the medio-lateral or antero-posterior forces. A ‘proportionality scheme’ was applied to estimate horizontal forces based on the relative distribution of vertical forces across each segment, for each time frame (Equation 1 and Equation 2).^{109, 114}

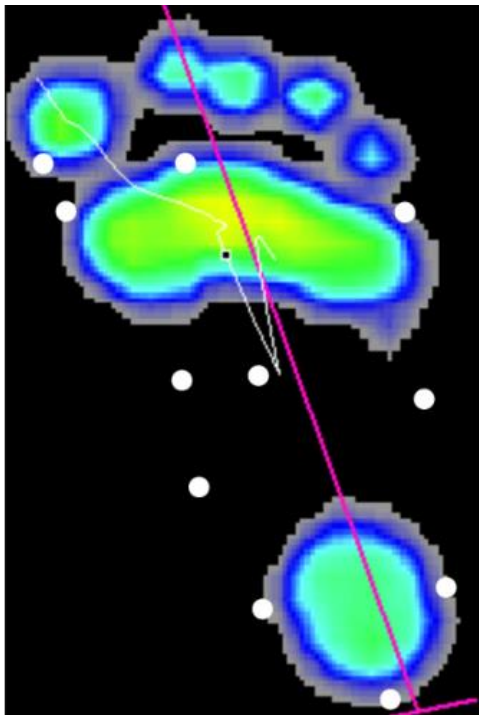


Figure 8. Vertical projection of the position of reflective markers (white dots) on the pressure matrix.

Equation 1: $Fx_i = Fx \times \frac{Fz_i}{Fz}$,

Equation 2: $Fy_i = Fy \times \frac{Fz_i}{Fz}$,

Where Fx_i is the antero-posterior force of segment i (unknown), Fy_i is the medio-lateral force of the segment i (unknown), Fx and Fy are respectively the total antero-posterior and medio-lateral forces recorded by the force plate, and $\frac{Fz_i}{Fz}$ is the proportion total foot pressure that apply to segment i (pressure plate).

Ankle/foot joint kinetics characterize joint loading patterns. They have already been investigated during walking among individuals with diabetes,¹¹⁶ foot joint osteoarthritis,¹¹⁷⁻¹¹⁸ and tibialis posterior tendon dysfunction.¹¹⁹ It has also been reported among healthy participants during running, serving as a proof of concept to acquire foot joint kinetics during sporting tasks.¹¹²

Statistical methods

Kinematics, kinetics, and ground reaction forces are smooth spatiotemporal signals, also called ‘one-dimensional’ signals.¹²⁰ Discretization is the process of calculating ‘zero-dimensional’ values of interest from one-dimensional data (Figure 9).¹²⁰ Discretized variables consist in peaks, slopes, ranges, areas under curves, and durations.

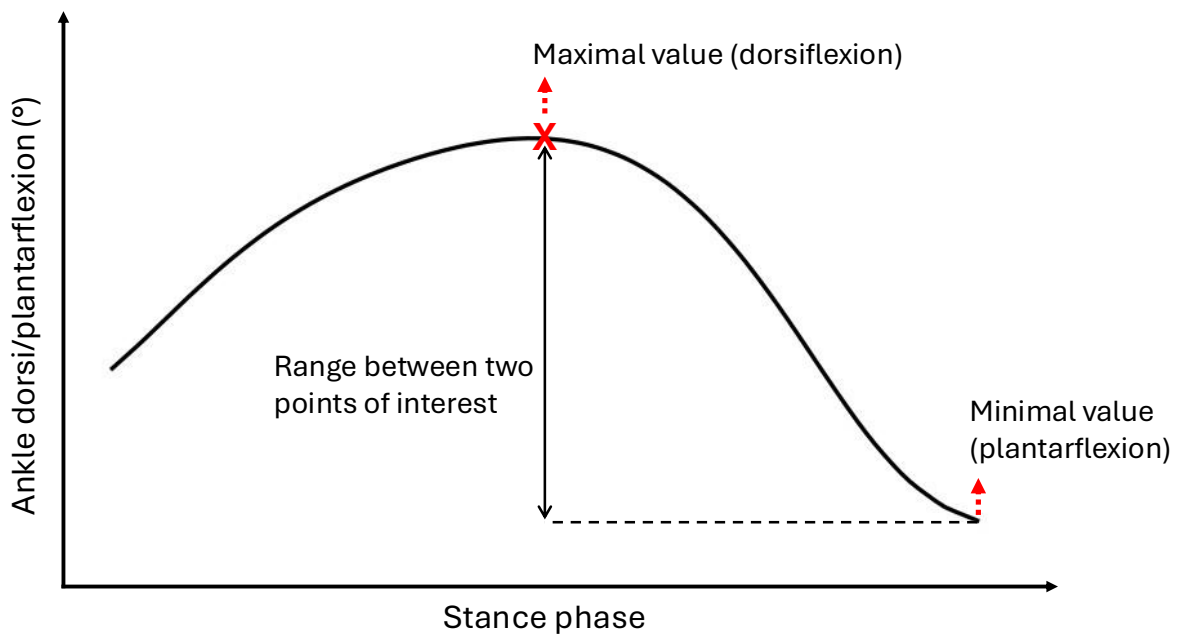


Figure 9. Discretization of a one-dimensional signal into zero-dimensional values of interest. In this example, sagittal plane ankle range of motion is computed from the corresponding continuous spatiotemporal signal captured during the stance phase of running.

Performing statistical inference on zero-dimensional, discretized data is a valid practice, especially if values of interest are selected based on sound *a priori* hypotheses. Still, discretization can compromise the integrity of the underlying continuous signals, potentially obscuring or reversing trends,¹²⁰ and it increases the probability of false positives.¹²¹ Statistical Parametric Mapping (SPM) and its non-parametric equivalent (SnPM) overcome these limitations by applying the general linear model in combination with random field theory to smooth continuous data, allowing statistical inference across entire waveforms. SPM preserves the original spatiotemporal structure and maps results directly onto the biomechanical domain, facilitating clear and meaningful interpretation of significant effects (Figure 10).^{120, 122-124}

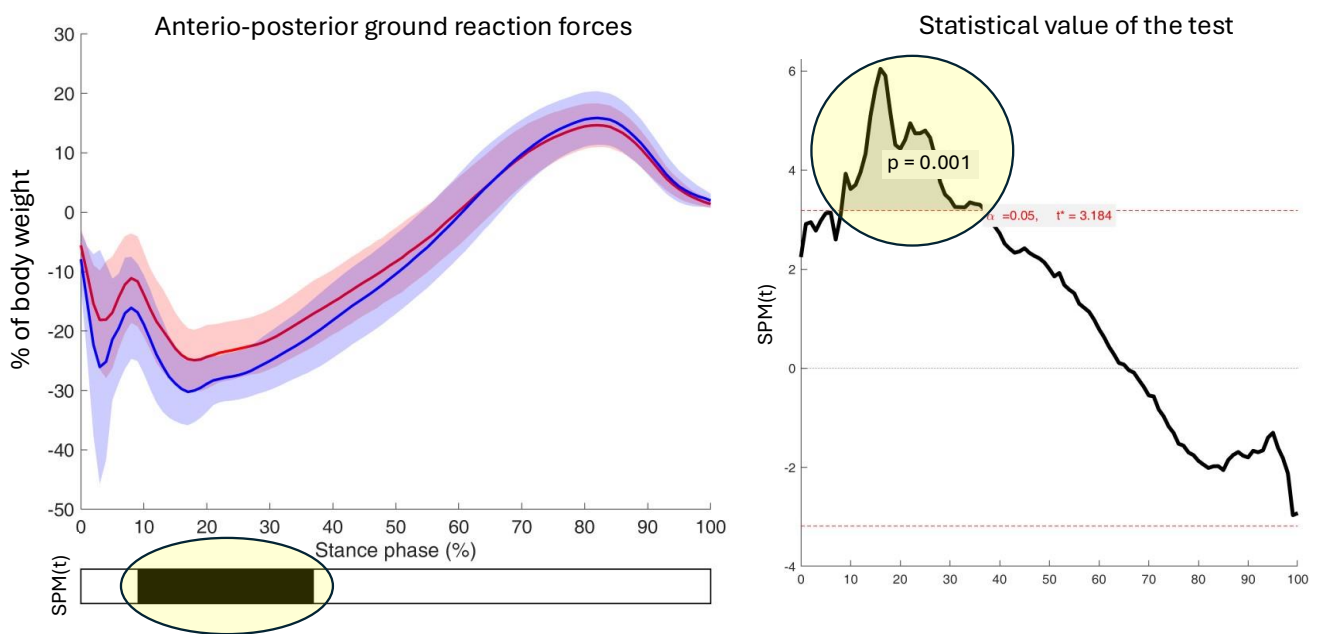


Figure 10. Inference using statistical parametric mapping. This example shows how one-dimensional data (left) can be compared using SPM (right). A difference between conditions is considered significant (yellow circles) when $SPM(t)$ exceeds $\pm t^*$ critical values (dotted red lines).

Relevance of side-step cutting

Altered movement patterns during locomotion and sporting tasks are part of the impairments suspected to contribute to CAI.³⁹ Many sports involve changes of direction, which place the ankle in a vulnerable supinated position,¹²⁵⁻¹²⁷ potentially contributing to the high occurrence of LAS among such sports.¹²⁸⁻¹²⁹ Because of the high recurrence of LAS and the risk of developing CAI, studies should investigate foot joint motion and loading patterns to better understand ‘motor-behavioural’ impairments, as part of Hertel and Corbett’s model of CAI-pathways.³⁹

Side-step cutting is defined as performing a change of direction in the contralateral direction relative to the supporting foot and mainly varies in terms of cutting angle and approach speed (Figure 11).^{125, 127, 130-132} As the cutting angle increases, greater deceleration demands are imposed, requiring larger braking forces and, in some cases, multiple steps to complete the manoeuvre. A 45° side-step cutting represents a compromise that is representative of cutting actions in team sports.¹³³⁻¹³⁴ It is acute enough to require substantial deceleration while still being obtuse enough to allow a change of direction within a single foot contact.¹³³

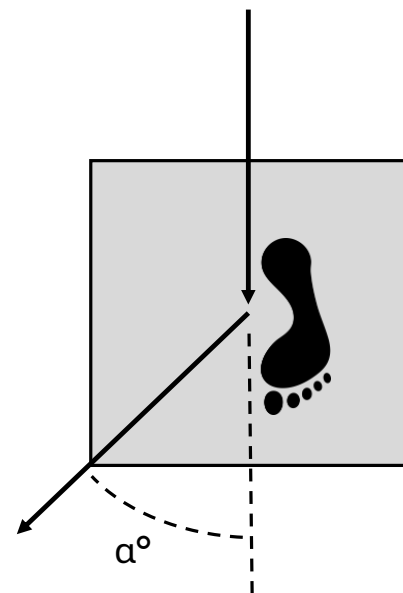


Figure 11. Side-step cutting. Cutting tasks mainly differ in approach speed and cutting angle (α). Written consent has been obtained to use this picture.

The ground reaction forces observed during a 45° side-step cutting are thus characterized by posterior forces up to mid-stance, followed by anterior forces, as well as medial forces throughout the stance phase (Figure 12).¹³⁴

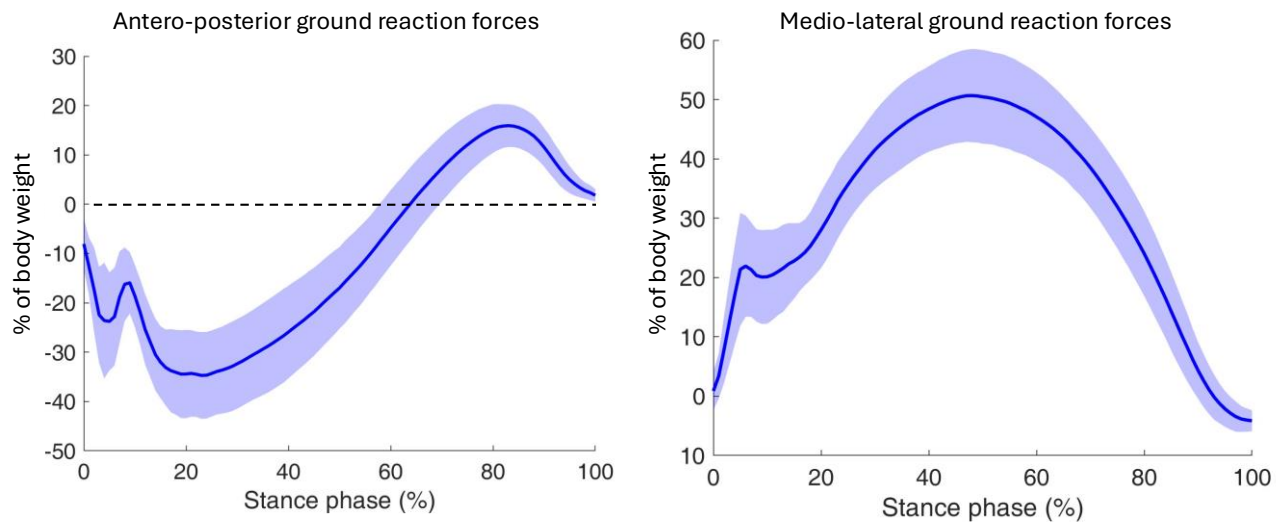


Figure 12. Anteroposterior (left) and mediolateral (right) components of the ground reaction forces during side-step cutting. Mean (solid line) and standard deviation (cloud). Anteroposterior ground reaction forces are characterized by a period of horizontal deceleration (forces below the dotted line), followed by a period of horizontal acceleration (forces above the dotted line). Mediolateral ground reaction forces are oriented medially during the entire stance phase.

Besides, the vertical ground reaction forces generally include an impact peak force shortly after foot strike and an active peak force occurring later,^{83, 134-135} which can serve as markers to define the impact peak, active peak and propulsive subphases of stance (Figure 13).^{86, 136}

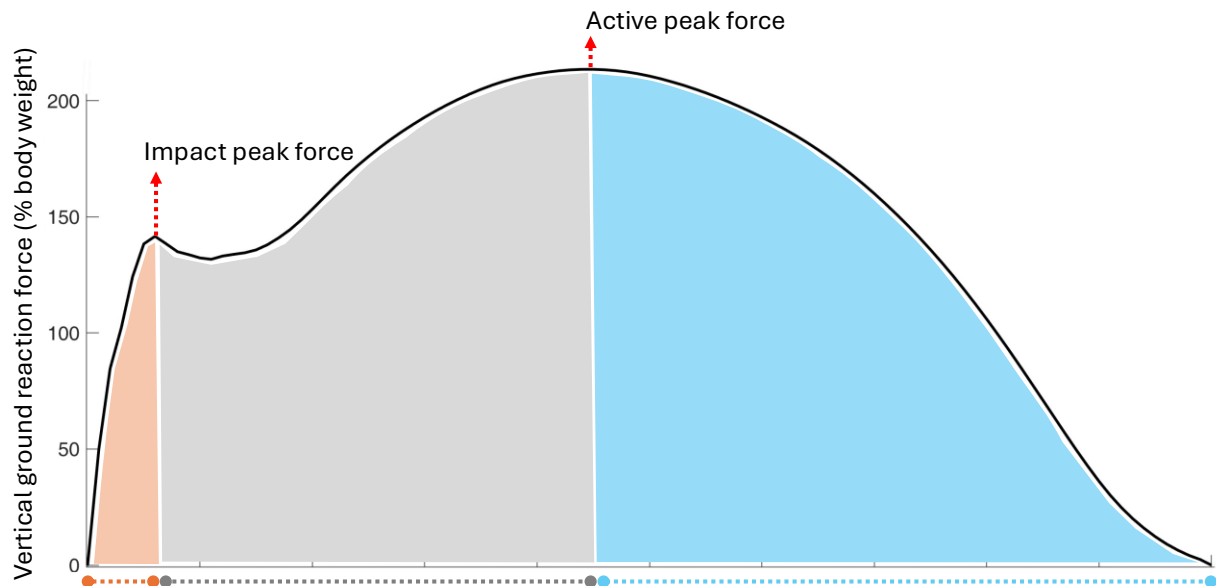


Figure 13. Vertical ground reaction forces during the stance phase of 45° side-step cutting. The impact peak and active peak forces define the impact peak (orange), active peak (grey) and propulsive (blue) subphases.

Critical appraisal

During side-cutting, literature compared foot joint kinematics between females with and without CAI,⁸³ between CAI individuals with or without laxity,¹²⁷ and between copers with or without laxity.¹²⁵ Overall, joint laxity was found to increase ankle abduction in late stance,^{125, 127} but the paucity of research limits our understanding of ankle and foot mechanics in CAI.

Although limited evidence is available during running, this movement shares key mechanical characteristics with side-step cutting – such as occurring within a single stance phase – and can therefore provide complementary insights.⁸⁴⁻⁸⁸ Compared to healthy controls, these studies reported more ankle eversion and more medial forefoot inversion during the second half of stance using SPM,⁸⁴ altered sagittal plane and transverse plane ankle ranges of motion, increased Chopart frontal plane range of motion,⁸⁶ and a more adducted ankle in early stance.⁸⁵

Side-step cutting foot joint kinematics is thus mostly unexplored among individuals with CAI. Further, no prior study compared CAI and copers, despite the crucial importance of copers to formulate hypotheses about CAI development pathways.³⁹ Moreover, foot joint kinetics have not yet been investigated in CAI, although structural changes and limitations in primary and accessory joint movements associated with CAI could alter foot kinetics,³⁹ as it was seen in other foot joint-related disorders.¹¹⁶⁻¹¹⁹ This PhD aimed to fill these gaps by investigating ankle and foot joint kinematics and kinetics during side-step cutting among recreationally active participants with CAI.

IV. Intrinsic foot muscles

Foot strength, foot muscle sizes and ultrasound imaging

The intrinsic and extrinsic foot muscles are closely related to the function of the IFMs (*cfr.* pp. 33-34), and altered IFM function could thus contribute to lower-limb musculoskeletal disorders.¹⁰ Analytically, IFM function can be investigated as their task-specific capacity to generate force and contribute to foot strength.

Foot strength can directly be assessed through hallux flexion, toe flexion and foot doming tasks, using dynamometers or plantar pressure platforms.¹³⁷⁻¹⁴⁰ Hand-held dynamometry of hallux flexion and toe flexion is a clinically-accessible technique that benefits from good to excellent test-retest and inter-rater reliability.^{139, 141} Placing the ankle in plantarflexion keeps constant the contribution of the IFMs and decreases that of the extrinsic foot muscles compared neutral ankle position.^{139, 142} However, measures of foot strength are not able to determine the exact contribution of the IFMs during such tasks.

Imaging techniques are used to estimate the strength of individual muscles indirectly, supported by the fair to good correlations between the sizes and strength of a muscle.¹⁴³⁻¹⁴⁶ Magnetic resonance imaging is the gold standard method, showing excellent reliability for IFM sizes.¹⁴⁷⁻¹⁴⁸ However, it is also costly and poorly available, whereas ultrasound imaging benefits from excellent concurrent validity with magnetic resonance imaging.¹⁴⁸ In a systematic review, we highlighted that ultrasound imaging had good to excellent reliability to measure IFM sizes.¹⁴⁹ However, most studies used high-end ultrasound devices, which typically have better image resolution.¹⁵⁰ Therefore, reliability of IFM size needs to be confirmed using a low-end ultrasound device.

Foot muscle strength and sizes in individuals with chronic ankle instability

Reduced strength (i.e. weakness) belongs to the motor-behavioral impairments suspected to contribute to CAI after an initial LAS.³⁹ Weakness has been well demonstrated at the extrinsic foot muscles, with lower ankle inversion, eversion,⁷⁸⁻⁷⁹ and plantarflexion⁷⁹ strength in CAI compared to healthy controls. At the foot level, preliminary evidence suggests impaired hallux and toe flexion strengths (Cohen's $d = 1.3-1.7$ in CAI vs. controls),^{65, 79} that needs to be confirmed.

Reduced muscle sizes (i.e. atrophy) is listed among the pathomechanical impairments that could contribute to CAI, including thigh and leg muscle groups.³⁹ However, it is unclear if CAI is associated with diminished IFM sizes, since it has been very little examined. A first paper published in 2016 reported important deficits in adductor hallucis obliquus and flexor hallucis brevis muscle sizes (Cohen's $d > 2.3$) – but no significant differences in other IFMs – compared to healthy controls. That study used magnetic resonance imaging and involved a restricted sample size.⁶² Later articles employed ultrasound imaging and pointed towards no between-group differences,^{65, 151} except Jaffri et al. who very recently showed smaller abductor hallucis and flexor digitorum brevis cross-sectional areas in CAI compared to healthy controls ($d = 1.4-1.5$).⁶⁶

Since the IFMs contribute to dynamic foot postural control,⁹⁻¹⁰ further research should aim to investigate foot muscle strength and sizes in CAI, to inform assessment and rehabilitation guidelines for CAI.

Neuromuscular fatigue

As detailed above (*cfr.* p. 33), the IFMs contribute actively to energy storage and release during locomotion,¹⁹⁻²² in addition to their role in postural balance.¹⁵² A part of these roles were deducted by observing functional changes under tibial nerve block,^{22-23, 153-154} a procedure that drastically decreases muscular activity.¹⁵⁵ The main drawback of tibial nerve block is the associated decreased proprioception and cutaneous perception, potentially altering lower limb function.¹⁵⁶

Neuromuscular fatigue is defined as a loss of maximal force-generating capacity resulting from muscular activity,¹⁵⁷ which thus mimics altered muscle activation. Various protocols were tested to produce IFM fatigue, leading to contrasting results on strength, static foot biomechanics, and walking biomechanics.¹⁵⁸⁻¹⁶³ These inconsistent results could be attributed to the low loads applied to the feet and the short intervention durations. Performing protocols under a loaded condition would benefit from the spontaneous increase in IFM activity when load is applied to the lower-limbs and feet.^{19, 164-165}

Besides, voluntary contraction of the IFMs can be difficult to achieve, and electrical stimulation applied to the medial longitudinal arch has been proposed to assist IFM activation.¹⁶⁶⁻¹⁶⁷ Electrical stimulation is also sometimes used as a standalone intervention to achieve neuromuscular fatigue.^{158, 168-169} Therefore, combining voluntary contractions with electrical stimulation could help achieve higher levels of fatigue.

No study has previously investigated the effects of IFM fatigue on foot joint kinematics or kinetics among individuals with CAI. Such studies would provide insights into the consequences of altered IFM function in that population.

Intrinsic foot muscles strengthening

Foot muscle function may be impaired among individuals affected with various musculoskeletal disorders such as CAI,⁶⁶ plantar fasciitis,¹⁷⁰ and hallux valgus deformity.¹⁷¹ Foot muscle strengthening is a neglected intervention modality in the rehabilitation of lower-limb injuries.¹⁶⁷ Besides, individuals affected with neurological and systemic diseases¹⁷²⁻¹⁷³ and older people may also have IFM atrophy,¹⁷⁴ and they may benefit from interventions aiming to strengthen the IFMs.¹⁷⁵⁻¹⁷⁶

Specific exercises and the use of minimalist footwear were suggested to improve foot strength and/or IFM function.¹⁶⁶⁻¹⁶⁷ In the literature, ‘foot muscle exercises’ refer to specific interventions that were proved to activate the IFMs,¹⁷⁷ such as the short foot exercise.¹⁶⁶ Minimalist footwear is characterized by high flexibility, light weight, and the absence of stabilizing technologies to ensure minimal interference between the foot and the environment.¹⁷⁸ Despite the growing literature on the subject of foot muscle exercises and minimalist footwear (Figure 14), no previous meta-analysis of randomized controlled trials has investigated the effects of these interventions to increase foot muscle sizes. Besides, previous systematic reviews and meta-analyses only partially addressed the effect of these interventions on foot strength and locomotion biomechanics, have focused on a particular intervention¹⁷⁹⁻¹⁸⁰ or population^{175, 180-181} of interest. Therefore, a comprehensive systematic review with meta-analysis should be conducted to summarize the effects of foot muscle exercises and the use of minimalist shoes on foot strength, foot muscle sizes and locomotion biomechanics.

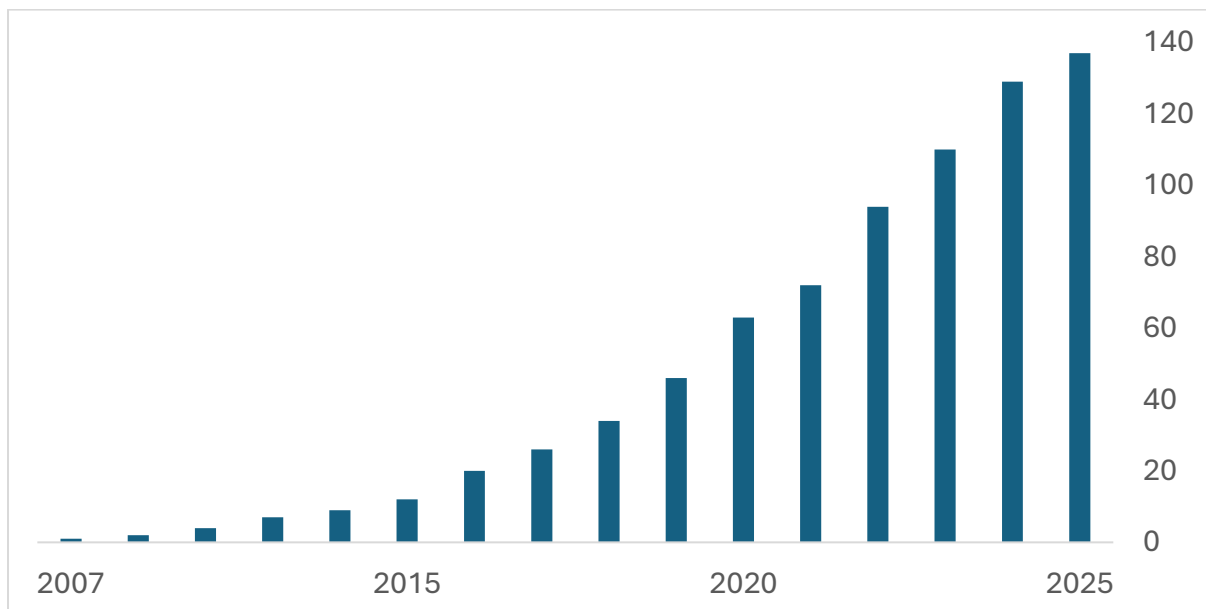


Figure 14. Cumulative number of publications available on PubMed. Search query (used September, 1, 2025): (intrinsic foot muscle[Title/Abstract] OR intrinsic foot muscles[Title/Abstract] OR foot core[Title/Abstract]) AND (exercise*[Title/Abstract] OR short foot[Title/Abstract] OR doming[Title/Abstract] OR strengthening[Title/Abstract] OR minimalist[Title/Abstract]).

V. Aims of the PhD project

Rationale

Chronic ankle instability (CAI) is characterized by altered movement patterns, which may contribute to the clinical presentation of these patients. Changes of direction are frequent in sports and place considerable mechanical stress on the ankle and foot. Despite the relevance of these sporting tasks, insufficient research has been conducted on foot joint kinematics and kinetics, particularly when comparing individuals with CAI those who have successfully recovered from lateral ankle sprains (LAS copers).

Furthermore, the intrinsic foot muscles (IFMs) contribute to foot arch support, energy restitution during propulsion, and proprioception. It is unclear if the sizes and strength of these muscles are altered in CAI, although it would have consequences on the suspected pathomechanical pathways for CAI and the content of rehabilitation. How IFMs should be strengthened is unclear, and the biomechanical consequences of IFM fatigue, which mimics altered function, are not yet fully understood. Ultrasound imaging appears as a reliable, non-invasive, and accessible method to assess muscle sizes, but its reliability needs to be confirmed using a low-end device.

Research objectives

This PhD project focused on two primary themes: foot joint dynamics and alterations in IFM sizes associated with CAI. In addition, the biomechanical effects of foot muscle fatigue were examined as a means of further elucidating movement patterns. Objectives related to individuals with CAI were therefore classified as **clinical objectives**.

Complementary challenges were also addressed, including the investigation of IFM strengthening and foot muscle fatigue protocols, as well as the assessment of ultrasound imaging reliability. These aims were classified as **methodological objectives**.

Clinical objectives

Objective 1. To characterize foot joint kinematics and kinetics in individuals with CAI, in comparison to LAS copers and healthy controls during side-step cutting (*Chapter 4*).

Objective 2. To evaluate IFM sizes and strength in individuals with CAI, relative to LAS copers and healthy controls (*Chapter 4*).

Objective 3. To assess the impact of a fatigue protocol targeting the IFMs on foot joint biomechanics during side-step cutting in participants with CAI (*Chapter 5*).

Methodological objectives

Objective 4. To conduct a systematic review of existing interventions aimed at strengthening the IFMs, including exercises and the use of minimalist footwear (*Chapter 1*).

Objective 5. To compare the effects of two fatigue protocols targeting the IFMs on hallux flexion strength and running biomechanics (*Chapter 2*).

Objective 6. To determine test-retest and inter-rater reliability of a low-cost ultrasound device to measure foot muscle sizes (*Chapter 3*).

Hypotheses

During 45° side-step cutting, we hypothesized that individuals with CAI would exhibit lower ankle sagittal plane range of motion (ROM) in early stance, along with lower ankle and Chopart joint power absorption and generation compared to healthy controls and copers (*Chapter 4*).

We expected lower IFM sizes in CAI, while controlling for body mass index (*Chapter 4*), and that a low-end ultrasound device could reliably measure foot muscle sizes (*Chapter 3*). We also assumed that specific exercises and the use of minimalist footwear would increase IFM sizes and foot strength (*Chapter 1*).

We hypothesized that repeated isometric and concentric toe flexor contractions would produce foot muscle fatigue and alter running ground reaction forces among healthy participants (*Chapter 2*). We expected that repeated isometric toe flexor contractions combined with electrical stimulation would increase frontal plane ankle ROM and frontal plane ankle peak moments and ankle peak power in individuals with CAI (*Chapter 5*).

Methodological studies

Chapter 1. The effects of foot core exercises and minimalist footwear on foot muscle sizes, foot strength, and biomechanics: a systematic review and meta-analysis

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Abstract

Background: Specific foot exercises and the use of minimalist shoes during running or daily life were suggested to strengthen the intrinsic foot muscles and to modify locomotion biomechanics. We aimed to review the effectiveness of these interventions to modify foot muscle sizes, foot strength, and biomechanical outcomes.

Method: PubMed, Embase, Cochrane Library and SportDiscus databases were searched (last update: 12 March 2024). Randomized controlled trials with intervention duration of four weeks or more were included. A random-effect meta-analysis was performed when at least two studies were available for an outcome. Standardized mean differences and corresponding 95% confidence intervals were reported. The certainty of the evidence was assessed by considering risk of bias, imprecision, inconsistency, and indirectness. The evidence not summarized in the meta-analysis was synthesised.

Findings: Twenty-eight trials (1399 participants) were included in the systematic review and seventeen meta-analyses were performed. From meta-analyses, minimalist shoes lead to greater strength of toes 2 to 5 (confidence interval = 0.02 to 0.76), and foot exercises resulted in lower medial longitudinal arch motion during running (confidence interval = 0.08 to 0.82). The certainty of the evidence was low to very low, mainly due to the limited number of available studies and high risk of bias. The systematic synthesis supported exercises and minimalist shoes to increase foot strength, but was conflicting for muscle sizes. The interventions occasionally modified walking and running biomechanical variables.

Interpretation: Foot exercises and minimalist shoes may be appropriate to increase foot strength and to induce biomechanical changes during dynamic tasks.

Introduction

The foot is a complex structure regulated by both the extrinsic and intrinsic foot muscles. The origins and attachments of the plantar intrinsic foot muscles (PIFMs) are located within the foot, and their resulting short moment arms designate them as local stabilizers for the foot.¹⁰ As a result, the PIFMs contribute to maintaining medio-lateral balance during single-leg stance,¹⁵² and during walking and running, they help attenuate foot impact upon landing and support foot stiffness during push-off.¹⁵³

Thicker PIFMs were observed in athletes compared to non-athletes.¹⁸² In contrast, decreased PIFM sizes have been seen in various musculoskeletal disorders,¹⁷⁰⁻¹⁷¹ neurological and systemic diseases,¹⁷²⁻¹⁷³ and among older individuals,¹⁷⁵ compared to healthy people. Common rehabilitation and return-to-sport approaches have given poor attention to interventions targeting the PIFMs.¹⁶⁷ These encompass foot exercises and specific footwear.¹⁶⁶ First, so-called foot muscle exercises – such as the short foot, foot doming, and the toe-spread-out exercises – were proved to activate the PIFMs.¹⁷⁷ Authors suggested to familiarize with these exercises through active-assisted contractions or with the aid of neuromuscular electrostimulation (NMES) targeting the medial foot muscles. They then recommend to combine the exercises into more global movements with the whole lower limb as progression.¹⁶⁶⁻¹⁶⁷ Second, the use of minimalist running shoes (MRS) during running or daily life activities was also associated with hypertrophy and strengthening of the PIFMs.¹⁸³⁻¹⁸⁴ MRS exhibit a high flexibility, lightweight construction, and a lack of stability features. As a result, they offer minimal interference between the foot and the surrounding environment.¹⁷⁸

The strength of the foot refers to the ability of the intrinsic and extrinsic foot muscles to generate force. Foot muscle strength can be assessed directly through toe dynamometry, foot mobility under load and foot posture.¹³⁸ Indirect muscle strength assessment methods have also been described, using imaging techniques such as ultrasound or magnetic resonance imaging to measure muscle sizes.^{138, 149, 185} Previous meta-analyses have demonstrated that PIFM exercises were linked to improved toe flexion strength, foot mobility, foot posture, and balance.^{138, 179-180, 186-187} Additionally, systematic reviews have outlined the growth in muscle sizes and modifications in lower limb biomechanics following these exercises,^{175, 181} as well as following the use of MRS.^{181, 188} To our knowledge,

no previous meta-analysis investigated the effects of PIFM exercises on PIFM sizes, and no meta-analysis of randomized controlled trials investigated the benefits of MRS. Additionally, the existing meta-analyses and systematic reviews on foot strength and biomechanics have included a limited number of articles. This may be explained by their focus on specific PIFM exercises (e.g. the short foot exercise),¹⁷⁹⁻¹⁸⁰ their population of interest, which is often pes planus, older people or asymptomatic participants,^{175, 180-181} and search strategies that may have missed important synonyms.^{175, 181, 189-190} Finally, the field of PIFM strengthening is rapidly evolving,¹⁹¹⁻¹⁹⁴ suggesting that more articles could be included into existing meta-analyses and that new meta-analyses may be feasible.

Therefore, the objective of this systematic review and meta-analysis was to thoroughly examine the effects of PIFM exercises and the use of MRS on PIFM sizes, foot strength, and biomechanics during dynamic tasks in both asymptomatic and symptomatic participants.

The hypotheses were that both foot exercises and MRS would result in increased PIFM sizes and foot strength. During running and walking, it was expected that these interventions would result in kinematic changes characterized by a forefoot strike pattern and a decreased medial longitudinal arch (MLA) mobility, along with kinetic and spatiotemporal changes.

Methods

1. Study design

This systematic review follows the Preferred Reporting Items for Systematic Reviews and Meta-Analysis (PRISMA). The protocol was prospectively registered in the PROSPERO database (CRD42023411084).

2. Sources and research strategy

PubMed (Medline), Embase, Cochrane Library and SportDiscus databases were consulted. The Cochrane Collaboration recommends at least searching in the three first databases,¹⁹⁵ while SportDiscus is more specific to sports medicine. Databases were searched using equations based on two search strings: The first included synonyms of PIFMs and the second synonyms of foot exercises alongside MRS. These terms were searched using Title, Abstract, Thesaurus, and Keyword fields, without filter. A team of librarians helped in creating this search strategy. The initial search was conducted on April 14, 2023, and subsequently updated on March 12, 2024, using the same search methodology (Appendix 1.A). Additionally, a manual search was conducted within the reference lists of previous systematic reviews and meta-analyses.^{131, 175, 179-181, 186-189, 196}

3. Eligibility criteria

Randomized controlled trials (RCT) with full texts available in English or French were included if their intervention lasted a minimum of 4 weeks and involved either PIFM exercises or the use of MRS during running or daily activities. The main outcomes were PIFM sizes as well as hallux, toes, and doming strength. Secondary outcomes included lower limb biomechanics during dynamic tasks such as walking, running, side-cutting, and jump-landing. Studies were included if at least one of these outcomes was assessed at baseline and immediately following the intervention.

4. Study selection

The search of available resources was conducted by two investigators (M.R. and L.C) among the four databases. Duplicates were automatically removed using EndNote software, and the remaining articles were imported to Rayyan website where any remaining duplicate was manually removed. Two reviewers (J-L.P-D. and M.R.) independently selected articles based on titles and abstracts before

resolving any disagreement. The articles were retrieved online, and authors were contacted whenever necessary. Then, the same two reviewers, who were still blinded from each other, selected the retrieved articles through full-text reading and specified the reason for exclusion, such as language barrier, study design, intervention, outcomes, or the population. A third investigator (L.C.) was consulted to resolve any remaining disagreement.

5. Risk of bias assessment

Version 2 of Cochrane risk-of-bias tool for randomized trials (RoB2) was used to assess the risk of bias in RCT across five domains: (1) randomisation process, (2) deviations from intended interventions, (3) missing outcome data, (4) measurement of the outcomes, (5) and selection of the reported results. Two investigators (J-L.P-D. and M.G-B.) independently applied RoB2, and then met to reach consensus on the assessment of bias as “low risk”, “some concerns”, or “high risk”.¹⁹⁷

6. Data collection process and analysis of data

Data extraction from the selected articles was conducted by one reviewer (J-L.P-D.) using pre-established tables. The extraction covered study information, number of participants per group, demographics, intervention details (duration, frequency, volume in early and late intervention stages, and supervision), pre-and post-intervention results, adherence, study retention, and adverse events.

Whenever possible, a meta-analysis was performed using RevMan web review manager (version 8.1.1.), considering two subgroups: PIFM exercises and MRS. For each outcome, a meta-analysis was conducted if two studies were available in a subgroup. A random effect model utilizing standardized mean differences (SMD) was applied to analyse post-intervention means and standard deviations. In case of missing means and standard deviation, they were calculated from medians and interquartile ranges or maximum and minimum values using the formulas provided by Wan et al. (2014).¹⁹⁸ Standard deviation was also calculated from standard error of the mean using the following formula (Equation 3):

$$\text{Equation 3: } SD_j = SEM \times \sqrt{n_j},$$

Where SD_j is the standard deviation of the group, SEM is the standard error of the mean, and n_j is the sample size of the group.

Subgroup SMD values with 95% confidence intervals (CI) and measures of heterogeneity (I^2) were reported.¹⁹⁹ The meta-analysis was non-significant if the CI of its SMD contained zero. For each meta-analysis, the certainty of the evidence was assessed using the Grading of Recommendations Assessment, Development, and Evaluation (GRADE) methodology.²⁰⁰ Because only RCT were included, the grading started at “high certainty of the evidence”. It was then downgraded to moderate, low and very low certainty if serious (one level) or very serious (two levels) concerns were raised for the following criteria: (1) Risk of bias, i.e. one or more studies with two domains of RoB2 tool scored at “high risk”, (2) imprecision, i.e. low sample size if there were less than 400 participants per group and/or contradictory findings under the CI (e.g. evidence of harm and evidence of benefits, with very small effect sizes regarded as “no effect”), (3) inconsistency, i.e. I^2 higher than 50% was regarded as “substantial”,¹⁹⁹ (4) indirectness, i.e. only a restricted part of the research question was addressed, and (5) publication bias was investigated by funnel plots when at least 10 RCT were available.²⁰⁰

A synthesis of results (systematic review) was performed for data that could not be combined in the meta-analysis. Therefore, SMD were determined following Lakens' (2013) guidelines. Specifically, SMD values were calculated using Hedge's g_s (Equation 4), which is based on Cohen's d_s (Equation 5). Hedge's g values are considered more reliable for studies with small sample sizes. The SMD were interpreted as following: < 0.2 very small, < 0.5 small, < 0.8 medium, < 1.2 large, < 2 very large, and ≥ 2 huge.²⁰¹

Equation 4: Hedge's $g = \text{Cohen's } d \times \left(1 - \frac{3}{4 \times (n_{IG} + n_{CG}) - 9}\right)$

Equation 5: Cohen's $d = \frac{\text{Mean}_{IG} - \text{Mean}_{CG}}{\sqrt{\frac{(n_{IG}-1) \times SD_{IG}^2 + (n_{CG}-1) \times SD_{CG}^2}{n_{IG} + n_{CG} - 2}}}$

Where, for equations 4 and 5, n is the sample size, SD is the standard deviation, IG refers to the intervention group, and CG refers to the control group.

Results

1. Study selection

A total of 2485 references were identified through the databases, with 5 additional articles found among the reference lists (Figure 15).²⁰²⁻²⁰⁶ After removing 668 duplicates, the title and abstract screening identified 89 articles for full-text review. Three articles were not retrieved,^{193, 207-208} and one article was withdrawn between identification and screening.²⁰⁹ Ultimately, 28 RCT were included in the systematic review, with 9 studies included in the meta-analyses for PIFM sizes, 9 for strength, and 10 for biomechanic parameters.

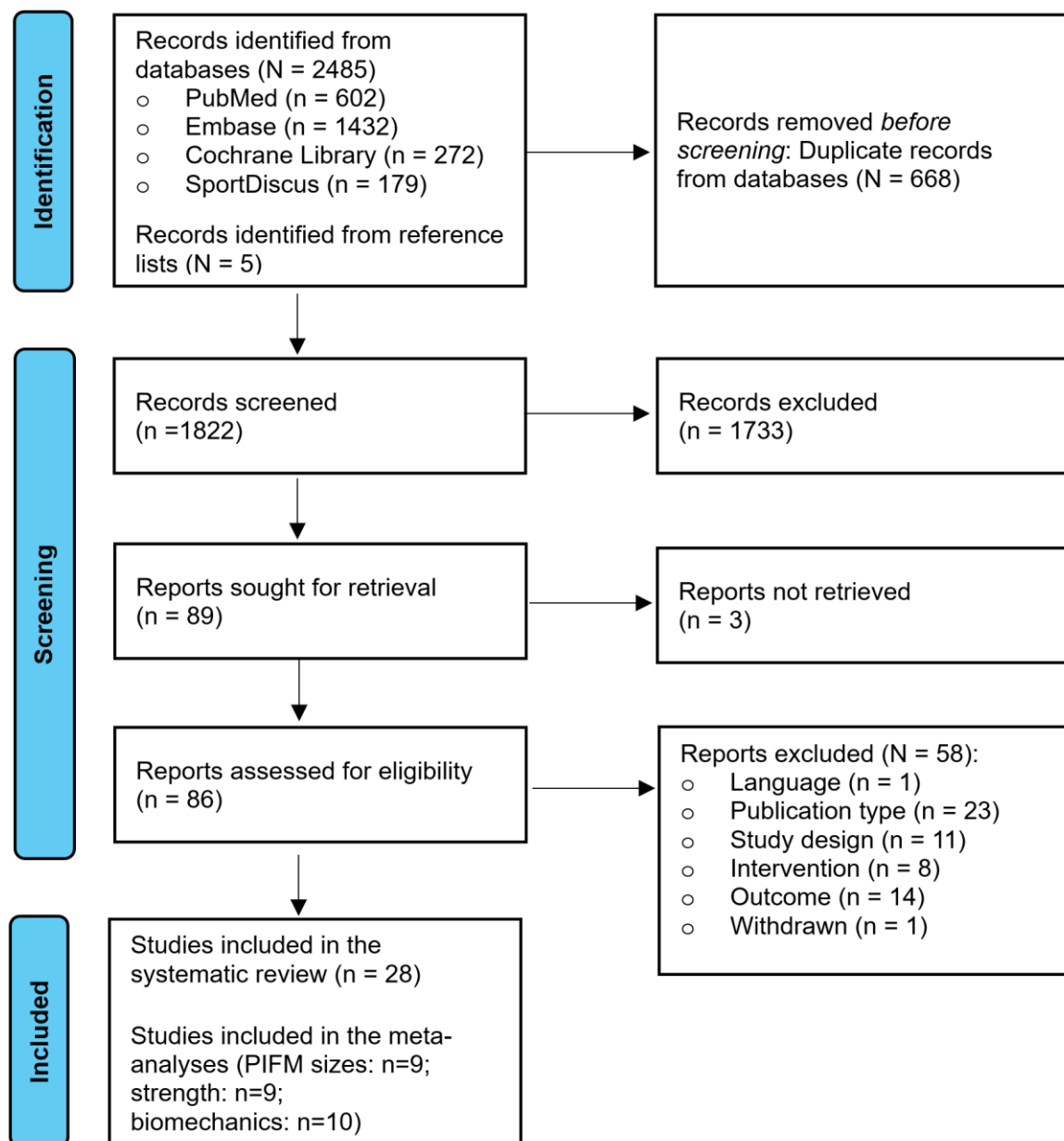


Figure 15. PRISMA flow chart. N (sum of the number of articles), n (number of articles per category), PIFM (plantar intrinsic foot muscles).

2. Risk of bias

The complete risk of bias assessment can be found in Appendix 1.B. In total, 5 studies were classified as low risk, 3 as some concerns, and the remaining 20 as high risk of bias.

3. Characteristics of the studies

The characteristics of each study are summarized in Table 2. This systematic review includes the results from 1399 participants. Some studies did not provide enough information regarding the biological sex of participants.²¹⁰⁻²¹⁵ Nevertheless, there was relative gender parity, with approximately half men and half women (male participation between 640 and 716 individuals).

Twenty-one studies implemented PIFM exercises. Suda et al. (2024) performed a subgroup analysis on the sample included in Matias et al.'s study.^{191, 216} Six studies involved symptomatic participants.^{194, 203, 217-220} Five studies included participants over the age of 60.^{203, 205, 219-221} The intervention durations ranged from 1 and 6 months. The differences between the intervention and control groups were seen in the performance of PIFM exercises,^{192, 194, 202, 213, 217-218, 222} including the short foot exercise carried out during wall squat,¹⁹² as well as PIFM exercises combined with various interventions such as exercises targeting the extrinsic foot muscles,^{184, 191, 205, 210, 216, 219, 223-224} physiotherapy oriented towards low-back pain,²¹⁸ and optimized podiatric treatment.²⁰³ Adjuvant methods such as NMES, transcutaneous electrical nerve stimulation,^{221, 225} electromyographic biofeedback, and electromyographically-controlled electrostimulation were used during the PIFM exercises,^{220, 226} with NMES and transcranial direct current stimulation also being used as stand-alone technique.^{211, 224}

Besides, nine studies utilized MRS during running,^{183, 204, 206, 211-212, 214, 227} or daily life activities.^{184, 215} Three of these studies combined MRS during running along with exercises targeting the PIFMs and extrinsic foot muscles,^{204, 206, 227} while one study included both a MRS group and a PIFM exercise group.¹⁸⁴ In one study, children aged 11 years old wore MRS during school hours.²¹⁵ The mean age of participants in the other eight studies ranged between 20 and 40 years old. The MRS interventions lasted between 1.5 and 9 months.

Adherence to the interventions was often high, but many RCT did not adequately report this outcome. Reported retention rates varied between 74 and 100%. Adverse events included injuries and overloading symptoms. For PIFM exercise studies, adverse events were reported in 5 participants of intervention groups versus 13 participants of control groups. For MRS studies, the adverse events affected 23 participants of intervention versus 6 participants of control groups. For more details on supervision, adherence, drop out rates, retention, and adverse events, see Appendix 1.C.

4. Summary of results

The results related to PIFM sizes and strength are summarized in Table 3. The results related to biomechanics are available in Appendix 1.D. In both tables, data used in the meta-analyses are highlighted using “MA” in superscript.

Table 2: Study population and intervention characteristics

	Study (Author, year)	Study sample size (N) Participant characteristics	Intervention group Group size (M:F ratio); mean age (SD) Intervention characteristics	Control group Group size (M:F ratio); mean age (SD) Intervention characteristics	Duration x frequency or volume Volume: Sets x reps (initial → ... → final)
MRS during running or daily life	Chen et al. (2016)	N=38 Runners (≥20km/w) Asymptomatic	n=20 (11M:9F); 34.55 (6.62) Y Transition to MRS (Vibram® Five-fingers; 92% index) + calf strengthening + single-leg balance + rapid jogging in place	n=18 (11M:7F); 35 (5.39) Y Maintain running with habitual running shoes + calf strengthening + single-leg balance + rapid jogging in place	6Mo x see below. MRS: 2 to 3d/w → <u>5d/w</u> (according to symptoms). 30min run/w + 2h walk/w → <u>1h50 run/w</u> . Calf strengthening: 3d/w x 2 x 30: Stance → Single leg. Balance: 3d/w x 2: 10 reps lateral movements → <u>5 reps SEBT movements</u> . Jogging in place: 3d/w x 2: 1.5min → <u>5 min</u> .
	Johnson et al. (2016)	N=37 Runners (24 to 48km/w) Asymptomatic	n=18 (10M:8F); approx. 24.4 (6.2) Y Transition to MRS (Vibram® Five-fingers; 92% index)	n=19 (10 to 14M:9 to 5F); approx. 27.7 (6.2) Y Maintain running with habitual running shoes	10w x 24 to 48km/w MRS: 1.6 to 3.2km during w1 → increased by 1.6 to 3.2km/w for 3w → increased as tolerated (final distance ≥ <u>4.8km/w</u>).
	Miller, E et al. (2014)	N=29 (13 to 17M:12 to 16F) Runners (≥48.3km/w) Asymptomatic	n=16 (M:F NA); approx. 30.5 (4.0) Y Transition to MRS (New Balance® Road Minimus 10; 72% index (<i>unsure</i>) or Merrel® Pace glove; 80% index) <i>n=14 were analysed for biomechanical outcomes</i>	n=13 (M:F NA); approx. 29.9 (5.5) Y Maintain running with habitual running shoes <i>n=12 were analysed for biomechanical outcomes</i>	12w x mean weekly distance of 48.3km (approx. 5 sessions/w) MRS: 3.2km during w1 → <u>20.9km during w11</u>
	Quinlan, S et al. (2022)	N=64 Healthy children Asymptomatic	n=30 (15 to 19M:21 to 15F): 11.26 (0.76) Y MRS during school hours (shoe model NA; 46% index)	n=34 (15 to 17M:17 to 19F): 11.28 (0.83) Y Maintain wearing habitual shoes	9Mo x see below. MRS: 3d/w. = approx. 18h/w No progression.
MRS combin	McCarthy et al. (2013)	N=19 (100% F) Recreational runners (>15km/w) Asymptomatic	n=9; 29 (3) Y Transition to MRS (Vibram FiveFingers; 92% index) + running with habitual shoes + PIFM ex.	n=10; 30 (4) Y Maintain running with habitual running shoes	12w x 3/w MRS: walk → walk and run → <u>run</u> (3x5min → <u>3x30 min</u> → <u>individually-adapted</u>)

MRS or PIFM ex.	Shen, B et al. (2022)	N=26 (100% M) Recreational runners (mean approx. 30km/w) Asymptomatic	n=13; 25.2 (4.8) Y Transition to MRS (Vibram® Five-fingers; 92% index) while running with a forefoot strike + PIFM ex. (towel curls, foot doming, TSO) + heel raises + balance board + foot auto-massage	n=13; 23.8 (1.7) Y Maintain running with habitual shoes	12w x see below. Total running: ≥20km/w FFS running in MRS: 3d/w x 5 min/d → <u>48 min/w</u> All ex.: 1d/w (<i>unsure</i>) Towel curls: 3x10 → <u>3x15</u> Doming and TSO: 2x10 → <u>4x25</u> Heel raises: bipedal (1x10 → 2x15) → bipedal on step (2x20 → <u>3x30</u>) + unipedal on step (1x10 → <u>3x15</u>) Balance board: 2x20s → <u>3x40s</u>
	Warne et al. (2017)	N=24 (100% M) Runners (≥40km/w) Asymptomatic	n=12; 36 (7) Y Transition to MRS (Vibram FiveFingers; 72% index) + conventional shoe running (Asics GEL-Columbus; 16% index) + gait retraining (step rate increased by 10%, forefoot strike pattern, upright posture, soft landing) + towel curls + heel raises + single leg balance + ankle mobility	n=12; 34 (9) Y Conventional shoe running (Asics GEL-Columbus; 16% index) + same interventions as IG except MRS.	6w x see below. MRS: 3/w x (5-8 min → 30 - <u>40 min</u>) Exercises: 3 → 2/w x 3x15 reps per ex. Balance: 60s
	Ebrecht et Sichtung (2018)	N=59 (23 to 37M:23 to 37F); approx. 25.3 (7.2) Y Healthy, Asymptomatic	IG1 n=32 (M:F NA): Transition to MRS (Leguan® Aktiv; 88% index) (<10 km·h ⁻¹ =“beginners”; ≥10 km·h ⁻¹ =“advanced”) <i>n=11 were analysed for muscle sizes</i> IG2 n=15 (M:F NA): NMES in sitting without active contraction <i>n=7 were analysed for muscle sizes</i>	n=12 (M:F NA): Maintain habitual running <i>n=5 were analysed for muscle sizes</i>	8w x 2d/w MRS: 10 or 15 min (beginners and advanced runners, respectively). Duration increased (unclear). NMES: 75 reps (15 min). Intensity individually increased (unclear).

Ridge et al. (2019)	N=57 Runners (24.1 to 28.3km/w) Asymptomatic	IG1: n=19 (9M:10F); 22.9 (3.3) Y MRS during daily life (Inov-8® Bare-XF 210; 86% index or Bare-XF 260; 76% index) (not used during running) IG2: n=19 (10M:9F); 22 (1.8) Y Towel curls + TSO + toe squeeze + doming (bipedal and hopping) + heel raises	n=19 (11M:8F); 22.6 (2.7) Y Maintain running with habitual shoes	8w x see below. MRS: 2500 steps/d → 5000 steps/d (w3 and w4) → <u>7500 steps/d</u> . All exercises except doming while hopping: ≥5d/w; 3 sets x 10 to 20 reps → <u>30 reps</u> . Doming while hopping (in place → <u>in square</u>): 3x10 → <u>3x20</u> Heel raises (stance → <u>single leg stance</u> ; flat surface → <u>edge of step</u>)
Day and Michael (2019)	N=20 Elite runners (≥50km/w), Asymptomatic	n=11 (M:F NA); 24 (6) Y SFE + foot curl + eccentric band curl + concentric band curl + range of motion (<i>unclear</i>). <i>n=9 were analysed for biomechanics at 14 km·h⁻¹ and 18 km·h⁻¹, and n=8 were analysed at 22 km·h⁻¹.</i>	n=9 (M:F NA); 30 (12) Y Maintain habitual training <i>n=7 were analysed for biomechanics at 14 km·h⁻¹ and 18 km·h⁻¹, and n=5 were analysed at 22 km·h⁻¹.</i>	10w x 3d/w Range of motion: 30 Each other ex.: 3 x 10 <u>Bands of greater resistance</u>
Elsayed, W et al. (2023)	N=40 Flexible pes planus (ND≥1 cm and forefoot abduction), Symptomatic	n=20 (7M:13F); 26.8 (6.5) Y SFE with towel + shoe insoles	n=20 (5M:15F); 24.9 (4.7) Y Shoe insoles	6w x see below. SFE: 7d/w x 3 x 10 Sit → stance → <u>single leg stance</u> (every 2w) Insoles: 8h/d
Goldmann et al. (2013)	N=27 (100% M) Healthy Asymptomatic	n=15; 24 (4) Y Isometric metatarsophalangeal joint flexion contractions	n=12; 26 (2) Y No intervention	7w x 4/w IG: 4 sets x 5 reps (3s contraction – 3s relaxation) at 90% of maximal voluntary contraction
Hartmann et al. (2009)	N=56 Older adults Asymptomatic (<i>probably</i>)	n=28 (10M:18F); 75.9 (5.5) Y In-person and home-sessions: Same as CG + moving toes + spreading toes + grabbing small objects + calf stretching.	n=28 (11M:17F); 75.9 (5.4) Y In-person: Warm-up + 15 min aerobic + resistance training (knee and hip) + stretching	12w x see below. In-person session: 2/w Resistance training: 2-3 sets x 12 reps Home-session: 7/w x 10 min Stretching: 2 x 30 s per stretch

Jung et al. (2011)	N=28 (<i>gender NA</i>) Pes planus ($\geq 4^\circ$ calcaneal eversion while standing; ND>1.3cm) Asymptomatic	n=14 (M:F NA); 22.36 (2.41) Y SFE + foot orthoses	n=14 (M:F NA); 21.93 (2.73) Y Foot orthoses	8w x see below. SFE (single leg stance): 7d/w x 2/d x 3 sets x <i>unknown</i> reps of 5s (per foot) → <u>5 reps of 10s</u> Foot orthoses: during outdoor walking (duration NA)
Kim, H et al. (2022)	N=48 Patello-femoral pain syndrome (>3Mo) and $\geq 6^\circ$ calcaneal eversion at rest), Symptomatic	IG1 $\bar{n}$ =16 (7M:9F); 21.81 (2.29) Y SFE IG2 $\bar{n}$ =16 (8M:8F); 22.69 (2.77) Y SFE + talonavicular joint mobilization	$\bar{n}$ =16 (6M:10F); 21.38 (1.82) Y Talonavicular joint mobilization	4w x 3d/w SFE: 2 x 5 min (5s per contraction). Sit → <u>stance</u> (from w3) Mobilization: 2 x 5 min (frequency 1/s)
Kim, M et al. (2015)	N=24 (13M:11F) Hallux valgus (>15°) Symptomatic status NA	$\bar{n}$ =12 (M:F NA); 22.2 (2.04) Y TSO + orthoses	$\bar{n}$ =12 (M:F NA); 22.8 (2.83) Y Orthoses	8w x see below. TSO: 4d/w x 20 min/d Orthoses: >8h/d
Mahmoud, W (2022)	N=46 (100% M) Low back pain + flexible pes planus (FPI $\geq$ 10 and ND>1cm) Symptomatic	IG1 : n=15; 50.28 (8.28) Y SFE IG2 : n=17; 47.42 (7.10) Y SFE + walking + stretching (hamstring, calf, plantar fascia) + TENS (lumbar) + massages (lumbar)	n=14; 47.64 (7.76) Y MTP joint and ankle DF/PF without load	6w x see below. SFE: 7d/w x 30 x 30s (10s rest). Sit → <u>stance</u> and <u>half-squat</u> (from w4). Walking: 2d/w x 20 min. Stretching: 2d/w x 5 x 60s. TENS: 3d/w x 20 min (intensity according to tolerance). Massages: 3d/w x 15 min. CG: 1/d x 5 sets (for ankle and MTP joints) x 30s.
Matias et al. (2022)	N=87 Runners (20 to 100km/w) Asymptomatic	n=41 (17M:24F); 40.3 (7.7) Y (A) In-person: PIFM ex. (toes abd-adduction, SFE, plantar arch raises, ball grabbing, squeeze toe separators) + heel raises + ex. with rubber bands (abduction, toes and ankle flexion) + ex. sitting (inv- and eversions, foot taps, toe mobilisation, plantar self-massage, ball slide. (B) At-home: slight differences.	n=46 (25M:19F); 40.3 (6.1) Y Warm up and static stretching before each running practice	8w x 4d/w PIFM ex: total=50 reps; approx. 15 min. Sit → stance → <u>single leg stance</u> Heel raises: 1 x 30 → 2 x 30 → <u>2 x 40 reps</u> . Ex. with rubber band: 10 per ex. Ex. sitting: approx. 4min CG: 5 minutes Volume and difficulty increased each w. Further details: see article protocol.

Mickle et al. (2016)	N=68 Older people Both symptomatic and asymptomatic	n=36 (8M:28F); 70.6 (5.6) Y In-person: Warm-up + short-foot + heel raises + ex. against band (toe flexion, hallux flexion, hallux adduction, ankle dorsiflexion, inversion and eversion) + calf and plantar fascia stretching. <i>n=16 were analysed for muscle sizes</i>	n=32 (11M:21F); 69.5 (6.0) Y Home-based: Warm-up + ex. without resistance: toe squeezes, toe pulls, marble pickups, towel pulls and short foot exercise <i>n=13 were analysed for muscle sizes</i>	12w x 3/w IG: 45 min Short foot: 3 x 5 reps x 5 s → <u>5 x 5 reps x 10s</u> Other ex.: 1 x 10 reps → <u>2 to 3 x 10 reps</u> CG: no progression.
Namsawang, J et al. (2019)	N=36 Flexible pes planus (criteria NA) Asymptomatic	n=18 (8M:10F); 20.17 (1.20) Y SFE + NMES	n=18 (4M:14F); 19.78 (1.59) Y SFE + “sham” NMES	4w x 3d/w SFE: 30 x 5s (sitting). NMES: 30min (maximal intensity without discomfort) during stance.
Namsawang, J. et al. (2022)	N=68 Sedentary older adults (age criteria NA) Asymptomatic	n=34 (6M:28F); 68.6 (4.6) Y SFE + TENS	n=34 (6M:28F); 68.7 (4.0) Y SFE + sham TENS	4w x 3d/w SFE: 30 x 5sec during stance. TENS (standing): 30 min (maximal intensity without discomfort) No progression.
Okamura et al. (2020)	N=17 Older adults (≥65 Y) Both symptomatic and asymptomatic	n=9 (9M:0F); 79.1 (4.4) Y SFE + TSO with EMG biofeedback	n=8 (7M:1F); 79.6 (4.2) Y SFE + TSO	6w x 2d/w SFE: 3 x 10 x 5s TSO: 40 x 3 to 5s SFE and TSO: Sit → <u>Stance</u>
Okamura et al. (2020b)	N=20 Pes planus (FPI≥6) Asymptomatic	n=10 (1M:9F); 19.7 (0.9) Y SFE + EMG biofeedback + EMG-controlled electrostimulation	n=10 (2M:8F); 20.2 (1.5) Y No intervention	8w x 3d/w 3 x 10 x 5s Sit → stance → <u>single leg stance</u>
Spink et al. (2011)	N=305 Older adults with disabling foot pain and increased risk of falling Symptomatic	n=153 (47M:106F); 74.2 (6.0) Y Foot orthoses + provision of footwear + education in falls prevention + home-based ex.: toe flexion strength, ankle ROM, ankle strength (eversion, inversion, dorsiflexion, heel raises), calf and hallux stretching.	n=152 (47M:105F); 73.6 (5.7) Y Continue current podiatry treatment	Total duration is 12Mo but outcomes were assessed at 6Mo x see below. Home-based ex.: 3/w x 30 min (<u>increase reps and resistance</u>) Strengthening ex.: 3x10 Stretching: 2-3 x 20s

Suda et al. (2024)	This is a secondary analysis of the data of Matias et al. (2022). We refer the reader to the description of this article upper in this table.			
Taddei, U et al. (2020)	N=28 Recreational runners (20 to 100km/w) Asymptomatic	IG: n=14 (5M:9F); 41.9 (7.4) Y; CG: n=14 (9M:5F); 41.6 (6.0) Y This is a proof-of-concept article using the same interventions as Matias et al. (2022). We refer the reader to the description of this article upper in this table.		
Utsahachant et al. (2023)	N=45 Flexible pes planus (ND>10mm) Asymptomatic	IG1: n=15 (5M:10F); Median (IQR): 21.0 (21; 29) Y SFE IG2: n=15 (5M:10F); Median (IQR): 21.0 (20; 22) Y SFE during wall squat	n=15 (5M:10F); Median (IQR): 21.0 (21; 26) Y No intervention	6w x 7d/w IG1: 3→4 sets x 10 reps x 3→5s (sit → stance → single leg stance) IG2: 3→4 sets x 10 reps x 3→5s (sit → wall squat → unilateral wall squat)
Xiao et al. (2022)	N=36 (100% M) Healthy sedentary adults Asymptomatic	IG1: n=12; 22.7 (2.0) Y SFE + TSO with toe squeeze + towel curls + balance board training. IG2: n=12; 21.9 (2.1) Y tDCS	n=12; 23.5 (1.5) Y No intervention	4w x 3d/w SFE and TSO: 2 x 10 → 3 x 20 Towel curls: 3 x 10 → 3 x 15 with 0.5kg additional weight Balance board: 2 x 20s → 3 x 30s tDCS: 20 min

Legend: /w (per week), approx. (approximately), CG (control group), d (day), DF (dorsiflexion), EMG (electromyography), ex. (exercise), F (female), FFS (fore foot strike), FPI (foot posture index), IG (intervention group), IQR (interquartile range), M (male), min (minutes), Mo (month), MRS (minimalist running shoes), MTP (metatarsophalangeal), N (sample size included in the analyses), n (sample size of the group), NA (non-available), ND (navicular drop), NMES (neuromuscular electrostimulation), PF (plantarflexion), PIFM (plantar intrinsic foot muscle), SD (standard deviation), SEBT (star excursion balance test), SFE (short foot exercise), tDCS (transcranial direct current stimulation), TENS (transcutaneous electrical nerve stimulation), TSO (toe spread out exercise), w (week), w1 (first week after the beginning of the study), and Y (year).

Table 3: Study results regarding foot muscle sizes and foot strength

	Outcome domain	Study Instrument	Outcome (units)	Intervention group Mean (SD), within-group p-value	Control group Mean (SD), within-group p-value	Between group p-value Effect size
MRS during running or daily life	PIFM sizes	Chen et al. (2016) <i>MRI</i>	Total foot muscle volume (cm ³)	Pre: 4566.96 (753.51) Post: 4973.51 (766.5); p<0.01*	Pre: 4613.94 (723.98) Post: 4739.53 (914.34); p=0.38	p: NA, g=0.27
			Rearfoot muscle volume (cm ³)	Pre: 2592.21 (630.42) Post: 2764.32 (531.33); p=0.10	Pre: 2526.27 (485.81) Post: 2647.36 (660.84); p=0.33	p: NA, g=0.19
			Forefoot muscle volume (cm ³)	Pre: 1974.27 (303.23) Post: 2209.64 (417.21); p<0.01*	Pre: 2087.66 (333.13) Post: 2092.16 (330.48); p=0.95	p: NA, g=0.30
		Johnson et al. (2016) <i>USI</i>	AbH CSA (cm ²) ^{MA}	Pre: 2.29 (0.45) Post: 2.53 (0.40); p: NA	Pre: 2.22 (0.55) Post: 2.26 (0.48); p: NA	p=0.01* (IG>CG), g=0.60
			FDB CSA (cm ²) ^{MA}	Pre: 2.04 (0.37) Post: 2.04 (0.39); g=0	Pre: 2.12 (0.56) Post: 2.03 (0.49)	p=0.22, g=0.02
			FHB thickness (cm) ^{MA}	Pre: 1.01 (0.17) Post: 1.00 (0.18)	Pre: 1.07 (0.22) Post: 1.01 (0.22)	p=0.08, g=-0.05
		Quinlan, S et al. (2022) <i>USI</i>	AbH CSA (% of foot length) ^{MA}	Pre: 6.53 (2.05) Post: 7.32 (2.35)	Pre: 6.29 (1.92) Post: 6.77 (1.53)	p=0.1, g=0.28
			FDB CSA (% of foot length) ^{MA}	Pre: 6.42 (1.01) Post: 6.82 (1.05)	Pre: 6.27 (1.37) Post: 6.36 (1.23)	p=0.09, g=0.40
	Foot strength	Quinlan, S et al. (2022) <i>Fixed hand-held dynamometer</i>	Hallux flexion (% of BMI) ^{MA}	Pre: 2.33 (1.06) Post: 2.58 (0.86)	Pre: 2.15 (0.99) Post: 2.20 (0.72)	p=0.10, g=0.48
			Toes 2-to-5 flexion (% of BMI) ^{MA}	Pre: 2.18 (1.01) Post: 2.26 (0.75)	Pre: 1.91 (0.83) Post: 1.85 (0.76)	p=0.10, g=0.54
MRS + PIFM ex.	Foot strength	Shen, B et al. (2022) <i>Dynamometer, except MTPJ flexion with force sensors</i>	Hallux flexion (N·kg ⁻¹) ^{MA}	Pre: 1.43 (0.39) Post: 1.74 (0.44); p=0.001*	Pre: 1.67 (0.44) Post: 1.7 (0.5); p=0.762	p=0.036* , g=0.08
			Toes 2-to-5 flexion (N·kg ⁻¹) ^{MA}	Pre: 0.94 (0.21) Post: 1.29 (0.43)	Pre: 0.98 (0.34) Post: 1.05 (0.46)	p>0.05, g=0.52
			MTP joint flexion (all toes) (N·kg ⁻¹)	Pre: 1.17 (0.39) Post: 1.55 (0.4); p=0.006*	Pre: 1.58 (0.6) Post: 1.54 (0.6); p=0.813	p NA, g=0.02

MRS or PIFM ex.	PIFM sizes	Ebrecht et al. (2018) <i>USI</i>	AbH CSA (cm ²)	<i>Pre-post changes (median and IQR) ^λ:</i> IG1: 0.42 (0.53; 0.87); IG2: 0.32 (0.35; 0.56)	<i>Pre-post changes (median and IQR) ^λ:</i> 0.36 (0.32; 0.57)	p=0.203, g NA
		Ridge et al. (2019) <i>USI</i>	AbH CSA (cm ²) ^{MA}	IG1: Pre: 2.08 (0.61); Post: 2.22 (0.54); p<0.05* IG2: Pre: 1.98 (0.57); Post: 2.19 (0.55); p<0.05*	Pre: 2.22 (0.45) Post: 2.20 (0.42); p=0.704	p<0.001*, g=0.04 (IG1>CG) and p<0.05*, g=-0.02 (IG2>CG)
			FHB thickness (cm) ^{MA}	IG1: Pre: 1.56 (0.20); Post: 1.61 (0.20); p<0.05* IG2: pre: 1.33 (0.22); Post: 1.52 (0.23); p<0.05*	Pre: 1.60 (0.23) Post: 1.60 (0.22); p=0.516	p<0.05* , g=0.05 (IG1>CG) and p<0.05* , g=-0.35 (IG2>CG)
			FDB CSA (cm ²) ^{MA}	IG1: pre: 2.06 (0.35); Post: 2.17 (0.35); p<0.05* IG2: pre: 1.83 (0.46); Post: 2.04 (0.50); p<0.05*	Pre: 2.05 (0.22) Post: 2.04 (0.23); p=0.687	p<0.05* , g=0.43 (IG1>CG and), p<0.05* , g=0 (IG2>CG)
			QP CSA (cm ²)	IG1: pre: 1.84 (0.41); Post: 2.02 (0.46); p<0.05* IG2: pre: 1.70 (0.35); Post: 1.95 (0.33); p<0.05*	Pre: 1.76 (0.45) Post: 1.75 (0.41); p=0.875	p<0.05* , g=0.61 (IG1>CG) and p<0.05* , g=0.53 (IG2>CG)
	Foot strength	Ridge et al. (2019) <i>Dynamometer</i>	Hallux flexion (kg) ^{MA}	IG1 pre: 4.68 (0.97); Post: 6.98 (2.35); p<0.05* IG2: pre: 5.27 (2.34); Post: 7.14 (3.99); p<0.05*	Pre: 6.03 (2.90) Post: 6.38 (2.35); p=0.483	p<0.05* , g=0.25 (IG1>CG) and p<0.05* , g=0.23 (IG2>CG)
			Toes 2 to 5 flexion (kg) ^{MA}	IG1: pre: 4.62 (1.91); Post: 5.70 (1.80); p<0.05* IG2: pre: 5.13 (2.27); Post: 6.85 (3.22); p<0.05*	Pre: 5.81 (2.33) Post: 5.87 (2.27); p=0.477	p<0.05* , g=-0.08 (IG1>CG) and p<0.05* , g=0.34 (IG2>CG)
			Doming (kg)	IG1: 6.28 (3.03); Post: 9.47 (4.16); p<0.05* IG2: pre: 4.98 (2.10); Post: 10.28 (4.31); p<0.05*	Pre: 7.53 (2.26) Post: 8.14 (3.31); p=0.082	p<0.05* , g=0.35 (IG1>CG) and p<0.05* , g=0.55 (IG2>CG)

PIFM sizes	Jung et al. (2011) <i>USI</i>	AbH CSA (cm ²) ^{MA}	Pre: 2.36 (0.66) Post: 2.63 (0.65); p<0.001*	Pre: 2.49 (0.41) Post: 2.62 (0.44); p<0.001*	p= 0.008* ; g=0.02
	Kim, M et al. (2015) <i>USI</i>	AbH CSA (cm ²) ^{MA}	Pre: 2.04 (0.43) Post: 2.52 (0.66); p<0.05*	Pre: 2.04 (0.62) Post: 1.93 (0.51); p>0.05	p<0.05* ; g=0.97
	Mahmoud, W (2022) <i>USI</i>	AbH CSA (mm ²) ^{MA}	IG1^{MA} : Pre: 216.63 (17.15); Post: 233.5 (11.59); p=0.001* IG2 : Pre: 220.79 (21.68); Post: 256.75 (17.59); p<0.001*	Pre: 211.55 (21.5) Post: 212.45 (20.33); p=0.216	p=0.038* , g=1.25 (IG1>CG) p<0.001* , g=2.29 (IG2>CG)
	Mickle et al. (2016) <i>MRI</i>	AbH CSA (mm ²)	Pre: 13.5 (1.0) Post: 15.1 (1.0); p=0.27	Pre: 13.9 (0.6) Post: 12.8 (0.6); p=0.2	p=0.0016* , g=2.26
		FDB CSA (mm ²)	IG: Pre: 14.9 (1.0) Post: 16.4 (1.1); p=0.3	Pre: 12.1 (0.6) Post: 13.7 (0.7); p=0.08	p=0.9881, g=2.78
		AbDM CSA (mm ²)	Pre: 10.5 (0.5) Post: 12.4 (0.5); p=0.007*	Pre: 11.4 (0.6) Post: 11.3 (0.7); p=0.9	p=0.001* , g=1.79
		AbH muscle volume (mm ³) ^{MA}	Pre: 688.5 (61.0) Post: 838.4 (62.8); p=0.1	Pre: 711.3 (39.3) Post: 729.3 (42.5); p=0.76	p=0.0011* , g=1.94
		FDB muscle volume (mm ³) ^{MA}	Pre: 125.1 (8.2) Post: 152.3 (8.5); p=0.03*	Pre: 111.1 (6.4) Post: 133.8 (6.9); p=0.03*	p=0.5542, g=2.30
		AbDM muscle volume (mm ³)	Pre: 398.3 (24.6) Post: 496.8 (25.3); p=0.009*	Pre: 429.1 (26.2) Post: 441.3 (31.0); p=0.77	p=0.0256* , g=1.93
	Namsawang, J et al. (2019) <i>USI</i>	AbH CSA (mm ²)	Pre: 218.66 (53.23) Post: 256.96 (NA); p<0.05*	Pre: 225.97 (64.10) Post: 244.36 (NA); p>0.05	p>0.05, g=0.21

Namsawang, J. et al. (2022) <i>USI</i>	AbH CSA (cm ²)	<i>Median (IQR)</i> Pre: 1.86 (1.61 – 2.31) Post: 2.05 (1.76 – 2.53); p<0.05*	<i>Median (IQR)</i> Pre: 1.79 (1.56 – 2.20) Post: 2.16 (1.77 – 2.50); p<0.05*	p=0.49, g=-0.05
Okamura et al. (2020b) <i>USI</i>	AbH thickness (mm) ^{MA}	Pre: 12.3 (2.4) Post: 13.4 (3.7)	Pre: 12.2 (1.7) Post: 13.2 (2.1)	p>0.05, g=0.06
	FDB thickness (mm) ^{MA}	Pre: 7.3 (1.0) Post: 7.1 (1.1)	Pre: 7.7 (1.7) Post: 7.6 (1.6)	p>0.05, g=-0.35
	FHB thickness (mm) ^{MA}	Pre: 13.9 (1.6) Post: 14.0 (1.3)	Pre: 14.3 (1.4) Post: 14.7 (1.1)	p>0.05, g=-0.56
Taddei, U et al. (2020) <i>MRI</i>	AbH CSA (mm ²)	Pre: 135.8 (64.4) Post: 141.2 (58.5)	Pre: 157.7 (79.8) Post: 145.2 (64.6)	p=0.1, g=-0.06
	AbDM CSA (mm ²)	Pre: 105.4 (36.1) Post: 112.8 (31.4)	Pre: 119.3 (33.7) Post: 113.9 (31.7)	p=0.256, g=-0.03
	FDB CSA (mm ²)	Pre: 116.9 (45.5) Post: 116.7 (30.8)	Pre: 128.3 (44.4) Post: 122.0 (44.1)	p=0.169, g=-0.14
	FHB CSA (mm ²)	Pre: 143.3 (38.9) Post: 147.2 (42.3)	Pre: 174.9 (37.6) Post: 166.2 (52.2)	p=0.053, g=-0.39
	AbH volume (cm ³) ^{MA}	Pre: 11.0 (6.1) Post: 13.7 (6.3); p NA	Pre: 15.4 (8.1) Post: 13.3 (6.3); p NA	p=0.019* , g=0.06
	AbDM volume (cm ³)	Pre: 9.2 (5.0) Post: 11.2 (4.8); p NA	Pre: 12.5 (4.6) Post: 12.0 (4.6); p NA	p=0.013* , g=-0.17
	FDB volume (cm ³) ^{MA}	Pre: 10.4 (4.6) Post: 12.5 (4.7); p NA	Pre: 14.0 (4.6) Post: 13.3 (4.3); p NA	p=0.039* , g=-0.17
	FHB volume (cm ³) ^{MA}	Pre: 9.1 (4.8) Post: 10.2 (4.4); p NA	Pre: 12.6 (3.6) Post: 12.0 (4.1); p NA	p=0.009* , g=-0.41

Foot strength	Day and Michael (2019) <i>Biodex dynamometer</i>	5-toes flexion (N·kg ⁻¹) ^{MA}	Pre ^λ : 2.6 (0.6) Post ^λ : 3.3 (0.9); p=0.006*	Pre ^λ : 2.6 (0.6) Post ^λ : 2.6 (0.9); p=0.607	p=0.002* , g=0.74
	Goldmann et al. (2013) <i>Dynamometer</i>	MTP joint flexion moment (all toes) (Nm·kg ⁻¹), right side ^{MA}	Pre: 0.24 (0.06) Post: 0.40 (0.08); p<0.001*	Pre: 0.25 (0.05) Post: 0.26 (0.06); p=0.11	p<0.001* , g=1.86
		MTP joint flexion moment (all toes) (Nm·kg ⁻¹), left side	Pre: 0.21 (0.06) Post: 0.38 (0.07); p<0.001*	Pre: 0.24 (0.04) Post: 0.25 (0.05); p=0.12	p<0.001* , g=2.03
	Jung et al. (2011) <i>Fixed dynamometer</i>	Hallux flexion (kgf) ^{MA}	Pre: 6.35 (2.98) Post: 8.14 (3.17); p<0.001*	Pre: 6.38 (3.53) Post: 7.26 (3.24); p<0.001*	p=0.008* , g=0.27
	Mickle et al. (2016) <i>Pressure platform</i>	Hallux flexion (% BW)	Pre ^λ : 9.1 (4.3) Post ^λ : 10.9 (4.9); p<0.001*	Pre ^λ : 9.3 (5.0) Post ^λ : 9.3 (3.1); p=0.92	p NA, g=0.38
		Toes 2 to 5 flexion (% BW)	Pre ^λ : 5.3 (2.5) Post ^λ : 7.3 (2.9); p<0.05*	Pre ^λ : 7.1 (4.5) Post ^λ : 6.9 (2.4); p>0.05	p NA, g=0.15
	Okamura et al. (2020) <i>Dynamometer</i>	5-toes flexion (grip), seated, dominant (N)	Pre: 12.2 (4.2) Post: 13.8 (3.8)	Pre: 13.1 (4.5) Post: 14.4 (5.4)	p>0.05, g=-0.12
		5-toes flexion (grip), seated, non-dominant (N)	Pre: 9.8 (4.3) Post: 13.1 (4.9); p=0.0008*	Pre: 9.6 (4.6) Post: 10.7 (4.9); p>0.017	p=0.029* , g=0.46
	Spink et al. (2011) <i>Hand-held dynamometer</i>	Hallux flexion (N) ^{MA}	Pre: 68.23 (26.97) Post: 69.69 (30.15); p NA	Pre: 68.95 (27.98) Post: 69.86 (28.16); p NA	p=0.85; g=0.0
		“Lesser toe” flexion (N) ^{MA}	Pre: 68.00 (25.06) Post: 74.11 (30.04); p NA	Pre: 67.95 (27.18) Post: 70.13 (27.11); p NA	p=0.13; g=0.14
	Taddei, U et al. (2020) <i>Pressure platform</i>	Isometric 5-toes flexion (% BW)	Pre: 19.95 (6.24) Post: 27.31 (9.33)	Pre: 23.46 (4.86) Post: 25.61 (7.4)	p=0.707, g=0.20
	Xiao et al. (2022) <i>Dynamometer</i>	MTP joint flexion (all toes) (N·kg ⁻¹) ^{MA}	IG1 ^{MA} : pre: 3.03 (0.87); Post: 3.37 (1.05) IG2 : pre: 3.63 (1.21); Post: 4.17 (1.71)	Pre: 2.91 (0.99) Post: 2.88 (0.95)	p=0.027* , g=0.47 (IG1>CG); p=0.008* , g=1.49 (IG2 > CG)
		Toes 2 to 5 flexion (N·kg ⁻¹) ^{MA}	IG1 ^{MA} : pre: 1.76 (0.70); Post: 1.99 (0.79) IG2 : pre: 1.76 (0.49); Post: 1.90 (0.52)	Pre: 1.72 (0.72); Post: 1.74 (0.74)	p>0.05, g=0.32 (IG1 vs CG) and p=0.015* , g=0.24 (IG2 > CG)

Legend: AbDM (abductor digiti minimi), AbH (abductor hallucis), BMI (body mass index), **Bold *** (significant p-value at p<0.05 or lower if it results from a post-hoc test), CG (control group), CSA (cross-sectional area), ex. (exercise), FDB (flexor digitorum brevis), FHB (flexor hallucis brevis), g (Hedge's g effect size), IG (intervention group), IQR (interquartile range), kgf (kilogram force), ^{MA} (data included in the meta-analysis), MRI (magnetic resonance imaging), MRS (minimalist running shoes), MTP (metatarsophalangeal), NA (non-available), p (p-value), PIFM (plantar intrinsic foot muscle), Post (post-intervention), Pre (pre-intervention), QP (quadratus plantae), SD (standard deviation), USI (ultrasound imaging), and ^λ (data was estimated from graph).

4.1. Plantar intrinsic foot muscle sizes

Meta-analyses were conducted for abductor hallucis, flexor digitorum brevis, and flexor hallucis brevis muscles (Figure 16). Therefore, volume, cross-sectional area (CSA), and thickness were used depending on availability in the articles. The CI crossed zero for abductor hallucis, with a SMD of 0.37 for PIFM exercises ($[-0.07, 0.82]$, $I^2 = 47\%$, $p = 0.09$) and 0.66 for MRS ($[-0.03, 1.34]$, $I^2 = 75\%$, $p = 0.06$). Similarly, there were no significant differences for flexor digitorum brevis between interventions and control (PIFM exercises: SMD = $-0.16 [-0.61, 0.30]$, $I^2 = 0\%$, $p = 0.50$ and MRS: SMD = $0.71 [-0.08, 1.50]$, $I^2 = 81\%$, $p = 0.08$), as well as flexor hallucis brevis (PIFM exercises: SMD = $-0.43 [-0.89, 0.04]$, $I^2 = 0\%$, $p = 0.07$ and MRS: SMD = $-0.01 [-0.50, 0.48]$, $I^2 = 0\%$, $p = 0.97$).

The synthesis of data that could not be combined into the meta-analyses revealed the following: PIFM exercises did not result in increased CSA in the abductor hallucis, abductor digiti minimi, flexor digitorum brevis and flexor hallucis brevis muscles compared to a warm-up and stretching routine before running sessions.²²³ The effect sizes for quadratus plantae CSA were moderate in favour of PIFM exercises in runners (SMD = 0.53, $p < 0.05$).¹⁸⁴ Individuals with low back pain experienced a significant increase in abductor hallucis CSA after PIFM exercises combined with other non-invasive interventions to the lumbar spine, compared to exercises with no expected effects (SMD = 2.29, $p < 0.001$).²¹⁸ Adding NMES or transcutaneous electrical nerve stimulation to PIFM exercises did not result in significant differences in abductor hallucis CSA compared to exercises with sham-NMES or sham-transcutaneous electrical nerve stimulation.^{221, 225} The same was true for NMES alone compared to no intervention.²¹¹ Regarding MRS, a six-month transition to MRS used during running combined with extrinsic foot muscle exercises did not lead to significantly increased total and forefoot muscle sizes compared to the group that ran in traditional footwear.¹⁸³ Similarly, an eight-week transition to MRS running did not significantly increase the abductor hallucis CSA compared to no intervention (median changes = 0.42cm^2 for MRS and 0.36cm^2 for control, $p = 0.203$).²¹¹ A three-month transition to MRS running showed significant increases in abductor hallucis CSA, abductor digiti minimi CSA and volume – but not flexor digitorum brevis CSA – compared to running with traditional footwear (SMD ranging from 1.79 to 2.78).²¹⁴ Lastly, using MRS during daily activities

significantly increased quadratus plantae CSA compared to no intervention, with a moderate effect size (SMD = 0.61, $p < 0.05$).¹⁸⁴

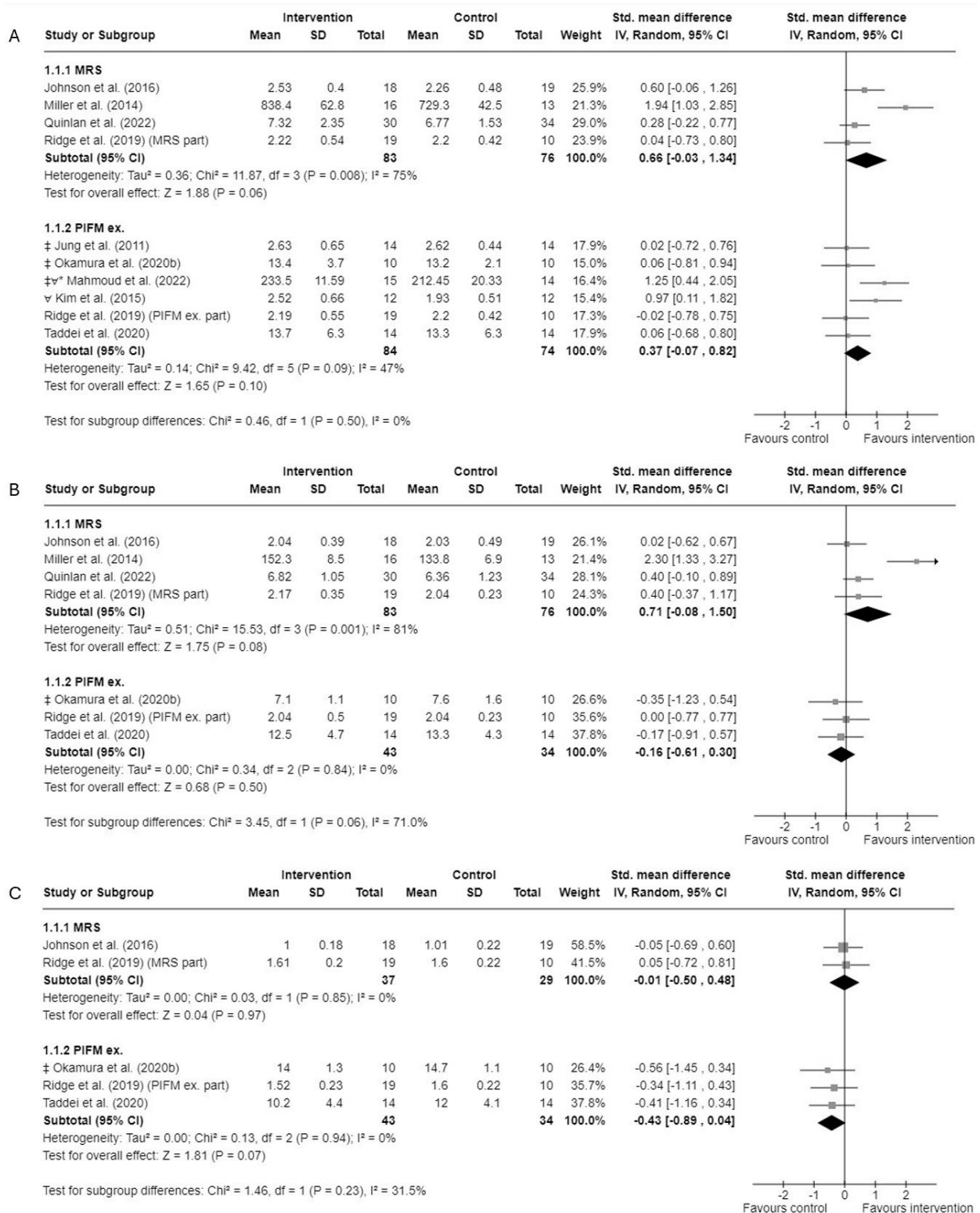


Figure 16. Meta-analyses of plantar intrinsic foot muscle sizes. A: Abductor hallucis, B: Flexor digitorum brevis, and C: Flexor hallucis brevis. MRS (minimalist running shoes), PIFM ex. (exercises targeting the plantar intrinsic foot muscles), CI (confidence interval), SD (standard deviation), * (participants with a symptomatic status), ‡ (participants diagnosed with pes planus), ∇ (participants affected with another musculoskeletal disorder).

4.2. Foot strength

Meta-analyses were conducted for hallux, toes 2-5 and metatarsophalangeal joint (MTPJ) flexion (Figure 17). The participants who used MRS showed significantly greater flexion strength in their toes 2-5 compared to control group (SMD = 0.39 [0.02, 0.76], $I^2 = 0\%$, $p = 0.04$), but this was not the case for PIFM exercises (SMD = 0.16 [-0.04, 0.37], $I^2 = 0\%$, $p = 0.12$). Hallux flexion strength did not change significantly between intervention and control groups (PIFM exercises: SMD = 0.04 [-0.17, 0.25], $I^2 = 0\%$, $p = 0.71$ and MRS: SMD = 0.33 [-0.03, 0.70], $I^2 = 0\%$, $p = 0.07$). Similarly, there was no significant difference between the PIFM exercises and the control group for hallux and toes together (SMD = 0.41 [-0.16, 0.99], $I^2 = 0\%$, $p = 0.16$) nor for MTPJ flexion (SMD = 1.16 [-0.23, 2.55], $I^2 = 80\%$, $p = 0.10$).

The synthesis of data not included in the meta-analyses included strength exerted while flexing the hallux, the toes 2-to-5, the 5 toes together, and MTPJ, alongside foot doming strength. Adding PIFM exercises to in-person extrinsic exercises in older adults significantly increased hallux flexion (SMD = 0.38) and toes 2-to-5 flexion (SMD = 0.15) in the intervention group but not in the control group that performed similar exercises at home without increasing intensity.²¹⁹ PIFM exercises increased MTPJ strength with huge effect sizes (SMD = 2.03, $p < 0.001$) compared to no-intervention.²⁰² PIFM exercises combined with other exercises also increased doming strength compared to maintaining habitual running.¹⁸⁴ PIFM exercises combined with electromyographic-biofeedback increased toe grip strength compared to no biofeedback on the non-dominant side (SMD = 0.46, $p = 0.029$), but not for the dominant side.²²⁶ Lastly, transcranial direct current stimulation was beneficial with small but significant effect sizes for toes 2-5 strength (SMD = 0.24, $p = 0.015$), and very large effect sizes for MTPJ strength compared to no intervention (SMD = 1.49, $p = 0.008$).²²⁴ The use of MRS during daily activities improved foot doming strength compared to no intervention (SMD = 0.35, $p < 0.05$).¹⁸⁴ Besides, combining MRS during running with PIFM exercises and additional exercised increased MTPJ strength in the intervention group but not in the group that maintained habitual running.²²⁷

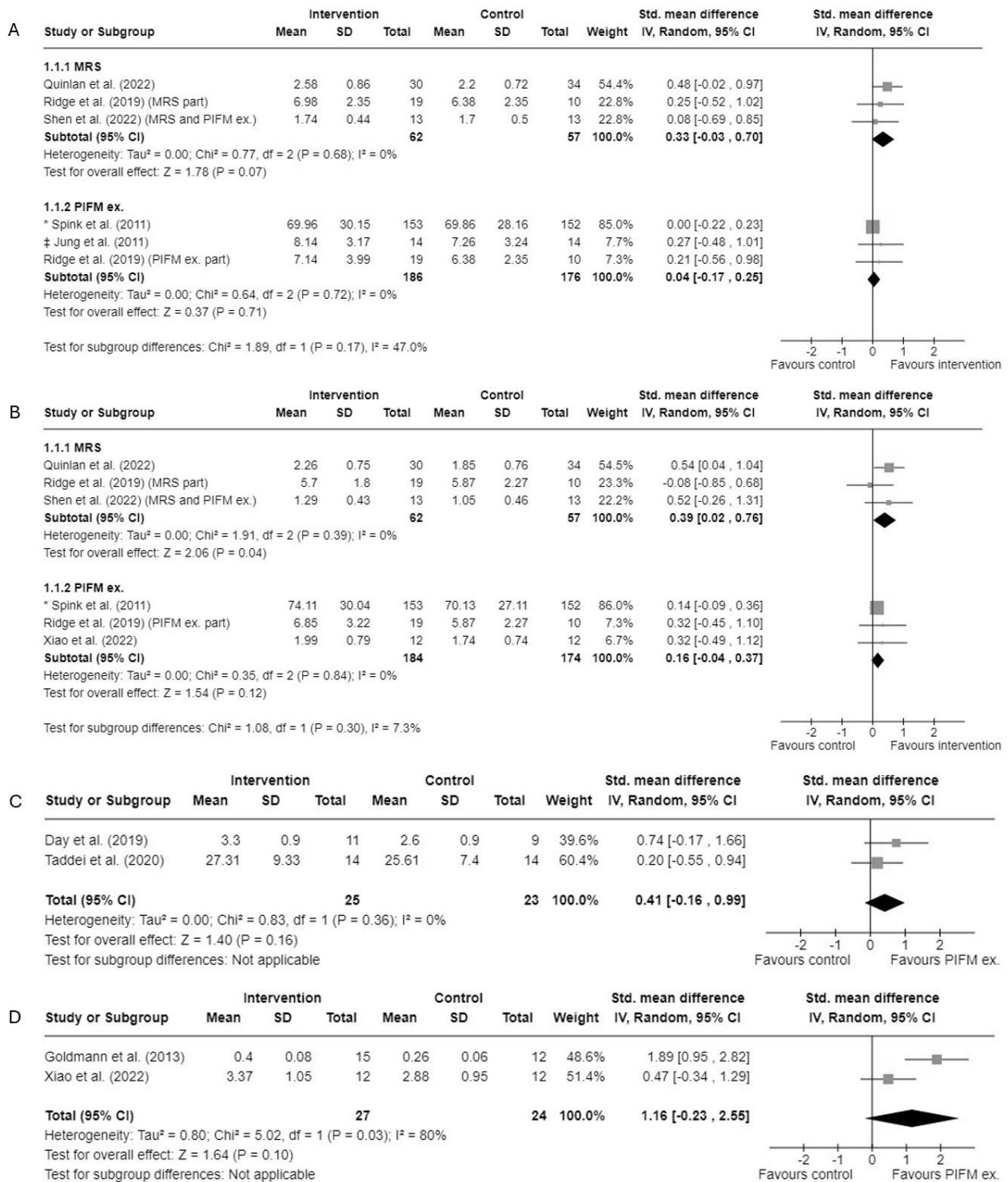


Figure 17. Meta-analyses of foot strength. A: Hallux flexion, B: Toes 2-to-5 flexion, C: Hallux and toes flexion, and D: Metatarsophalangeal flexion. MRS (minimalist running shoes), PIFM ex. (exercises targeting the plantar intrinsic foot muscles), CI (confidence interval), SD (standard deviation), * (participants with a symptomatic status), ‡ (participants diagnosed with pes planus).

4.3. Biomechanics

Meta-analyses of biomechanical outcomes are available in Appendix 1.E. Meta-analyses were first conducted for MLA motion during walking and running as well as foot incidence at foot strike while running. MLA motion was significantly lower during running in the PIFM exercise group compared to the control group (SMD = 0.45 [0.08, 0.82], $I^2 = 0\%$, $p = 0.02$), but not during walking (SMD = -0.01 [-0.59, 0.56], $I^2 = 5\%$, $p = 0.96$). For foot incidence at foot strike after MRS compared to control, the CI lower bound was zero (SMD = 0.78 [0, 1.56], $I^2 = 59\%$, $p = 0.05$). Meta-analyses were also conducted for vertical and horizontal ground reaction forces, ankle sagittal moment, and the first MTPJ sagittal moment. The PIFM exercise and control groups did not show significant differences in vertical ground reaction forces during walking and running (respectively SMD = -0.09 [-0.60, 0.42], $I^2 = 0\%$, $p = 0.73$ and SMD = 0.21 [-0.50, 0.93], $I^2 = 64\%$, $p = 0.56$), anterior ground reaction forces during running (SMD = -0.03 [-0.39, 0.34], $I^2 = 0\%$, $p = 0.89$), and the peak sagittal moments during running at the ankle and first MTPJ (respectively SMD = 0.07 [-0.87, 1.02], $I^2 = 61\%$, $p = 0.88$ and SMD = -0.22 [-0.80, 0.35], $I^2 = 0\%$, $p = 0.45$). Lastly, meta-analyses were performed on walking stance phase duration and time to lowest MLA height, which also did not show significant differences between the PIFM exercises and control (respectively SMD = -0.01 [-0.66, 0.63], $I^2 = 41\%$, $p = 0.97$ and SMD = 0.54 [-0.03, 1.11], $I^2 = 0\%$, $p = 0.06$).

The synthesis of data showed that PIFM exercises were found to be associated with biomechanical changes compared to the control groups during running and walking. During running, there was a significant increase in the shank-calcaneus inversion peak angle after PIFM exercises compared to warm-up and stretching (SMD = -0.37, $p = 0.037$).²¹⁶ Additionally, at 14 km·h⁻¹, the control group showed a significantly higher increase in the first MTPJ sagittal plane range of motion (ROM) compared to the PIFM exercise group, and at 18km·h⁻¹, foot contact time was decreased.²¹⁰ Significant differences were observed using Statistical Parametric Mapping for Chopart dorsiflexion and inversion, calcaneus-metatarsals adduction, Lisfranc dorsiflexion, and first MTPJ abduction and dorsiflexion ($p < 0.05$).²¹⁶ Coordination of two joint couples – calcaneus-midfoot and midfoot-metatarsus – showed a significant increase following PIFM exercises compared to warm-up and stretching ($p < 0.05$), but not for the other joint couples.¹⁹¹ During walking, PIFM exercises did not have significant impact on kinematic variables

compared to control interventions. For walking kinetic variables, the medial ground reaction peak significantly decreased in PIFM exercise group compared to the control group (SMD = -0.66, $p = 0.033$). Walking peak plantar pressure and foot contact area were not significantly affected,¹⁹⁴ but centre of pressure excursion index was more laterally deviated in the intervention group.¹⁹² Single-leg squat dynamic valgus index and jump (horizontal and vertical) kinetics did not show significant difference between groups.^{202, 217}

Distinct biomechanical changes were also identified following MRS used during running. In a study implementing a 3-month transition to MRS running, there was a notable decrease in maximum MLA angle during rearfoot strike running compared to no intervention ($p = 0.031$), although no significant changes were observed during a forefoot strike running task. During the rearfoot strike running task, the intervention group also exhibited significant changes in arch angle range of motion, arch height range of motion, and arch stiffness when compared to the control group, though no statistical significance was reported between groups.²²⁷ In another 3-month transition to running in MRS combined with PIFM exercises, there was a significant increase in ankle range of motion while running shod, but not when running barefoot, along with a change in foot strike pattern (heel-toe latency) from rearfoot strike to forefoot strike pattern compared to the habitual running shoe group. Differences in knee kinematics were not statistically different between groups.²⁰⁶ A third study of running in MRS also implemented PIFM and extrinsic foot muscle exercises. This intervention did not lead to significant changes in loading rate, impact peak force, vertical stiffness, stiffness of the knee and ankle during running in traditional shoes (between group p -values not available).²⁰⁴ Exhaustive biomechanical findings are displayed in Appendix 1.D.

5. GRADE rating for meta-analysis

The evidence of the meta-analysis assessing the impact of PIFM exercises on abductor hallucis muscle sizes was of low certainty. The quality of the evidence was very low for all other meta-analyses (Table 4). This was mainly due to high risk of bias, low sample sizes and indirectness. Funnel plots could not be performed due to the limited number of RCT in each meta-analysis.

Table 4: Summary of findings for plantar foot muscle exercises or minimalist shoes

	Outcome	A	B	C	D	E	F
Plantar foot muscle exercises	Abductor hallucis sizes	0	-1 ^b	0	-1 ^f	0 ^h	Low
	Flexor digitorum brevis sizes	-1 ^a	-2 ^{b, c}	0	-1 ^{f, g}	0 ^h	Very low
	Flexor hallucis brevis sizes	-1 ^a	-2 ^b	0	-1 ^{f, g}	0 ^h	Very low
	Hallux flexion strength	-1 ^a	-1 ^b	0	-1 ^{f, g}	0 ^h	Very low
	Toes 2-to-5 flexion strength	-1 ^a	-1 ^b	0	-1 ^f	0 ^h	Very low
	Hallux and toes strength	0	-2 ^b	0	-1 ^{f, g}	0 ^h	Very low
	Metatarsophalangeal flexion	-1 ^a	-2 ^{b, c}	-1 ^d	-1 ^{f, g}	0 ^h	Very low
Minimalist shoes	Abductor hallucis sizes	-2 ^a	-1 ^b	-1 ^{d, e}	-1 ^{f, g}	0 ^h	Very low
	Flexor digitorum brevis sizes	-2 ^a	-1 ^b	-1 ^{d, e}	-1 ^{f, g}	0 ^h	Very low
	Flexor hallucis brevis sizes	-2 ^a	-2 ^{b, c}	0	-1 ^{f, g}	0 ^h	Very low
	Hallux flexion strength	-2 ^a	-2 ^b	0	-1 ^{f, g}	0 ^h	Very low
	Toes 2-to-5 flexion strength	-2 ^a	-2 ^b	0	-1 ^g	0 ^h	Very low

Legend: A (Risk of bias), B (Imprecision), C (Inconsistency), D (Indirectness), E (Publication bias), F (Certainty of the evidence). ^a (-1 if one article has 2 high risk domains and -2 if most articles have 2 high risk domains), ^b (small sample size), ^c (overall confidence interval includes contradictory findings), ^d (substantial heterogeneity), ^e (confidence intervals do not overlap), ^f (blended interventions in the intervention groups), ^g (restricted population with respect to the research question), and ^h (funnel plot could not be performed due to the limited number of studies). For each outcome, the corresponding meta-analysis can be seen in Figure 16 for muscle sizes and in Figure 17 for strength.

Discussion

This systematic review and meta-analysis aimed to investigate the impact of PIFM exercises and MRS on foot muscle sizes, foot strength, and biomechanical variables.

Out of the systematic review, performing foot muscle exercises seem to be appropriate to increase foot strength and it could increase the sizes of some PIFMs. Wearing MRS during running or during daily activities could be appropriate to increase the sizes of multiple PIFMs and may be helpful to increase foot strength. Biomechanical effects during walking and running were observed for both interventions. Besides, we conducted 17 meta-analyses and assessed the certainty of their evidence using GRADE methodology.²⁰⁰ The uncertain to very uncertain evidence summarized in the meta-analyses revealed the following: First, PIFM exercises may have moderate effects on abductor hallucis size, all toes flexion strength, and metatarsophalangeal flexion strength. It may have trivial effects on hallux strength and toes 2-to-5 strength. Regarding the biomechanical variables, the meta-analyses revealed a decreased MLA mobility while running and decreased time to lowest MLA height while walking, with very low quality of evidence. Second, MRS may have a moderate effect on abductor hallucis size, flexor digitorum brevis size and foot strike pattern while running. It may also increase hallux flexion and toes 2-to-5 flexion strength.

Several experimental studies have highlighted the increased recruitment of the PIFM during specific exercises, including the short foot, toe-spread out, first toe extension, second-to-fifth toes extension,¹⁷⁷ toe grip, and metatarsophalangeal flexion exercises.²²⁸ Also, PIFM were shown to be thicker after years of MRS used during daily life compared to wearing traditional footwear.²²⁹ Therefore, several weeks to months of these interventions have the potential to generate hypertrophy and increased strength. However, the RCT included in this systematic review did not consistently report increased strength and muscle sizes. These RCT used ultrasound and magnetic resonance imaging to estimate PIFM sizes, as supported by various reliability studies.¹⁴⁸⁻¹⁴⁹ To estimate foot strength, hand-held and fixed dynamometers were used, as well as plantar pressure technologies, which are also reliable tools to estimate toe flexion and doming strength.^{137, 166}

The conflicting findings found in the included RCT could be explained by various sources of heterogeneity. First, the focus of this systematic review was on

symptomatic and asymptomatic participants. However, for MRS, no study included symptomatic participants, neither older people, so the evidence applies to young healthy participants transitioning from traditional to minimalist running shoes or wearing them during daily life activities. Contrarily, PIFM exercises were implemented in symptomatic participants (hallux valgus and low back pain with pes planus) and seemed to be effective at increasing PIFM sizes.^{218, 222} These exercises were also implemented in older people and were more effective than active control to increase PIFM sizes, but not strength.²¹⁹ Adding electromyographic-biofeedback to PIFM exercises was also more effective on strength than the same exercises alone in older people,²²⁰ suggesting that older people can benefit from these interventions in an attempt to decrease fall risk.¹⁷⁵ Second, the duration, frequency and volume of the interventions varied considerably. Durations varied from 1 to 9 months, and although strength already increases in the first few days of interventions primarily due to neuromuscular adaptations,²³⁰ muscle hypertrophy requires more time and is not expected to occur before 4 or even 6 to 10 weeks of training.²³¹ Besides, training volumes varied from light (e.g. exercises performed once a week or MRS used less than an hour per week)^{211, 226-227} to heavy (e.g. exercises performed or MRS used daily).^{184, 202, 213, 218} Interestingly, PIFM sizes seemed to be increased following heavy training volumes,^{152, 202, 213, 218} but not following lighter interventions.^{211, 226} This relationship could also be true for foot strength, but less of these “light” or “heavy” training volume studies investigated that outcome. For example, Shen et al. implemented a transition to MRS worn less than one hour a week combined with PIFM exercises performed once a week,²²⁷ and Ridge et al. included both a group wearing MRS daily and another performing various PIFM exercises at least 5 times per week.¹⁸⁴ Although Shen et al.’s intervention lasted 3 months, they reported increased strength for hallux flexion, but not for toes 2-to-5 flexion while Ridge et al.’s study, that lasted two months, resulted in benefits for both interventions over control for hallux flexion, toes 2-to-5 flexion and doming strength. Third, PIFM exercises were often included in blended interventions, for example combined with exercises known to recruit extrinsic foot muscles such as heel raises.^{184, 203, 216, 219}

During walking and running, the PIFM help attenuate foot impact at landing through excentric contraction and support foot stiffness during push-off through rapid concentric contraction.¹⁵³ Sprint performance seem to be determined by toe strength,²³² and habitual running foot strike pattern influences toe strength, with

more strength developed in forefoot strikers.²³³ Therefore, PIFM strengthening exercises, as well as walking and running MRS have the potential to modify dynamic task biomechanics. Other published meta-analyses showed that PIFM exercises resulted in decreased sit-to-stand navicular drop as well as a more supinated standing foot posture.^{180, 186-187} These assessments of MLA mobility and foot posture are regarded as non-validated surrogate measures for PIFM strength.¹⁶⁶ Indeed, previous experiments highlighted that an increased activation of the PIFM results in a lower navicular drop and a more supinated foot posture.¹⁶⁵ Moreover, MLA mobility is correlated with abductor hallucis and flexor hallucis brevis sizes,²³⁴ and foot posture is partly explained by flexor hallucis brevis and flexor digitorum brevis muscle sizes.²³⁵ Stronger PIFMs should thus theoretically result in a more elevated MLA and a more supinated foot posture, along with subsequent kinematic and kinetic changes during dynamic tasks. Evidence in favour of these hypotheses were found in some of the included RCT. First, PIFM exercises may result in less MLA mobility, a more plantarflexed Chopart joint, a more inverted shank-calcaneus joint,²¹⁶ and an increased vertical impulse during running.²²³ Second, walking following PIFM exercises may be characterized by a more laterally oriented centre of pressure,¹⁹² a decreased medial ground reaction force, and a decreased time to minimal MLA height.²²⁶ Third, MRS transition consistently led to a foot strike pattern more oriented towards forefoot strike,^{206, 214, 227} although this was biased in one study that assessed the intervention group in MRS and the control group in their habitual footwear.²¹⁴ However, one should admit the numerous non-significant findings associated with these interventions. Biomechanical changes are thus unpredictable and could be limited to isolated variables.

Transition to MRS can be associated with an increased risk of sustaining a running related injury.²³⁶ Therefore, a safe transition requests a gradual increase in MRS use over a long period of time, in parallel with additional injury prevention exercises. These injury prevention programs typically include exercises oriented towards the foot muscles (intrinsic and extrinsic) and balance training.²³⁷ Opposite to these recommendations, some of the included RCT implemented transitions to MRS over short periods of time and without injury prevention policy,^{211-212, 214} which could partly explain the high rates of adverse events that some authors reported.²¹¹⁻²¹² Contrarily, performing PIFM exercises does not seem to be associated with a particular risk of injury, as evidenced by the very low number of

adverse events reported. Even better, these PIFM exercises could decrease the rate of running related injuries.²³⁸ Besides injury, adherence rates were sometimes low in the included RCT.^{183, 203} Physiotherapists should be attentive to maintaining the motivation of their participants in order to promote the correct application of the interventions.

Although safe, exercises targeting the PIFM can be difficult to learn,¹⁶⁶ and activation of these muscles can also be obtained through functional exercises which are easier to perform and could result in benefits not only in the foot muscles.²³⁹

Limitations

The results of this systematic review with meta-analysis should be considered in light of its limitations. First, a single investigator performed systematic data extraction in the pre-established tables. Second, the limited quality of the evidence – mainly due to the very reduced number of study participants, the high risk of bias and the variety of interventions – prevents the recommendation of specific PIFM exercises and MRS transition programs as standalone interventions in clinical practice. Third, the evidence cannot be generalized to children for PIFM exercises and MRS used during running, nor can it be applied to older individuals or symptomatic people for MRS. Lastly, study durations were limited, especially for PIFM exercises, so effects of longer periods of interventions are unknown.

Perspectives

Future research should focus on the long-term effects of PIFM exercises and MRS, as well as the benefits of MRS used in children. In symptomatic populations, future studies on PIFM exercises should incorporate these interventions into existing rehabilitation. The feasibility and safety of using MRS could be evaluated in patients with specific musculoskeletal disorders. Besides, the outcomes of interest need to be assessed using reliable methods, with examiners blinded whenever possible.

Practical applications

Performing PIFM exercises or wearing MRS can be used in attempts to improve foot strength. Runners performing PIFM exercises can expect to have a less mobile medial longitudinal arch, and runners using MRS are likely to gradually change from a rearfoot to a forefoot strike pattern. PIFM exercises appear to be safe, while MRS require a long transition period that should include strengthening exercises.

Conclusion

Exercises targeting the plantar intrinsic foot muscles seem to increase foot strength and could result in a more elevated medial longitudinal arch during running, but there is too much uncertainty regarding their effects on foot muscle sizes. The use of minimalist shoes during running or daily life seem to result in increased foot muscle sizes and foot strength, and a more forefoot-oriented strike pattern during running.

Acknowledgements

The authors would like to acknowledge Mrs. Lisa Croce for her contribution to the article selection, Mrs. Sophie Patris and all the members of the UCLouvain librarian team and the Désiré Collen learning centre for their help on the search strategy, as well as Dr Gauthier Everard for sharing his expertise on meta-analysis methodology.

Summary

Context

Impaired intrinsic foot muscle function in some individuals with CAI could support the integration of interventions aiming to strengthen those muscles. It was unclear whether specific foot exercises and the use of minimalist footwear could increase foot muscle sizes, foot strength and modify lower-limb biomechanics.

Findings

This work summarizes the findings of 28 randomized controlled trials. Most studies were related to performing various exercises, with interventions performed for durations often close to one month, among both asymptomatic and symptomatic participants. In nine studies, participants used minimalist footwear during daily life or during running, for durations up to 9 months, among asymptomatic young participants.

Risk of bias was generally high, mainly due to possible deviations from intended interventions and missing outcome data. We were able to conduct 17 meta-analyses, but each meta-analysis included the data of a restricted number of studies and participants, leading to imprecision. Many meta-analyses included both ‘no effects’ and moderate-to-important effects under the confidence interval (CI) of their standardized mean difference. For these reasons, the certainty of the evidence was regarded as (very) low.

Meta-analyses showed that performing specific exercises may have trivial to moderate effects on abductor hallucis sizes (CI = -0.1, 0.8) and foot strength outcomes (e.g. CI = -0.2, 1 for hallux and toe flexion). These exercises were also able to decrease medial longitudinal arch mobility during running, as a possible consequence of increased stiffness at midfoot joints. Using minimalist footwear may moderately increase the sizes of abductor hallucis (CI = -0.0, 1.3) and flexor digitorum brevis (CI = -0.1, 1.5), along with trivial effects on strength outcomes (e.g. CI = -0.0, 0.7 for hallux flexion). Foot incidence is also likely to moderately increase among runners using minimalist shoes, who will thus exhibit a more ‘forefoot’ strike pattern.

Implications for clinical practice and future research

Exercises aiming to strengthen the intrinsic foot muscles seem to be safe among symptomatic participants, but studies should implement interventions of higher dosage (load, frequency, repetitions) applied for long periods of time (several months). Other studies should assess the safety and effectiveness of using minimalist shoes during daily life and running among symptomatic participants.

Chapter 2. Comparison of concentric and isometric repeated foot muscle contraction protocols on neuromuscular fatigue and running biomechanics: a crossover randomized controlled study

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Contextualisation

The reader should be informed that data collection for *Chapter 2* occurred after that for *Chapter 5*. In *Chapter 5*, participants with CAI completed repeated isometric toe flexion tasks combined with electrical stimulation. After recruitment of approximately half of the planned sample, preliminary analyses indicated only minimal reductions in hallux flexion strength. Consequently, a decision was made to compare the isometric protocol with a concentric protocol, as detailed below.

Abstract

Background: Foot muscles contribute to receptive and propulsive dynamics during running. Prior studies have attempted to fatigue these muscles using short and unloaded protocols, producing inconsistent results. Consequently, the specific impact of foot muscle fatigue on running biomechanics remains poorly understood.

Research questions: Does repeated concentric versus isometric toe-flexor contractions result in fatigue and altered ground reaction forces while running?

Methods: Crossover randomized controlled study. Recreationally active participants performed repeated concentric toe flexions and repeated isometric toe flexions with concurrent electrical stimulation of the medial longitudinal arch, one week apart. Hallux flexion strength and ground reaction forces during running at 9 and 12 km·h⁻¹ were analysed using repeated-measures general linear models.

Results: Nineteen participants (25±4.6 years, 16 males) completed the study. After concentric contractions, hallux flexion strength decreased by 20 % ($p < 0.001$, Hedge's $g = -1.07$). Leg stiffness increased by 5.2 % ($p < 0.001$, $g = 1.03$) and propulsive impulse decreased by 2.9 % ($p = 0.01$, $g = -0.82$) while running at 12 km·h⁻¹. After isometric contractions, hallux flexion strength did not significantly change. Leg stiffness increased by 3.6 % at 9 km·h⁻¹ ($p = 0.02$, $g = -0.75$), and vertical impulse decreased by 1.5 % at 9 km·h⁻¹ and 1.8 % at 12 km·h⁻¹ ($p = 0.03$ and $g = -0.7$ for both).

Significance: Concentric, but not isometric, toe-flexor contractions induced hallux flexion fatigue. Both protocols altered running biomechanics, highlighting the acute influence of foot muscle activity on running, in presence and absence of fatigue.

Introduction

The muscular component of the foot is divided into two groups: the intrinsic foot muscles (IFMs) originate within the foot, while the extrinsic foot muscles (EFMs) originate from the lower leg. Analytically, flexor hallucis longus and brevis, abductor hallucis and adductor hallucis perform hallux flexion, and flexor digitorum longus, quadratus plantae and flexor digitorum brevis enable flexion of toes 2-5.²⁴⁰ During standing, the IFMs provide postural support through low-intensity isometric contractions.¹⁶⁴ During dynamic activities such as running, the IFMs help attenuate foot impact upon landing and support metatarsophalangeal stiffness during push-off.¹⁵³ The proper function of these muscles is crucial for the foot to act as a highly versatile and adaptable system,⁹ absorbing and generating energy during locomotor and sporting tasks.^{20, 23}

The function of the EFMs and IFMs can be impaired in various musculoskeletal, systemic, or neurological conditions, as well as with aging,¹⁴⁹ which may lead to structural overload and injury. In experimental research, nerve blocks have been used to simulate such impairments by producing a near-complete cessation of muscle activity.^{19, 23} However, nerve blocks not only suppress muscle activation but also diminish proprioception and cutaneous sensation, which may confound their effects on lower-limb function.¹⁵⁶ By contrast, neuromuscular fatigue, defined as a loss of maximal force-generating capacity following sustained or repeated muscle activity,¹⁵⁷ provides a more physiologically relevant approach to model altered muscle function.

Several studies have already attempted to produce foot muscle neuromuscular fatigue to further investigate their roles during postural and dynamic activities, with varying success.¹⁵⁸⁻¹⁶³ Hallux flexion and abduction strength decreased by 26 % after a sustained isometric hallux flexion and abduction contraction, maintained until failure.¹⁵⁹ Repeated concentric hallux flexion movements resulted in a 17 % decrease in IFM activity,¹⁶¹ but no change in strength in another study,¹⁶³ and fatigue levels were unclear in two studies.^{160, 162} Electrical stimulation-evoked hallux flexion strength decreased by 74 % following electrical stimulation applied to the medial longitudinal arch.¹⁵⁸ These contrasting results could be explained by protocols often being performed on unloaded feet (seated position) during short durations (1 to 8 minutes),^{160, 162} which emphasizes the need to develop more demanding protocols. Further, isometric and concentric modalities are both used

by the foot muscles during functional and sporting tasks, but no study has directly compared their effects on fatigue.

The selective recruitment of the IFMs is a particular challenge in the context of foot muscle fatigue and strengthening exercises, since IFMs and EFM functions overlap. Neuromuscular electrical stimulation applied to the medial longitudinal arch (MLA) has been proposed as a relevant strategy to enhance IFM recruitment during isometric exercises such as ‘the short foot exercise’.¹⁶⁶ Incorporating electrical stimulation into isometric fatigue protocols could therefore ensure IFM recruitment.

Impaired IFM function modifies ground reaction forces during sporting tasks.²³ Previous foot muscle fatigue protocols occasionally affected walking biomechanics, with for example modified hindfoot kinematics after repeated concentric hallux flexion movements,¹⁶² longer step length and longer single-support duration after a sustained isometric hallux flexion protocol.¹⁶⁰ Many sports involve bouts of running, but the effects of IFM fatigue on running biomechanics remain unexplored. This knowledge would contribute to a better understanding of the foot muscle roles during running.

Therefore, the main objective of this study was to compare the effects of concentric and isometric contraction protocols on hallux flexion strength. The second objective was to examine how toe flexor fatigue influences ground reaction force-related outcomes during running.

We hypothesized that both the concentric and isometric protocols would result in a decreased hallux flexion strength, quantifying neuromuscular fatigue. During running, we expected increased foot contact time, increased peaks in vertical ground reaction forces, and decreased vertical stiffness.

Methods

1. Experimental approach

This study was written in accordance with the CONSORT Guidelines Extension for Crossover Trials.²⁴¹ A crossover design provides greater statistical power with fewer participants than a parallel design.²⁴¹

2. Participants

Twenty recreationally active participants were recruited between December 2023 and April 2024, of whom 19 completed the study (Table 5). Inclusion criteria were being older than 18 years, absence of lower limb symptoms, and at least one hour of moderate-to-vigorous physical activity per week. Exclusion criteria encompassed neurological, musculoskeletal, or systemic diseases, pregnancy, and lower limb injury or surgery within six months prior to participation. All participants provided written informed consent (Comité d'éthique hospitalo-facultaire, B403201523492).

Table 5: Characteristics of the participants (n = 19)

Outcomes	Values
Sex (M/F)	16/3
Self-reported dominancy (L/R)	6/13
Age (year)	25.2 (4.6)
Weight (kg)	74.1 (11.5)
Additional weight used during the interventions (%BW)	26.4 (3.2)
Height (cm)	178.0 (7.6)
Body mass index (kg·m ⁻²)	23.3 (2.6)
6-item Foot Posture Index	5 [2, 6]
Self-reported moderate to intense physical activity per week (h·week ⁻¹)	3.7 (2.0)

Legend: Continuous quantitative data are presented as mean (standard deviation), discrete quantitative data are presented as median [interquartile range], and qualitative data are presented as numbers. Abbreviations: BW (body weight), L/R (left/right ratio), M/F (male/female ratio).

3. Sample size

Sample size was estimated based on the results of a pilot study on eight individuals (22 ± 4 years; 180.7 ± 8.2 cm height; 70.3 ± 13.7 kg) using the same interventions as described below. Long-sitting hallux flexion strength was 1.2 ± 0.2 N·kg⁻¹ before and 0.9 ± 0.2 N·kg⁻¹ after intervention A. Hallux flexion strength was 1.2 ± 0.3 N·kg⁻¹ before and 1.1 ± 0.3 N·kg⁻¹ after intervention B. The repeated-measures general linear model revealed a η^2_p effect size of 0.4. G*Power (version 3.1.) revealed that 18 participants would be necessary for $\alpha = 0.05$ and power = 0.8. To allow for dropouts, 20 participants were recruited.

4. Procedures

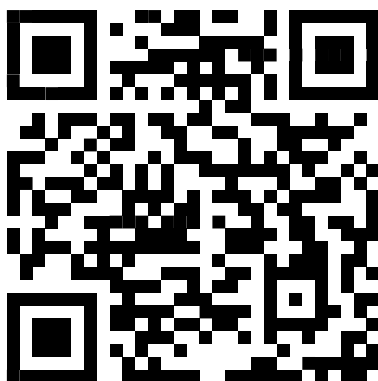
Participants attended two visits at the UCLouvain Neuromusculoskeletal Laboratory (Belgium), separated by a one-week washout period. During the first visit, baseline leg length and 6-item Foot Posture Index score were assessed on the dominant side.²⁴² Participants were randomly assigned to sequence order in blocks of four by drawing a sealed envelope. Participants remained barefoot and interventions targeted the self-reported dominant foot.²⁴³ During both interventions, participants wore a weighted vest of 20 kg for men and 15.2 kg for women. Due to the nature of the interventions, neither participants, therapists, nor the evaluator were blinded to the assignments.

Intervention A involved repeated concentric hallux and toe flexion movements against an Airex® balance pad (Airex AG®, Sins, Switzerland), with an estimated Young's modulus of 218 kN·m⁻².²⁴⁴ Participants stood on both feet, with their heel and midfoot placed on a steel platform level with the pad (Figure 18 and Video file 1). They pressed the hallux and toes of their dominant foot into the pad and controlled release at 40 repetitions/min, paced by a metronome.¹⁶¹⁻¹⁶³ Participants first familiarized themselves with this procedure for 3 minutes. The protocol consisted of eight sets of 75 contractions,¹⁶¹⁻¹⁶² separated by one recording of standing hallux flexion strength at the end of each set (approximately 10 seconds). The duration of intervention A was approximately 16 minutes.



Figure 18. Position during intervention A. The metatarsophalangeal joint was aligned with the edge of the steel platform. Participants performed toe flexions against the mat.

Video file 1: Demonstration of intervention A (repeated concentric toe flexions).

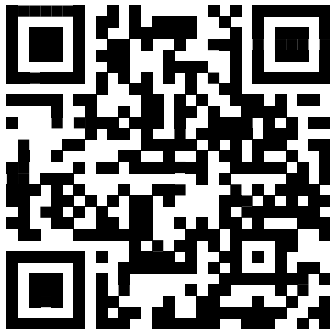


OR

<https://www.dropbox.com/scl/fi/l86lgj1rni5m78tgecxp2/Concentric-toe-flexion-protocol-JL-Peters-Dickie.mp4?rlkey=81w64wc5oegct6n7mzcfil9np&st=s8fogp0r&dl=0>

Intervention B consisted of repeated isometric hallux and toe flexor contractions combined with electrical stimulation applied to the MLA (Video file 2). Participants stood on both limbs and performed the intervention with their dominant foot. Two electrodes (5 x 5 cm EN-Trode, Enraf-Nonius®, Malle, Belgium) were placed on the MLA, between the medial process of the calcaneus and the first metatarsal head (Figure 19).²⁴⁵ A Tensmed S84 device (Arseus Medical®, Bornem, Belgium) delivered a biphasic current with a frequency of 30 Hz, an impulse width of 400 µs, a raising ramp of 1.5 s and a descending ramp of 0.5 s, by 8 s 'On' and 4 s 'Off' periods. This frequency had previously been associated with metabolic fatigue.¹⁶⁹ The 'On' period duration was chosen to promote greater fatigue compared to shorter periods commonly used in rehabilitation. Participants practiced for 3–5 min pushing down with toes during 'On' periods and relaxing during 'Off' periods, and the highest current intensity leading to tolerable discomfort was determined and served as starting intensity for Intervention B. During the 16-min protocol, participants shifted a part of their weight to the targeted limb by moving their hips and shoulders in that direction. They were frequently encouraged to increase current intensity.

Video file 2: Demonstration of intervention B (repeated isometric contractions)



OR

<https://www.dropbox.com/scl/fi/vqvva2er3mmosuq7vfc9v/Isometric-Electrical-stimulation-JL-Peters-Dickie.mp4?rlkey=m5cr4qrdd20yz1evzkpc1jgpx&st=dwbgr8ft&dl=0>



Figure 19. Placement of the adhesive electrodes used to deliver the electrical stimulation.

During both interventions A and B, a break was allowed if participants experienced significant discomfort in the involved lower limb. Participants were encouraged to continue their usual activities and avoid engaging in unusual physical efforts during the wash-out period.

Before interventions A and B, the sit-to-stand navicular drop was measured using a calliper (resolution 0.02 mm), followed by the recording of running ground reaction forces (Gaitway 3D treadmill, h/p/cosmos®, Arsalis, Belgium) and assessment of hallux flexion strength (microFET®2 hand-held dynamometer, Biometrics, France). The same investigator (J-L. P-D.) carried out all the measures. After each intervention, assessments were performed in the following order: hallux flexion strength, navicular drop, and running ground reaction forces. Before each running acquisition, participants familiarized with the treadmill by walking 1 min at $5 \text{ km}\cdot\text{h}^{-1}$. They then ran 1 min at $9 \text{ km}\cdot\text{h}^{-1}$ and 1 min at $12 \text{ km}\cdot\text{h}^{-1}$. Ground reaction forces were recorded at a frequency of 1000 Hz during the last 20 s of each running speed and were filtered using an 8th order Bessel low-pass filter.

5. Outcome measures

Primary outcome

The long-sitting hallux flexion strength quantified neuromuscular fatigue (Figure 20 part A)¹³⁹. To promote IFM activation compared to the EFM,¹⁴² participants were lying supine on an examination table, supporting themselves on their elbows, with the assessed foot at the edge of the table, knee flexed, and ankle in maximal plantarflexion without discomfort. The dynamometer was placed on a stool, transducer under the distal hallux phalanx and the evaluator stabilized the ankle. Participants pushed the hallux downward maximally for 5 s while keeping trunk and pelvis on the table,¹³⁹ followed by a break. They performed attempts until variation was less than 20 % across the last three trials. The mean of the two highest values normalized by body mass was then used in the analyses.

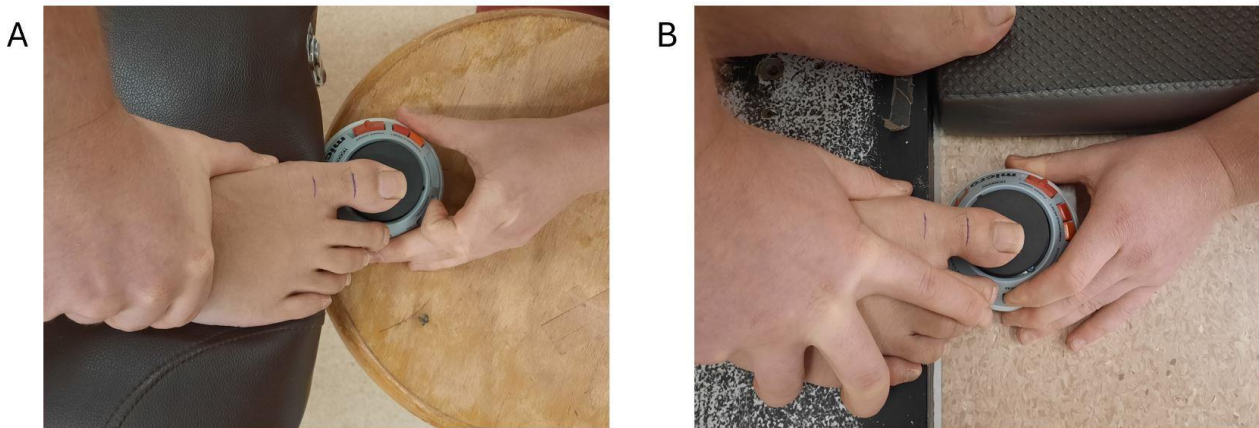


Figure 20. Assessment of hallux flexion strength in the long-sitting position (A) and while standing (B). The metatarsophalangeal joint of the hallux was vertically aligned with the edge of the table (A) or the edge of the steel platform (B), and the dynamometer transducer was placed under the distal hallux phalanx.

Secondary outcomes

Navicular drop was the difference in navicular height (marked with a pen) relative to the ground between sitting and standing.²⁴⁶

Ground reaction forces while running were used to compute the following hypothesis-driven outcomes: contact duration (% of stride), impact- and active-peak ground reaction forces (% of BW) and vertical stiffness ($\text{BW} \cdot \text{mm}^{-1}$). Other outcomes were also computed and definitions are given in Appendix 2.A.

Perceived fatigue was evaluated every four minutes during both interventions using a verbal numeric scale. Zero indicated no fatigue (same force as pre-test); ten, maximal fatigue with inability to exert force.

Hallux flexion strength ($\text{N}\cdot\text{kg}^{-1}$) was measured while standing before and at two-minute intervals during intervention A (Figure 20 part B and Video file 3). The dynamometer was placed on the ground, with the transducer horizontally aligned to the steel support and positioned under the distal phalanx of the hallux. An investigator ensured straight posture with ankles to shoulders aligned in the frontal plane. Moreover, the electrical stimulation current intensity (mA) was monitored every four minutes of intervention B.

Video file 3: Assessment of hallux flexion strength in standing (from 15th second)



OR

<https://www.dropbox.com/scl/fi/7h7mw4jhgzk0e8mmypqkb/Hallux-strength-standing-JL-Peters-Dickie.mp4?rlkey=2dfv1ooylinvuhq5ihzj1oyww&st=kz28ar3q&dl=0>

6. Statistical analyses

Analyses used SPSS v27 (IBM, Chicago, IL, USA). Demographics (continuous variables) were compared between sequence orders with Mann-Whitney non-parametric tests, because of normality violation observed in QQ-plots. Repeated-measures general linear models with four measurement times were used for each primary and secondary outcome. We reported partial eta squared (η^2_p) effect size, interpreted as the percentage of variance in each dependent variable explained by the main effect of time. In case of significant time effect ($\alpha = 0.05$), post-hoc multiple comparisons were applied (Bonferroni correction). The within-group Hedge's g effect sizes and corresponding 95 % confidence intervals (CI) were calculated and interpreted as: <0.2 very small, <0.5 small, <0.8 medium, <1.2 large, <2 very large, ≥ 2 huge.²⁰¹

Results

1. Participants

One participant withdrew from sequence A-B at the first visit due to dizziness (Figure 21). No significant between-sequence differences were found for any demographic outcome (Appendix 2.B).

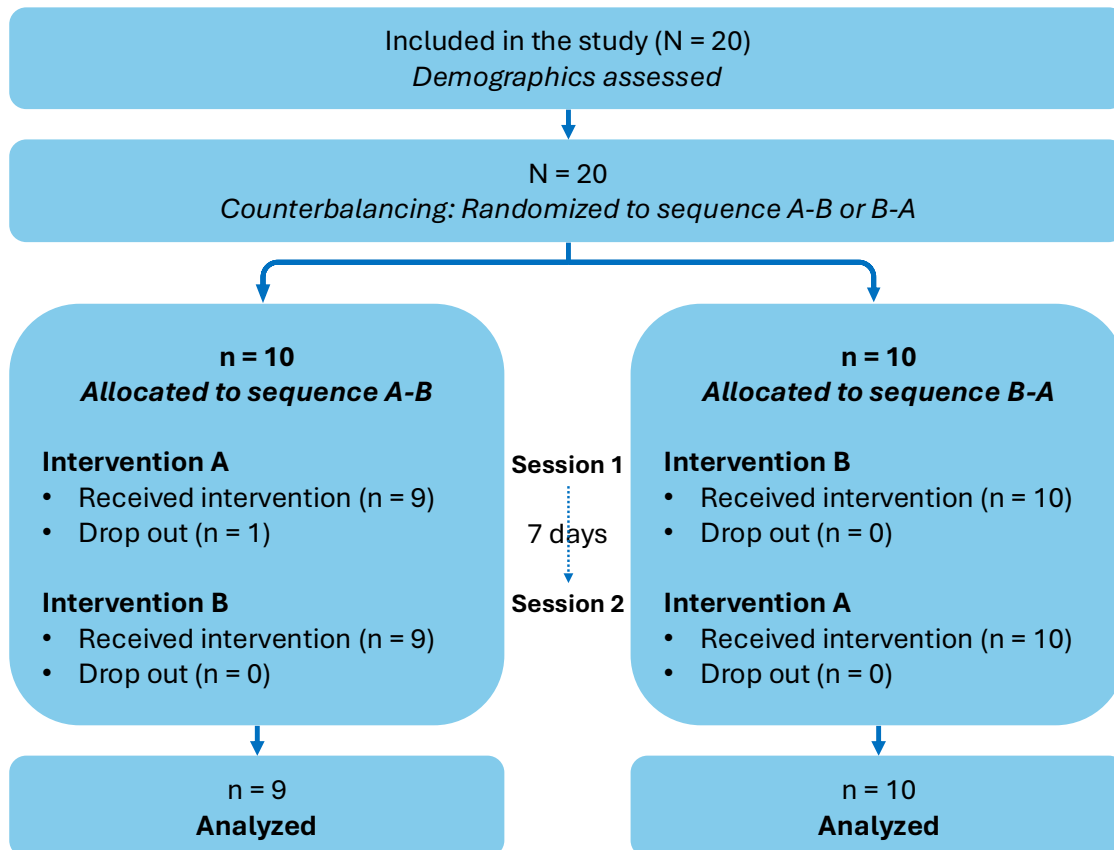


Figure 21. Flowchart of the interventions and data collection. Participants performed interventions with a 1-week wash-out period, in a random order depending on their sequence allocation. Interventions A consisted of concentric toe flexion contractions against a mat while intervention B involved isometric toe flexion contractions with electrical stimulation applied to the medial longitudinal arch of the foot.

2. Neuromuscular fatigue

There was a large decrease of 20 % in long-sitting hallux flexion strength after intervention A ($p < 0.001$), but not after intervention B, partially confirming our hypothesis. Navicular drop was not significantly affected by the interventions (Table 6). Standing hallux flexion strength during intervention A did not significantly change ($p = 0.44$; Appendix 2.C).

Table 6: Strength and navicular drop measurements before and after interventions

Outcomes	Intervention A		Intervention B		Inference	
	Before	After	Before	After	General linear model	Post-hoc p-values, g [95% CI]
Strength sitting ($\text{N}\cdot\text{kg}^{-1}$)	0.9 (0.3)	0.7 (0.3)	0.9 (0.3)	0.8 (0.2)	$p < \mathbf{0.001^*}$ $\eta^2_p = 0.29$	A: $p < \mathbf{0.001^*}$; g = -1.07 [-1.62; -0.50] B: $p = 0.30$; g = -0.47 [-0.93; -0.01]
Navicular drop (mm)	6.5 (2.8)	5.9 (2.7)	6 (3.6)	5.7 (4.0)	$p = 0.65$ $\eta^2_p = 0.02$	/

Legend: Data are presented as mean (standard deviation). Abbreviations: g (Hedge's g effect size), GLM (general linear model), η^2_p (partial eta squared effect size of the GLM), 95% CI (95% confidence interval), * and bold (significant at 0.05). Post-hoc p-values were corrected for multiple comparisons (Bonferroni).

3. Perceived fatigue and current intensity

Perceived fatigue increased during both interventions, reaching a median of 7 in intervention A and 6 in intervention B. Median electrical stimulation current intensity increased from 14 mA during familiarization to 16 mA at the end of intervention B ($p < 0.001$; Appendix 2.C).

4. Running biomechanics

None of the hypothesis-driven outcomes differed significantly between sequence order (Table 7).

Table 7: Hypothesis-driven biomechanical outcomes

Outcomes	Intervention A		Intervention B		Inference GLM
	Pre	Post	Pre	Post	
9 km·h⁻¹					
Contact duration (% of stride %)	38.5 (2.8)	38.5 (2.8)	38.8 (2.6)	38.6 (2.7)	p = 0.74; η^2_p = 0.02
Impact peak force (% BW)	143.2 (27.4)	142.8 (29.2)	136.2 (29.1)	141.7 (27.1)	p = 1; η^2_p = 0.12
Active peak force (% BW)	229.9 (21.3)	230.5 (21.6)	227.4 (17.9)	229.3 (20.1)	p = 0.47; η^2_p = 0.04
Vertical stiffness (% BW·mm ⁻¹)	2.9 (0.3)	2.9 (0.3)	2.9 (0.2)	2.9 (0.3)	p = 0.196; η^2_p = 0.082
12 km·h⁻¹					
Contact duration (% of stride)	35.0 (2.6)	34.7 (2.8)	34.9 (2.7)	34.8 (2.7)	p = 0.56; η^2_p = 0.04
Impact peak force (% BW)	160.6 (26.2)	163.9 (28.5)	163.0 (28.6)	163.9 (27.2)	p = 0.37; η^2_p = 0.06
Active peak force (% BW)	245.5 (25.4)	247.3 (25.1)	245.0 (21.4)	245.7 (22.5)	p = 0.51; η^2_p = 0.03
Vertical stiffness (% BW·mm ⁻¹)	3.2 (0.4)	3.3 (0.4)	3.2 (0.3)	3.3 (0.4)	p = 0.055; η^2_p = 0.13

Legend: Data are presented as mean (standard deviation). Abbreviations: BW (body weight), GLM (repeated-measures general linear model), η^2_p (partial eta squared effect size of the GLM), * and bold: significant at $p < 0.05$, † (significantly different from pre-intervention). The p-value and partial eta squared for the main effect of time are presented.

Secondary running outcome measures are detailed in Appendix 2.D. During running at 12 km·h⁻¹ following intervention A, leg stiffness had a large increase of 5.2 % ($p < 0.001$), while propulsive impulse had a large decrease of 2.9 % ($p = 0.01$) compared to pre-intervention values. During running at 9 km·h⁻¹ following intervention B, medium increases of 3.6 % in leg stiffness ($p = 0.02$) and medium decreases of 1.5 % in vertical impulse ($p = 0.03$) were observed. Vertical impulse also had a medium decrease of 1.8 % at 12 km·h⁻¹ ($p = 0.03$).

Discussion

This study examined the effects of two repeated toe flexion protocols on muscle fatigue and ground reaction force-related outcomes. Repeated concentric contractions decreased hallux flexion strength (-20 %), increased leg stiffness (+5 %), and reduced propulsive impulse (-3 %). In contrast, repeated isometric contractions with MLA electrical stimulation did not reduce hallux flexion strength but increased leg stiffness (+4 %) and reduced vertical impulse (-2 %). To our knowledge, this is the first study to compare concentric and isometric contraction modalities for inducing IFM fatigue.

Prior protocols investigating IFM fatigue lasted eight minutes at most and were often performed in a seated position.^{158-159, 161-163} They resulted in only modest fatigue,^{159, 161-163} justifying the development of two more challenging protocols in the present study. Both protocols lasted 16 minutes, were executed in standing position, and were performed with additional load using a weighted vest. Following the concentric protocol, hallux flexion strength decreased by 20 % and participants reported a median perceived fatigue of 7/10, indicating a high but submaximal level of fatigue. These findings align with previous reports of 36 % reduction in toe flexion strength after repeated toe flexions,¹⁶¹ and 10 % reduction in abductor hallucis activity after repeated hallux flexion movements.¹⁶² The observed strength decrease did not exceed the previously established minimal detectable change of 26–46 %.^{139, 141} This discrepancy may stem from methodological differences: in our study, the dynamometer was positioned on a stool, rather than on the investigator's knee as in previous validation studies. Our approach removed potential upward forces from the investigator and ensured that variations reflected only participants' performance. Given the paucity of reference values for hallux and toe flexion strength despite their clinical and athletic relevance, future studies should address this gap. Furthermore, rigid fixation of the dynamometer is recommended to improve measurement accuracy.²⁴⁷

Following the isometric protocol, hallux flexion strength did not significantly change, while perceived fatigue reached 6/10. These results differ from prior studies reporting a 26 % reduction in hallux flexion/abduction strength after isometric contractions¹⁵⁹ and a 74 % reduction in evoked hallux flexion strength after electrical stimulation of the MLA.¹⁵⁸

Duty cycle – the time under contraction relative to rest time –²⁴⁸ varied between our two protocols, which may help explain our results. The duty cycle of the concentric protocol is difficult to determine precisely, but it involved 40 repetitions/min (≈ 1 repetition/0.7 s). The isometric protocol consisted of repetitions of 8 s of isometric contractions followed by 4 s of rest (= 1 repetition/12 s), or a duty cycle of 2. Evidence from the knee extensors suggest that repeated concentric contractions induce greater fatigue than repeated isometric contractions, with 3 s ‘On’/2 s ‘Off’ (duty cycle = 1.5),²⁴⁹ but sustained isometric contractions induce equivalent or greater fatigue than repeated concentric contractions.²⁵⁰ The impact of duty cycle on foot muscle fatigue remains unclear and warrants future research.

The isometric contractions were combined with MLA electrical stimulation, based on earlier evidence of fatigue from both techniques applied separately.¹⁵⁸⁻¹⁵⁹ Electrical stimulation has also been proposed as an adjunct to ‘foot core’ exercises.²⁵¹ We applied a low-frequency current (30 Hz), as it is more likely to affect muscle contractility peripherally without involving central pathways.¹⁶⁹ However, participants tolerated a median current intensity of 16mA, lower than in previous foot core studies.^{211, 245} Low-frequency currents are generally less comfortable than higher-frequency currents,²⁵² which likely limited the increase of current intensity in our cohort.

In this study, participants carried an average load corresponding to 126 % of their body weight, when adding the weight of the vest to body weight. Previous work demonstrated higher IFM activation when seated participants bore 125 % of body weight on their lower limbs compared to 100 %.¹⁶⁵ Although we used fixed loads, rather than individualized percentages, the between-participant variation was low (SD = 3 % body weight). Future fatigue protocols could include real-time feedback such as electromyography or ultrasound imaging to maximize IFM activation and participants’ engagement.^{149, 220}

This study is also the first to examine running biomechanics after toe flexion fatigue. Both protocols increased leg stiffness by 3.6–5.2 %, which suggests a more effective transmission of vertical ground reaction forces,²⁵³ potentially associated with a greater activation of EFMs.²⁵⁴ The effects of protocols on ground reaction force impulses differed: after concentric contractions, propulsive impulse decreased, indicating reduced forward propulsion. After isometric contractions, vertical impulse decreased, reflecting altered vertical force application. Propulsive

impulse was similarly reduced following tibial nerve block in previous research,¹⁹ which strengthens the argument that our concentric protocol effectively impaired IFM function. These biomechanical changes highlight the contribution of the IFMs in propulsion during running, by supporting metatarsophalangeal joint stiffness in late stance.¹⁹

Several limitations should be acknowledged. First, participants did not receive real-time contraction feedback, preventing verification of maximal effort. Second, electrical stimulation was delivered at relatively low intensities, likely limiting its efficacy. Third, only hallux flexion strength was measured, excluding the contribution of lesser toes and providing an incomplete assessment of foot strength. Finally, biomechanical testing was performed barefoot, reducing external validity for shod running conditions.

Conclusion

Repeated concentric contractions performed under additional load decreased hallux flexion strength, increased leg stiffness, and reduced propulsive impulse, indicating IFM fatigue. In contrast, repeated isometric contractions with concurrent MLA electrical stimulation did not induce measurable strength loss, but did increase leg stiffness and reduce vertical impulse, suggesting biomechanical adaptations without overt fatigue. The lack of detectable fatigue may have been influenced by the low electrical stimulation intensity tolerated by participants. Future research should integrate real-time biofeedback to optimize IFM activation and provide a more comprehensive assessment of foot muscle fatigue.

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Summary

Context

Investigating the biomechanical effects of muscle fatigue enables to model muscle dysfunction. Previous studies already aimed to produce foot muscle fatigue, with varying success. We designed two protocols characterized by a longer duration and more load applied to lower-limbs compared to those used in previous studies.

The reader should be informed that the acquisitions related to *Chapter 2* were conducted after those of *Chapter 5*. In *Chapter 5*, participants with CAI performed repeated isometric toe flexions combined with electrical stimulation. After recruiting half of the intended sample size, preliminary analyses indicated only minimal declines in hallux flexion strength. Consequently, we decided to compare the isometric protocol to a concentric protocol to determine whether one of these approaches would better induce fatigue.

Findings

In this cross-over study, nineteen participants performed the concentric and the isometric protocol in a random order during two distinct sessions. Sixteen minutes of repeated concentric toe flexions performed while standing and wearing a weighted vest resulted in a 20 % decline in long-sitting hallux flexion strength. The corresponding effect size was important, although the change did not surpass the minimal detectable change of that outcome. Opposite to these findings, repeated isometric contractions performed for the same duration did not significantly modify hallux flexion strength. Participants tolerated low intensities of electrical stimulation, which help explain those results.

Ground reaction forces recorded while running were also used to compute various secondary outcome measures. Leg stiffness increased after the two protocols. Propulsive impulse (area under the anterior ground reaction force) declined after the concentric protocol, while vertical impulse declined after the isometric protocol.

Implications for clinical practice and future research

The proposed protocols may not be optimal to elicit foot muscle fatigue. The duration of future protocols could be individualized and performed until a predefined strength decline. Besides, higher electrical stimulation intensities should be encouraged. Real-time biofeedback could also support maximal effort of the participants and help investigators track biomarkers associated with fatigue.

Chapter 3. Quantitative ultrasonography of the foot muscles: a comprehensive perspective on reliability

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Abstract

Background: Quantitative ultrasound imaging is a popular technique to assess the structural properties of the intrinsic and extrinsic foot muscles. Although several studies examined test-retest reliability, specific gaps remain in assessing inter-rater reliability, particularly distinguishing between image acquisition and muscle measurement. Additionally, these studies utilized equipment that may not be generalizable across both clinical and research settings and often involved small sample sizes without prior sample size calculations. This study aimed to investigate test-retest reliability as well as global and measurement-based inter-rater reliability using a low-end ultrasound device to measure intrinsic and extrinsic foot muscle sizes.

Methods: This prospective reliability study included 21 active individuals. Five intrinsic (abductor hallucis, flexor digitorum brevis, flexor hallucis brevis, quadratus plantae, abductor digiti minimi), and three extrinsic muscles (peroneals, flexor digitorum longus, tibialis anterior) were scanned. Three investigators independently acquired images on two occasions and measured cross-sectional area and thickness in September and October 2023. Participants were assessed either at the Musculoskeletal Research Group laboratory (University of Leuven, Bruges) or in the Rehabilitation Sciences laboratory (Ghent University hospital). Test-retest (same investigator, one week in between), global inter-rater (each investigator measures own image set) and measurement-based inter-rater reliability (three investigators measure one image set) was performed following intra-class correlation, standard error of the measurement and coefficient of variation.

Results: Test-retest reliability showed intraclass-correlation coefficients of 0.60-0.88 for the flexor digitorum brevis and 0.38-0.73 for the tibialis anterior. Standard error of the measurement ranged from 0.16 to 0.41 cm² (cross sectional area) and from 0.05 to 0.31 cm (thickness) for the intrinsic, while they ranged from 0.19 to 1.13 cm² and from 0.12 to 0.44 cm for the extrinsic muscles. Global inter-rater correlation coefficients varied between 0.4 and 0.8 for the abductor hallucis and flexor digitorum brevis. Measurement based inter-rater correlation coefficient varied between 0.50 and 0.96 for abductor hallucis, flexor digitorum brevis, tibialis anterior and peroneal muscles. Standard error of the measurement ranged from 0.14 to 0.89 cm² (cross-sectional area) and from 0.07 to 0.24 cm (thickness) for

the intrinsic, while they ranged from 0.29 to 0.85 cm² (cross-sectional area) and from 0.12 to 0.51 cm (thickness) for the extrinsic muscles. Coefficients of variations were between 4 and 34%. For test-retest, they were consistently $\leq 10\%$ for abductor hallucis thickness, flexor digitorum brevis cross-sectional area, flexor hallucis brevis and tibialis anterior. Flexor digitorum brevis coefficients of variation were $\leq 10\%$ across all inter-rater reliabilities.

Conclusions: Most muscles demonstrated moderate to excellent test-retest reliability using a portable ultrasound device, supporting its generalizability. However, the greater variability in global inter-rater reliability suggests substantial variation during image acquisition. The same clinician should perform pre-intervention and follow-up assessments to minimize errors. If different clinicians are involved, caution is needed when comparing measurements.

Introduction

With each foot strike in locomotion, elastic energy is stored and released through deformation of the medial foot arch,¹⁷ regulated by the intrinsic (IFM) and extrinsic (EFM) foot muscles.^{22, 165} The IFM are small muscles with short moment arms that act as local foot stabilizers, helping to maintain foot posture and control foot deformation.¹⁰ Additionally, they work alongside the plantar aponeurosis to regulate foot stiffness during running.¹⁵³ The EFM, on the other hand, originate from the lower leg and attach to the foot via long tendons, making them global foot movers.¹⁰

IFM and EFM size and strength can be decreased in various disorders such as hallux valgus and diabetes,^{171, 173} and in older people.¹⁷⁴ Therefore, researchers and health care professionals aim to quantitatively evaluate IFM and EFM strength. However, isolating the contributions of each muscle in global toe and ankle strength is impossible with direct assessments such as dynamometry.^{166, 185} Consequently, imaging techniques are typically employed as indirect means of quantitative assessment, with magnetic resonance imaging as the golden standard.^{13, 145} However, it is costly and generally poorly accessible. Alternatively, two-dimensional ultrasound imaging (USI) is frequently used for assessing the structural properties of both IFM and EFM.^{65, 149, 255-256} The measurements obtained with USI are highly similar to those obtained with magnetic resonance imaging, with correlations superior to 0.9.¹⁴⁸

Several studies showed good reliability when quantitatively evaluating cross-sectional area (CSA) and dorso-plantar thickness of various IFM^{145, 148-149, 257} and EFM.²⁵⁷⁻²⁶⁴ Nonetheless, some aspects need to be considered in quantitative ultrasound: Firstly, USI reliability is susceptible to variation in how the image is acquired and then how the measurement is performed. This applies to conditions where there is a repeated measure design including individual therapists separately, however, the situation becomes even more complex when USI evaluations are performed across different observers. Therefore, inter-rater reliability can be fragmented into two aspects: the measurement-based inter-rater reliability (MIRR) and the global inter-rater reliability (GIRR). The MIRR emphasizes the reliability of the image measurement process itself, regardless of who acquired the images. The MIRR has, to our knowledge, only been studied for the tibialis anterior (TA).²⁵⁹ The GIRR replicates clinical contexts, where health care

professionals independently acquire and measure images. To our knowledge, GIRR has been reported for the abductor hallucis (AbH), flexor digitorum brevis (FDB), quadratus plantae (QP), tibialis anterior (TA) and peroneal muscles (PER),^{262, 265-267} but is lacking regarding the abductor digiti minimi (AbDM), flexor hallucis brevis (FHB) and flexor digitorum longus (FDL). Secondly, it is important to note that the majority of these studies utilized high-end USI devices,^{13, 262, 265} a factor that limits the generalizability of their findings to lower-end devices. The cost of USI equipment has been stated to directly relate to the attained images resolution and quality.¹⁵⁰ This therefore indicates that a high-resolution ultrasonography machine produces higher quality images, which are more easily interpreted, especially on small structures, such as the IFM. Moreover, time constraints during image acquisition are rarely discussed in literature. Nevertheless, this represents a significant challenge which clinicians daily face, that could affect the reliability of USI.

Thirdly, it is worth highlighting that only a single study mentioned an a priori sample size calculation,²⁶⁸ and that some of the aforementioned studies adopted small sample sizes of approximately 10 participants, raising concerns about the generalizability of their results.^{258, 261-264} As last consideration, intra-class correlation coefficients (ICC) and standard error of the measurement (SEM) are common reliability metrics. However, there are several ways to calculate ICC and SEM, depending on the study design of the reliability study and the future use of the device in actual practice.²⁶⁹ Some of the published USI reliability studies used inappropriate calculations,^{145, 148, 258-259, 261-263, 266-268, 270} which can lead to overestimations.²⁶⁹

While USI is a promising tool for assessing IFM and EFM size, attention must be paid to study design, equipment quality and sample size to ensure reliable and generalizable results. Therefore, the objectives of this reliability study were: to determine the test-retest reliability of IFM and EFM measurements using a low-end USI device, and to report the inter-rater reliability spanning from MIRR and GIRR. We present this article in accordance with the GRRAS reporting checklist.²⁷¹

Methods

1. Participants

The sample size was estimated based on the lookup table published by Borg (2022)²⁷². AbH and PER CSA were the primary quantitative outcomes of this study, because they are respectively the biggest foot and lower leg muscles assessed in the frame of this study. Based on previous ICC for these muscles ranging between 0.90 and 0.99, an ICC confidence interval width of 0.15,^{258, 265, 268, 273} a significance level of 5% and a power of 80% and 10% drop out, a sample size of 21 participants was deemed sufficient.

A convenience sample (non-probabilistic method) of 21 asymptomatic recreationally active adults was recruited at our two university campuses. “Recreationally active” was defined as the practice of at least 20 minutes of physical activity a day, at least three times a week. Exclusion criteria were foot deformities, neurological, musculoskeletal, or systemic diseases, pregnancy, lower limb injury or surgery in the six months prior participation, and if participants reported recent ankle or foot pain.

Each participant signed the informed consent form. This study was approved by the Ethical Committee UZ/KU Leuven (S67722) and UZ Ghent (number B6702023000319). The study was conducted in accordance with the Declaration of Helsinki (as revised in 2013).

2. Measures and procedures

In this study, each participant was assessed twice either in Bruges at the Musculoskeletal Research Group laboratory (University of Leuven) or in the university hospital of Ghent at the Rehabilitation Sciences laboratory (Ghent University), exactly 7 days apart, respectively session 1 and session 2 (Figure 22). During session 1, demographic data were collected including age, weight, height, sex, shoe size and dominant foot. Participants completed the Foot and Ankle Outcome Score (FAOS) and Baecke questionnaire. The FAOS is a self-reported questionnaire assessing the functional status related to ankle and foot conditions.²⁷⁴⁻²⁷⁵ Each subscale is scored from 0 to 100, with 0 indicating extreme problems and 100 indicating no problems. The Baecke questionnaire assesses physical activity levels in individuals. This questionnaire gathers information on various domains of physical activity, including work-related activities, sports, and leisure-time activities.²⁷⁶⁻²⁷⁷ Furthermore, the six items of the Foot-Posture-Index score were determined by agreement between the investigators,²⁷⁸ and navicular drop was calculated between sit to bipedal stance by one investigator (N.H.).²⁷⁹ This allowed to describe foot posture as this may have an effect on foot muscle sizes.²³⁵

During the two sessions, images of IFM and EFM of the dominant foot were collected using a low-end portable USI system with a 5-12 MHz 40 mm broadband linear array probe (MicrUs EXT-1H, Telemed UAB, Lithuania). Three musculoskeletal physiotherapists (N.H., J.L.PD. and V.S.) independently acquired muscle images with a time constraint of 20 to 30 minutes during the months of September and October 2023. The three investigators had varying USI and clinical experiences: V.S. (30 years, MSc, PhD, 3 year clinical experience) had 3 years of scanning experience while N.H. (25 years, MSc, 1 year clinical experience) and J.L. PD. (26 years, MSc, 1 year clinical experience) had both 1 year scanning experience. The three investigators conducted several pilot testing sessions together to ensure consistency in the protocol.

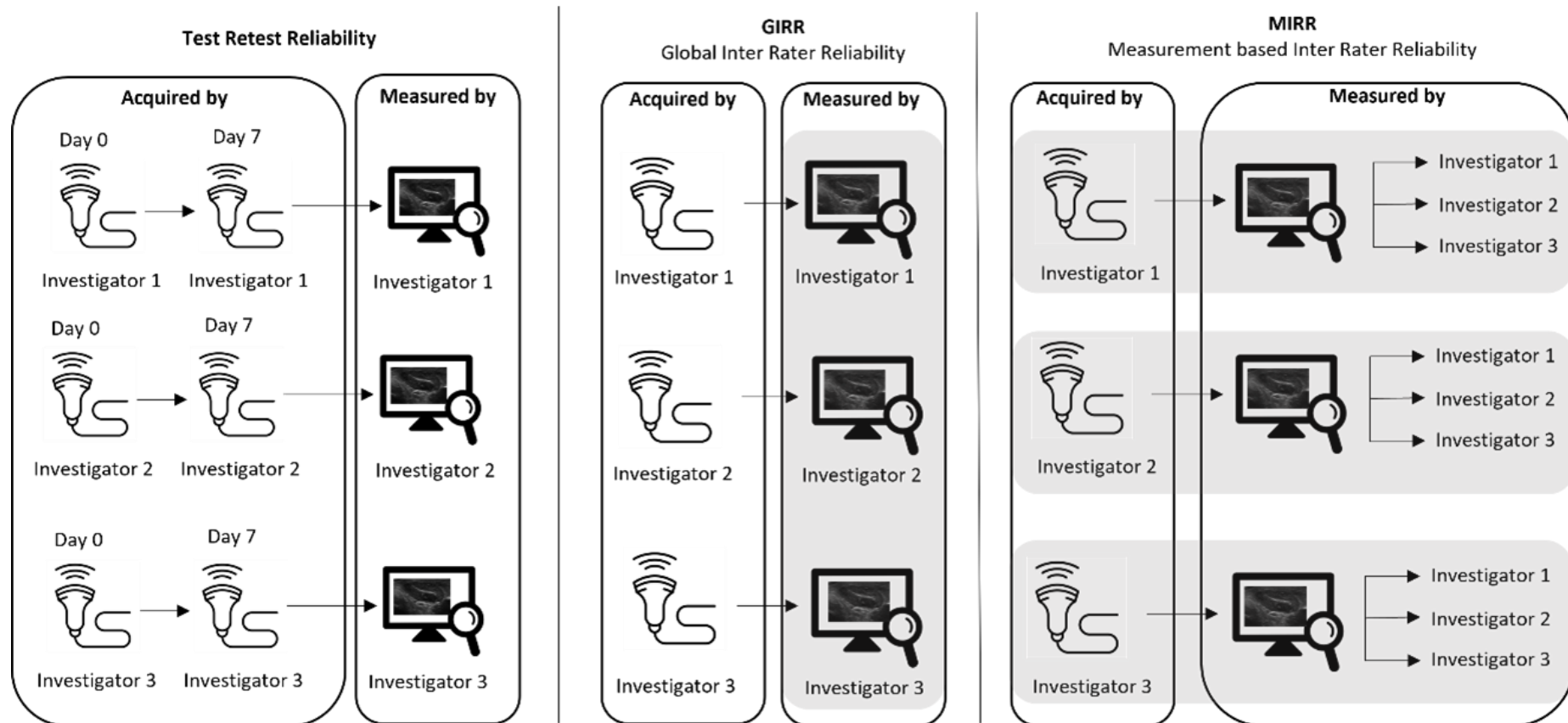


Figure 22. Study set up depicting the acquisition and measurement processes to obtain the three reliability types. Test-retest reliability was calculated for each investigator between day 0 and day 7. The grey zone indicates the measurements used to calculate GIRR (global inter-rater reliability) and MIRR (measurement-based inter-rater reliability): GIRR compares the measurements done by the investigators on their own images and MIRR compares for each of the three image sets the measurements of the three investigators.

The scanning protocol was adapted from previous studies and conducted with the participants lying on their back (i.e. relaxed and non-weight bearing position).^{258, 261, 280} A summary of probe positions and sample images is available in (Figure 23). Each investigator independently acquired transverse (short-axis) and longitudinal (long-axis) images of five IFM (AbH, FDB, FHB, QP, AbDM) and three EFM (PER, FDL and TA). Longitudinal imaging of the FHB was not possible due to the indefinite appearance of this muscle in the frontal plane. Longitudinal imaging of the TA was not possible as the muscle did not consistently fit within the USI display. The depth, focal point, frequency, and gain were adjusted for each participant and each muscle to optimize quality of image capturing. In case of doubt, subjects were instructed to perform specific movements to elicit muscular contraction, aiding the investigator in identifying muscle borders. These contractions were brief and performed without resistance to minimize the risk of fluid shifts significantly affecting muscle size. For each muscle, its relaxed state was captured in a cine-loop lasting up to 5 seconds. The investigator then saved a single image, before scanning the next muscle view. Muscle size measurement was performed in the days following image acquisition to avoid recall bias. During this step, investigators were blinded to participant information and to the measurements taken by other investigators. For the GIRR, each investigator individually measured muscle sizes using their own set of acquired images. For the MIRR, muscle measurement was performed by the three investigators on the image set acquired by one investigator. The image sets were taken from the second session and were measured using still images in Image J software (National Institute for Health, Bethesda, MD, USA). Previous studies used this software to measure muscle CSA and thickness from ultrasound images and demonstrated excellent reliability.^{258, 261, 281} Transverse-view and longitudinal-view images were respectively used to obtain CSA (cm²) and thickness (cm). Thickness was averaged from three measurements at the thickest part of the muscle on the image.

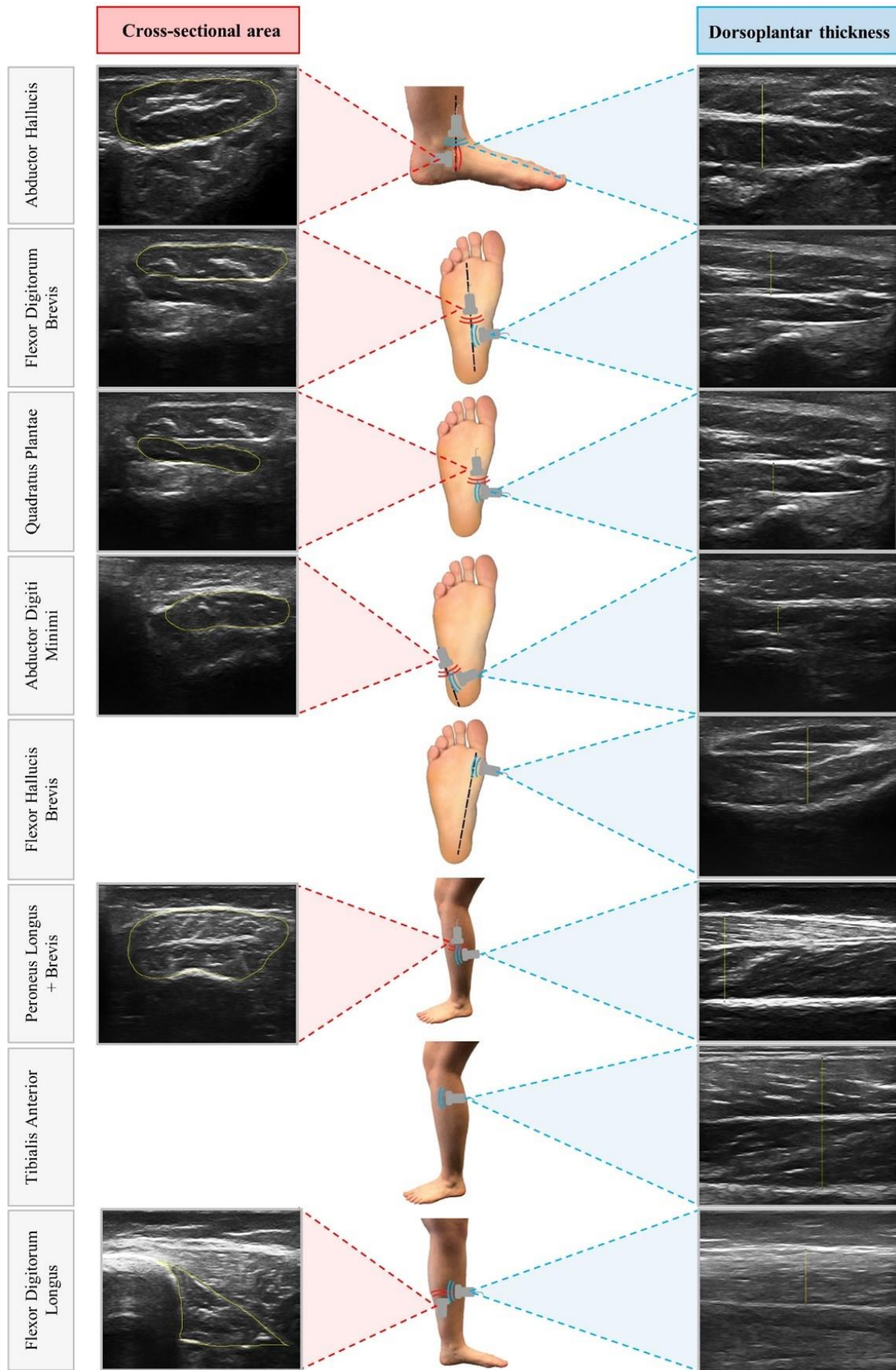


Figure 23. Summary of probe positions with sample images. (continued on the next page →)

Probe position during acquisition of transverse (red) and longitudinal (blue) images of the intrinsic and extrinsic foot muscles. The following landmarks were used during the acquisition: Abductor hallucis was scanned on a vertical line passing in front of the medial malleolus, Flexor digitorum brevis was scanned on a line passing through the third toe and the medial tubercle of the calcaneus, Quadratus plantae was scanned more medial than the scanning zone of flexor digitorum brevis and near the talo-calcaneo-navicular joint, Abductor digiti minimi was scanned between lateral tuberosity of the calcaneus and tuberosity of the 5th metatarsal, Flexor hallucis brevis was scanned on a line parallel to the muscle orientation, i.e. medio-plantar face of the first metatarsal bone, Peroneus brevis and longus were scanned together at the mid-distance between the fibular head and the lateral malleolus, Tibialis anterior was scanned at 20% of the distance between the fibular head and the two malleolus, Flexor digitorum longus was scanned at mid-distance between the medial tibial plateau and the medial malleolus.

3. Statistical analysis

Statistical analyses were performed in SPSS for Windows v27 (IBM SPSS Inc. Chicago, IL). Means and standard deviations were reported for age, height, weight, shoe size, navicular drop, FAOS, Baecke questionnaire and muscle measurements. Foot posture index scores were interpreted as neutral (between 0 and 5), pronated (higher than 5), extremely pronated (higher than 9), supinated (lower than 0) and extremely supinated (lower than -4).^{242, 278}

ICC and corresponding 95% confidence intervals were calculated. According to McGraw and Wong convention, test-retest reliability was calculated as single measure, absolute agreement, two-way mixed model while GIRR and MIRR were calculated using single measure, absolute agreement, two-way random effects model. These use the same formula and referred to as ICC_{2,1} in Shrout and Fleiss convention (Equation 6), but the interpretation of their results is different:²⁶⁹

$$\text{Equation 6: } ICC_{2,1} = \frac{MSR - MSE}{MSR(k-1)MSE + \frac{k}{n}(MSC - MSE)}$$

where MSR is the mean square rows, MSE is the mean square error, MSC is the mean square columns, k is the number of raters (=3) and n the number of participants (=21).

The scanning resulted in three sets of images (one per investigator). Test-retest reliability was considered between the two sessions held one week apart. For inter-rater, only the images of the second session were used. GIRR was calculated with each investigator measuring muscle sizes on their own images and MIRR was obtained with the three investigators interpreting muscle sizes on USI images on one image set. Therefore, three test-retest and three MIRR were obtained for each muscle. ICC were interpreted as follow: <0.5 poor, ≥0.5 to <0.75 moderate, ≥0.75 to <0.9 good, ≥0.9 excellent.²⁶⁹ Standard error of the measurement, agreement ($SEM_{agreement}$) were then calculated (Equation 7):²⁸²

$$\textbf{Equation 7: } SEM_{agreement} = \sqrt{(\sigma_{lp}^2 + \sigma_{residual}^2)}$$

where σ^2 is the variance, “IP” the interaction between the investigators and the participants and “residual” refers to the error.

The coefficient of variation (CV) was calculated for the three reliability types. The CV of the “i”th triplet of measurements (i.e. the measurements of the three investigators) is given by (Equation 8):²⁸³

$$\textbf{Equation 8: } CV(\%) = 100 \frac{\sum CV_i}{n}, \text{ with } CV_i = \frac{SD_i}{\bar{x}_i}$$

where n is the number of participants (=21), CV the coefficient of variation, $\bar{x}$ the mean of the measurements, SD the standard deviation and i refers to one “i”th triplet of measurements.

Results

The 21 participants were scanned for the five IFM and three EFM by the three investigators on two occasions, as per protocol. This included 10 participants recruited from the Bruges campus of KU Leuven University and 11 participants from the University Hospital of Ghent.

Descriptive characteristics are available in (Table 8). Participants were young adults, with an even distribution between males and females. Five participants had a pronated foot (23.8%), five participants (23.8%) had a supinated foot. The remaining eleven participants had a neutral foot (52.4%). All were physically active and pain free, as confirmed by the Baecke and FAOS scores. No significant differences in participant demographics were observed between the two scanning sites, ensuring consistency in the sample.

Table 8: Participant characteristics (n = 21)

Variable	Values
Age (years)	25.9 ± 2.47
Sex (M/F)	10/11
Body weight (kg)	70.68 ± 12.29
Height (m)	1.76 ± 0.09
Shoe size (European)	40.2 ± 3.08
Dominant foot, right (%)	18 (85.7%)
Foot Posture Index (-12 to 12 score)	4 ± 2.5
Neutral (score 0 to 5) (%)	11 (52.4%)
Pronated (score 6 to 9) (%)	5 (23.8%)
Extremely pronated (score 10 to 12) (%)	0 (0%)
Supinated (score -1 to -4) (%)	5 (23.8%)
Extremely supinated (score -5 to -12) (%)	0 (0%)
Navicular drop (mm)	5.2 ± 1.1
Baecke questionnaire (scores vary from 0 to 15)	10.02 ± 1.72
Foot and Ankle Outcome Score (FAOS, %)	96.16 ± 3.35

Legend: Data are presented as mean ± standard deviation or number of participants (%).

4. Test-retest reliability

4.1. *Cross-sectional area*

Test-retest reliability (Table 9) between sessions 1 and 2 was moderate to good for the three investigators for AbH (ICC 0.53-0.86) FDB (ICC 0.81-0.88), and QP (ICC 0.6-0.75). Estimated ICC were good in two investigators for PER (0.79-0.83). Estimated ICC were poor to moderate in all three investigators for AbDM and FDL. CV were lower or equal to 10% for the three investigators for FDB, and the other measurements had CV lower than 20%. SEM of the IFM ranged from 0.16 to 0.36 cm² for rater 1, from 0.18 to 0.31 cm² for rater 2, and from 0.19 to 0.41 cm² for rater 3. SEM of the EFM ranged from 0.46 to 1.13 cm² for rater 1, from 0.25 to 0.44 cm² for rater 2, and from 0.19 to 0.46 cm² for rater 3.

4.2. *Thickness*

Test-retest reliability (Table 9) between sessions 1 and 2 was moderate to good in the three investigators for AbH (ICC 0.51-0.78), FDB (ICC 0.6-0.8), and QP (ICC 0.51-0.87). Estimated ICC were reported to be poor in at least one investigator for AbDM, FHB, PER, TA. CV were lower or equal to 10% for the three investigators for AbH, FHB and TA, and the other measurements had CV lower than 16%. SEM of the IFM ranged from 0.05 to 0.31 cm for rater 1, from 0.09 to 0.17 cm for rater 2, and from 0.07 to 0.16 cm for rater 3. SEM of the EFM ranged from 0.18 to 0.44 cm for rater 1, from 0.12 to 0.22 cm for rater 2, and from 0.17 to 0.26 cm for rater 3.

Table 9: Test-retest reliability with mean muscle size

			Investigator 1				Investigator 2				Investigator 3			
	Muscles	Measure	Mean size (SD)	ICC [95% CI]	CV	SEM	Mean size (SD)	ICC [95% CI]	CV	SEM	Mean size (SD)	ICC [95% CI]	CV	SEM
Intrinsic Foot Muscles	AbH	CSA	2.4 (0.69)	0.78 [0.53-0.9]	10%	0.29	2.5 (0.69)	0.86 [0.42-0.95]	10%	0.27	2.65 (0.73)	0.53 [0.15-0.78]	12%	0.41
		Thickness	1.08 (0.2)	0.68 [0.36-0.86]	9%	0.11	1.17 (0.22)	0.78 [0.54-0.9]	6%	0.11	1.22 (0.18)	0.51 [0.1-0.77]	8%	0.14
	FDB	CSA	2.03 (0.47)	0.88 [0.74-0.95]	7%	0.16	2.2 (0.5)	0.85 [0.67-0.94]	6%	0.18	2.02 (0.42)	0.81 [0.58-0.92]	7%	0.19
		Thickness	1.04 (0.22)	0.6 [0.23-0.82]	11%	0.15	1.02 (0.19)	0.8 [0.57-0.92]	6%	0.09	1.04 (0.18)	0.79 [0.54-0.91]	7%	0.09
	QP	CSA	0.88 (0.26)	0.68 [0.37-0.86]	19%	0.17	1.24 (0.55)	0.75 [0.48-0.89]	18%	0.26	1.44 (0.33)	0.6 [0.23-0.82]	15%	0.24
		Thickness	0.6 (0.13)	0.87 [0.7-0.94]	7%	0.05	0.69 (0.18)	0.51 [0.1-0.77]	16%	0.13	0.83 (0.14)	0.74 [0.46-0.89]	7%	0.07
	AbDM	CSA	1.52 (0.46)	0.28 [0-0.6]	18%	0.36	1.57 (0.34)	0.18 [0-0.56]	16%	0.31	1.04 (0.22)	0.56 [0.19-0.79]	10%	0.19
		Thickness	0.7 (0.23)	0.66 [0.33-0.85]	15%	0.13	0.9 (0.23)	0.29 [0-0.64]	14%	0.17	0.92 (0.21)	0.82 [0.6-0.92]	9%	0.09
	FHB	Thickness	1.78 (0.35)	0.06 [0-0.47]	10%	0.31	1.55 (0.22)	0.79 [0.54-0.91]	5%	0.11	1.49 (0.21)	0.5 [0.1-0.76]	8%	0.16
	Extrinsic Foot Muscles	PER	CSA	5.35 (1.4)	0.32 [0-0.66]	17%	1.13	3.97 (1.01)	0.83 [0.64-0.93]	9%	0.44	4.29 (1.05)	0.79 [0.53-0.91]	7%
Thickness			2.45 (0.58)	0.21 [0-0.58]	14%	0.44	1.77 (0.35)	0.55 [0.15-0.79]	9%	0.22	2.02 (0.41)	0.79 [0.55-0.91]	7%	0.17
FDL		CSA	1.75 (0.61)	0.18 [0-0.56]	20%	0.46	1.76 (0.4)	0.58 [0.23-0.8]	9%	0.25	1.55 (0.35)	0.71 [0.41-0.87]	10%	0.19
		Thickness	1.12 (0.36)	0.75 [0.48-0.89]	14%	0.18	1.06 (0.17)	0.6 [0.23-0.81]	9%	0.12	1.74 (0.23)	0.02 [0-0.45]	12%	0.26
TA		Thickness	2.55 (0.47)	0.38 [0-0.69]	10%	0.36	2.42 (0.32)	0.67 [0.35-0.85]	7%	0.21	2.34 (0.37)	0.73 [0.45-0.88]	6%	0.19

Legend: Mean and SD muscle sizes were calculated from the second day of scanning. Negative lower ends of ICC confidence intervals were adjusted to zero. SD, standard deviation; ICC, intraclass correlation coefficients; CI, confidence interval; CV, coefficient of variation; SEM, standard error of the measurement; AbH, abductor hallucis; FDB, flexor digitorum brevis; QP, quadratus plantae; AbDM, abductor digiti minimi; FHB, flexor hallucis brevis; PER, peroneal; FDL, flexor digitorum longus; TA, tibialis anterior; CSA, cross-sectional area.

5. Inter-rater reliability

5.1. *Cross-sectional area*

GIRR (Table 10) was good for FDB (ICC = 0.80) and moderate for AbH (ICC = 0.74). Inter-rater reliability of the other muscles was poor. The CV were lower than 10% for FDB, and lower or equal to 20% for AbH, PER, and FDL. It reached 30% for QP. SEM of the IFM ranged from 0.21 to 0.43 cm², while SEM of EFM ranged from 0.39 to 0.99 cm².

MIRR was good to excellent for the three image sets for AbH (ICC 0.80-0.96) and FDB (ICC 0.81-0.93), and moderate to good for PER (ICC 0.67-0.81). MIRR was poor to moderate in the three image sets for QP (ICC 0.32-0.54) and FDL (ICC 0.24-0.59). CV was lower or equal to 10% for AbH and FDB. It was lower or equal to 20% for AbDM, PER and FDL. SEM of the IFM ranged from 0.14 to 0.89 cm², while SEM of EFM ranged from 0.29 to 0.85 cm².

5.2. *Thickness*

GIRR (Table 10) was moderate for FDB (ICC = 0.69). It was poor for the other muscles. The CV was lower than 10% for FDB, and lower than 20% for AbH, FHB, PER, TA. It reached 30% for FDL. SEM of the IFM ranged from 0.16 to 0.58 cm, while SEM of EFM ranged from 0.29 to 0.45 cm.

MIRR was moderate to good for AbH (ICC, 0.67-0.89), PER (ICC, 0.50-0.82), and TA (ICC 0.70-0.88). MIRR was poor to moderate for FDB (ICC, 0.49-0.69), QP (ICC, 0.03-0.51) and AbDM (0.50-0.61). Poor reliability was found for FDL (ICC, 0.05-0.14). CV was lower than 10% for the three image sets for AbH, FDB, FHB, PER, and TA. It was lower than 20% for AbDM and QP. SEM of the IFM ranged from 0.12 to 0.18 cm, while SEM of EFM ranged from 0.11 to 0.33 cm.

Table 10: Global inter rater reliability (GIRR) and measurement based inter rater reliability (MIRR)

			GIRR			MIRR			MIRR			MIRR		
						Image set investigator 1			Image set investigator 2			Image set investigator 3		
	Muscles	Measure	ICC [95% CI]	CV	SEM	ICC [95% CI]	CV	SEM	ICC [95% CI]	CV	SEM	ICC [95% CI]	CV	SEM
Intrinsic Foot Muscles	AbH	CSA	0.74 [0.54-0.87]	12%	0.37	0.88 [0.77-0.94]	8%	0.24	0.96 [0.91-0.98]	5%	0.14	0.8 [0.64-0.91]	10%	0.89
		Thickness	0.4 [0.14-0.65]	12%	0.16	0.67 [0.44-0.83]	7%	0.12	0.89 [0.79-0.95]	4%	0.07	0.76 [0.58-0.88]	5%	0.1
	FDB	CSA	0.8 [0.61-0.91]	9%	0.21	0.81 [0.66-0.91]	7%	0.21	0.92 [0.82-0.96]	6%	0.15	0.93 [0.87-0.97]	5%	0.33
		Thickness	0.69 [0.48-0.85]	9%	0.3	0.69 [0.48-0.85]	8%	0.13	0.67 [0.45-0.84]	6%	0.1	0.49 [0.23-0.73]	8%	0.14
	QP	CSA	0.28 [0.03-0.56]	31%	0.41	0.32 [0.06-0.6]	26%	0.35	0.51 [0.22-0.75]	23%	0.34	0.54 [0.28-0.76]	21%	0.32
		Thickness	0.17 [0-0.44]	22%	0.47	0.51 [0.24-0.74]	15%	0.14	0.32 [0.07-0.59]	16%	0.13	0.03 [0-0.29]	18%	0.18
	AbDM	CSA	0.12 [0-0.36]	26%	0.43	0.91 [0.76-0.96]	8%	0.14	0.46 [0.19-0.7]	15%	0.26	0.46 [0.2-0.7]	15%	0.22
		Thickness	0.3 [0.06-0.58]	22%	0.58	0.61 [0.35-0.8]	15%	0.14	0.5 [0.24-0.73]	16%	0.2	0.53 [0.28-0.75]	16%	0.17
	FHB	Thickness	0.1 [0-0.37]	13%	0.29	0.47 [0.21-0.71]	6%	0.24	0.76 [0.55-0.89]	5%	0.11	0.6 [0.36-0.8]	9%	0.24
Extrinsic Foot Muscles	PER	CSA	0.48 [0.11-0.75]	18%	0.99	0.67 [0.32-0.85]	15%	0.85	0.81 [0.62-0.92]	12%	0.5	0.76 [0.41-0.9]	11%	0.58
		Thickness	0.37 [0.05-0.66]	18%	0.45	0.65 [0.38-0.83]	10%	0.33	0.5 [0.24-0.73]	8%	0.51	0.82 [0.67-0.91]	7%	0.2
	FDL	CSA	0.32 [0.06-0.6]	20%	0.39	0.35 [0.08-0.62]	20%	0.5	0.24 [0-0.54]	18%	0.47	0.59 [0.35-0.79]	13%	0.29
		Thickness	0 [0-0.12]	34%	0.46	0.14 [0-0.41]	22%	0.31	0.06 [0-0.27]	30%	0.41	0.05 [0-0.27]	28%	0.45
	TA	Thickness	0.35 [0.09-0.62]	11%	0.33	0.7 [0.5-0.85]	7%	0.23	0.88 [0.77-0.95]	4%	0.12	0.75 [0.51-0.89]	5%	0.2

Legend: Mean and SD muscle sizes were calculated from the second day of scanning. Negative lower ends of ICC confidence intervals were adjusted to zero. GIRR, global inter-rater reliability; MIRR, measurement-based inter-rater reliability; ICC, intraclass correlation coefficients; CI, confidence interval; CV, coefficient of variation; SEM, standard error of the measurement; AbH, abductor hallucis; FDB, flexor digitorum brevis; QP, quadratus plantae; AbDM, abductor digiti minimi; FHB, flexor hallucis brevis; PER, peroneal; FDL, flexor digitorum longus; TA, tibialis anterior; CSA, cross-sectional area.

Discussion

Quantitative ultrasound is frequently used in research and health-care contexts to measure the sizes of IFM and EFM, including in patients with hallux valgus,^{171, 284} painful pes planus,²⁸⁵ chronic ankle instability,⁶⁵ and diabetic neuropathy.^{173, 286} This study aimed to determine its test-retest and inter-rater reliability in comparable settings using a low-end USI device. For the majority of the muscles, we demonstrated moderate to good test-retest reliability, poor to good GIRR, and poor to excellent MIRR.

Test-retest reliability was moderate to good for the three investigators for the CSA and thickness of AbH, FDB and QP in our study. For the other measures, at least two of the three investigators of our study had moderate or good ICC, except for AbDM CSA where two investigators had poor reliability. These findings thus differ from previous studies which consistently reported good to excellent test-retest reliability for IFM and EFM sizes,^{148, 257, 260-261, 266-268} and moderate to excellent reliability for TA thickness.²⁵⁹⁻²⁶⁰ Still, at least two of the three investigators of our study had moderate or good ICC for a given structure, except for AbDM CSA where two investigators had poor reliability. To our knowledge, this is the first report of IFM or EFM test-retest reliability involving two or more raters who all performed both the image acquisition and the measurement. Indeed, in previous test-retest studies that involved 2 or 3 raters, the investigators did the image acquisition once and then performed the measurement twice on that image,^{259, 265} or were limited to pathological populations.²⁶³ Selecting three raters is also in accordance with guidelines published by Koo & Li.²⁶⁹

MIRR was good to excellent for the three image sets for the CSA of AbH and FDB, and at least moderate for the thickness of AbH, AbDM, TA, and PER CSA and thickness. This aligns with a previous study reporting moderate to excellent MIRR for TA thickness (ICC 0.64–0.99).²⁵⁹

Moderate to good GIRR was found for AbH CSA, FDB CSA, and FDB thickness. Previous GIRR results have varied from moderate to excellent for IFM and EFM, with moderate GIRR reported for AbH, FDB, and QP thickness.^{262, 265} As expected, GIRR was generally worse compared to MIRR. Differences between investigators become more pronounced when both acquisition and measurement are done individually, suggesting that scanning is more dependent on the investigator, while measurements in Image J are easier to standardize due to agreed guidelines and

well defined muscle borders. This standardization is facilitated by several factors: the controlled environment in which image evaluation is often performed minimizes external variability; software tools can automate measurement processes, ensuring consistent results; and defined protocols for evaluation provide clear steps that can be universally applied. The availability of reference images for comparison also aids standardization, as raters can align their measurements against established benchmarks. Additionally, training and calibration focused on evaluation techniques can help standardize the process across different raters, while consensus processes allow for collaborative assessment of images, further enhancing reliability. Clinicians should therefore be cautious when comparing muscle sizes acquired by different peers. Similarly to MIRR, previous GIRR results have been reported to vary from moderate to excellent for IFM and EFM. More particularly, Battaglia et al. reported moderate GIRR for the thicknesses of AbH, FDB and QP.²⁶⁵ Our study is the first to investigate GIRR for FHB, AbDM, and FDL, showing poor reliability which needs to be interpreted cautiously until confirmation by future studies using similar low-end USI devices.

While ICC and CV are relative estimates of reliability providing insights on individuals maintaining their position in a sample over repeated measurement, SEM is an absolute estimate of reliability which express the variety of repeated measurements for an individual (for more detailed explanations, see Baumgartner 1989).²⁸³ According to Hopkins et al., CV equal or lower than 10% are considered reasonable.²⁸⁷ Therefore, the CV we obtained align well with the moderate to good test-retest, MIRR and GIRR ICC of AbH and FDB. Although an acceptable CV was generally associated with a moderate or good ICC, this was not always true as for example test-retest ICC of FHB and TA were poor for one of the investigators with a CV equal to 10%. CV related to GIRR were overall not acceptable, whereas CV of MIRR reached 5% for many structures.

We also calculated $SEM_{\text{agreement}}$, an outcome quantifying the uncertainty associated with USI measurements. SEM values indicate how representative the measured size is of the true size, with smaller SEM indicating that the measured size is more likely to be representative of the true size, increasing confidence in clinical decisions. In this study, $SEM_{\text{agreement}}$ ranged from 0.12 to 0.89 cm² for CSA and from 0.05 to 0.58 cm for thickness of IFM. For EFM, SEM values ranged from 0.19 to 1.13 cm² for CSA and from 0.11 to 0.51 cm for thickness. These values are

comparable to some previous studies for test-retest,²⁶⁶ and inter-rater reliability.^{263, 266} Although others reported impressive SEM values, lower than a millimetre.^{148, 259, 263-264, 268} Consistent with ICC results, SEM values were generally higher for GIRR than for MIRR, indicating greater variability in more clinically relevant settings.

Overall the majority of structures with moderate to excellent test-retest reliability highlights its generalizability, despite the use of a low-end USI device. However, the results obtained for the two perspectives on inter-rater reliability should raise the attention of people using USI in the intrinsic and extrinsic muscles of the foot. GIRR was indeed generally worse compared to MIRR, indicating greater variability in more clinically relevant settings. Importantly, the poor global inter-rater reliability observed is not attributable to varying levels of scanning or clinical experience, as there was no consistency in one rater performing better than the other. This highlights that errors can be performed during the image acquisition because of the difficulties that clinicians face during this process. For example, identifying the thickest muscle part of a muscle involves subjectivity. On the other hand, measurement in Image J is easier to standardize due to agreed guidelines and often well-defined muscle borders on the image. Two clinical applications related to orthopaedic interventions arise from these findings: First, the pre-intervention and follow-up assessments of muscle atrophy or hypertrophy in the frame of an orthopaedic intervention should be performed by the same clinician in order to minimize the sources of errors. Second, if the follow-up is performed by another clinician than the pre-intervention evaluation, caution is needed when comparing muscle sizes acquired by these peers.

The mean muscle sizes for AbH, FDB, and FHB align with reference values from a recent systematic review by Haelewijs et al. encompassing a large participant pool. This emphasises our consistency in the use of the ultrasound protocols. For example, our AbH CSA (2.40, 2.50 and 2.65 cm²) was similar to the reference value (2.43 cm²) reported in a cohort of 781 subjects.¹⁴⁹ Likewise, mean TA thickness (2.34, 2.42, 2.55 cm) also closely matched literature values (2.49 cm), while mean AbDM CSA and thickness are similar to the values reported by Swanson et al.,¹⁴⁸ despite another study finding higher values.²⁶¹ AbDM thickness,²⁶⁰⁻²⁶¹ QP CSA,¹⁴⁸⁻¹⁴⁹ and FDL thickness values are in the lower range of what could be expected from the literature,²⁶⁴ while mean QP thickness values were lower to

references values.¹⁴⁹ PER CSA and thickness values are higher compared to those reported elsewhere (approximatively 4 cm² and 1.4 cm, respectively).^{258, 260, 273} FDL CSA values face high discrepancy in previous articles,^{258-259, 280} and our results are comparable to those of Johnson et al.'s study.²⁶⁴ However, most of the aforementioned EFM size estimates are based on few cases (i.e. 10 to 28 participants), which may not be enough to result in valid norms,^{258-260, 264} or with PER separated into longus and brevis,²⁷³ which makes comparison difficult.

Differences between our findings and previous studies may be due to our attempt to emulate real clinical settings. Firstly, the utilization of a low-cost USI device represents a notable difference from the methodologies used in previous studies.^{148, 258, 261-262, 265-266} These devices typically provide lower image resolution and quality compared to high-end systems.²⁸⁸ Secondly, our study imposed a limited scanning time of 20-30 minutes per individual due to our pre-defined schedule, mimicking real-life clinical practice. This time constraint may have resulted in the acquisition of suboptimal images, as prior studies often do not specify such limits and may allow for longer examination times to optimize image quality. Also, although we performed a cine loop, we only registered a single image per muscle for evaluation. In contrast, prior studies may have had the opportunity to select multiple or higher-quality images for measurement, reducing variability. Lastly, we did not calculate the average of several independent measurements, which usually results in higher reliability.²⁶⁹

Our method to calculate ICC and SEM may also have influenced reliability estimates. Indeed, the calculation of ICC and SEM vary by model, type, and definition. According to Koo and Li,²⁶⁹ test-retest ICC should be calculated with “two-way mixed effects” model and “absolute agreement” definition. For inter-rater ICC, the model depends on the study design and clinical applications, while “absolute agreement” definition is preferred. Similarly, SEM_{agreement} should be preferred over SEM_{consistency}, because the exact muscle size matters more than the consistent differences between measurements.²⁶⁹ This was our rational to report “single measures, absolute agreement” ICC and SEM, hypothesized to better reflect clinical practice,²⁶⁹ while some previous studies used multiple measures^{264-265, 267} or consistency^{258, 262} models.

Our results should be interpreted in the light of the study. Sample size calculation was based on the hypothesis that the ICC results would be superior to 0.9.

However, the reliabilities associated with our primary outcomes were lower, and although the selected sample size allowed to expect confidence interval widths of less than 0.15, this was achieved in only two results. Hence, the desired power of this study is not guaranteed. Besides, although the investigators performed extended study piloting, they had approximately 1 year in USI experience, probably influencing the results. Despite these limitations, this study enhances our understanding of the generalizability of low-end USI in measuring IFM and EFM sizes.

Conclusion

This study investigated the reliability of ultrasound imaging for measuring intrinsic and extrinsic foot muscles across both clinical and research settings, using a low-end device. Our findings indicate moderate to good test-retest reliability, poor to good global inter-rater reliability, and poor to excellent measurement-based inter-rater reliability. While test-retest reliability was generally favourable, global inter-rater reliability tended to be lower, suggesting greater variability between investigators, particularly in image acquisition. Our results emphasize the need for caution when comparing muscle sizes acquired by different investigators, as variability in image acquisition can definitely impact reliability. Clinicians should be mindful of these limitations when using USI for muscle size assessment and should receive appropriate training. This training should transcend the common sources of technical error associated to the ultrasound device itself (depth, focus, frequency, intensity) by making clinicians aware of the intricacies at both the scanning and measurement stages. Future studies mimicking clinical settings are an essential next step in confirming generalizability of ultrasound imaging.

Acknowledgements

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Summary

Context

Many studies investigated test-retest and inter-rater reliability of ultrasound imaging to measure foot muscle sizes, as highlighted in our previously published systematic review.¹⁴⁹ Still, most of these reliability studies suffered from limitations including the lack of sample size calculation, restricted sample sizes, and optimized settings that may overestimate reliability compared to more ecological situations. These settings included the use of high end devices, placing landmarks on skin that remained visible between raters, and test-retest or inter-rater measurements being performed on the same images.

Findings

We determined an a priori sample size of 21 participants. Three raters independently acquired images of intrinsic and extrinsic foot muscles using a low-end ultrasound device. They also had time constraints to perform that acquisition and placed no visible landmark on the skin of participants. Muscle size measurements were performed later.

Test-retest reliability (7 days apart) was moderate to good for most structures. Similarly, inter-rater reliability was moderate to good for most measures when raters measured muscle sizes on the same images, removing variability due to image acquisition. Contrarily, reliability was poor for most structures for inter-rater reliability that combined image acquisition and measurement.

Implications for clinical practice and future research

These results – along with those of previous reliability studies – support the clinical applicability of a low-end ultrasound device to measure foot muscle sizes in a test-retest design. This can for example be useful to track the progression of patients when following interventions targeting the foot muscles. However, our results also highlight the need for strictly agreed guidelines when measures are intended to be compared between raters.

Clinical studies

Chapter 4. Side-step cutting foot biomechanics and intrinsic foot muscle sizes in people with chronic ankle instability

Jean-Louis Peters-Dickie, Christine Detrembleur, Sébastien Lobet, Kevin Deschamps

Abstract

Introduction: Lateral ankle sprains commonly occur during directional changes and often lead to chronic ankle instability (CAI), a condition associated with altered movement patterns, muscle weakness, and reduced lower limb muscle sizes. This study aimed to compare foot joint biomechanics during side-step cutting among individuals with CAI, copers, and uninjured controls, and to explore the relation between intrinsic foot muscle sizes and functional status.

Methods: Cross-sectional study. Sixty-nine recreationally active males (25.4 ± 4 years) were categorized based on perceived instability and self-reported function. Ankle and foot joint kinematics and kinetics were assessed using a multi-segment foot model during a 45° side-step cutting task, performed at a mean approach speed of $3.5 \text{ m}\cdot\text{s}^{-1}$. Group comparisons were conducted using ANOVA or Kruskal-Wallis tests, and continuous biomechanical data were analysed using statistical parametric or non-parametric mapping. Cross-sectional area and thickness of the abductor hallucis and flexor digitorum brevis were recorded using ultrasound imaging. Muscle sizes were explored using general linear models, and correlations with questionnaire scores and physical examination outcomes were calculated.

Results: Foot joint kinematics and kinetics did not differ significantly across groups during the side-step cutting task. Individuals with CAI exhibited lower peak posterior ground reaction forces compared to controls (mean difference: 6% of body weight, $p = 0.03$). Statistical parametric mapping revealed lower posterior ground reaction forces between 22–46% of stance in CAI versus controls ($p < 0.05$). Intrinsic foot muscle sizes showed weak but significant correlations with perceived instability and self-reported disability.

Conclusion: Despite having pronounced sensory-perceptual impairments, individuals with CAI demonstrated few differences in foot joint biomechanics compared to copers and controls during an anticipated side-step cutting task performed at moderate speed. Reduced braking forces may reflect altered deceleration strategies. These findings highlight the relevance of assessing deceleration abilities in people with CAI and underscore the potential role of foot muscles in CAI-related functional outcomes.

Introduction

Lateral ankle sprain (LAS) is the most common injury in physically active individuals.³³ Approximately 40% of individuals who sustained a LAS develop chronic ankle instability (CAI),³³ whereas those remaining asymptomatic are termed copers.⁴⁰ CAI is defined by recurrent sprains, ankle giving way episodes and perceived instability.³¹ Individuals with CAI and copers both sustained an initial ankle sprain, which is why direct comparisons between these populations are essential to understand the mechanisms underlying CAI development.⁴⁰

Hertel and Corbett's model summarizes the impairments involved in CAI.³⁹ Motor-behavioural impairments include altered movement patterns during locomotion and sporting tasks, balance deficits, and muscle weakness. Sensory-perceptual impairments involve perceived instability, lower self-reported function, and kinesiophobia. Pathomechanical impairments encompass structural deficits, among which reduced lower leg muscle sizes, and possibly lower foot muscle sizes.^{39, 62}

Sports that involve frequent changes of direction are at risk of LAS due to the vulnerable position of the ankle.¹²⁸⁻¹²⁹ Biomechanical studies found higher plantarflexion moment in early stance, lower ankle eversion moment at mid-stance,²⁸⁹ and higher ankle inversion position in late stance of side cutting tasks in people with CAI compared to controls.¹³² However, modelling the foot as a single segment may over-estimate ankle kinematics and kinetics,^{104, 290} and multi-segment foot models were therefore developed.^{109, 112, 290} To our knowledge, no study compared side-step cutting foot joint kinematics or kinetics among CAI, copers, and controls. Kunugi et al. found more hindfoot external rotation in late stance in CAI with severe laxity versus CAI without laxity,¹²⁷ and similar findings between copers with versus without laxity.¹²⁵ Besides, other studies found sagittal and frontal plane kinematic differences at the hindfoot and forefoot during running between CAI and controls.⁸⁴⁻⁸⁵ The arthrokinematic and osteokinematic restrictions associated with CAI³⁹ likely impair foot joint kinetics, as it was found in other foot joint disorders.¹¹⁷⁻¹¹⁹ Therefore, investigating foot joint biomechanics during side-step cutting may provide insights into functional alterations and assist in developing rehabilitation strategies for individuals with LAS and CAI.

Foot function is supported by both extrinsic and intrinsic foot muscles (IFMs), the latter originating within the foot.^{10, 12} The IFMs, with the abductor hallucis and flexor digitorum brevis being among the biggest, attenuate foot impact and support push-off during locomotion,^{22, 153} along with their proprioceptive roles.²⁷ Ultrasound imaging reliably assesses IFM sizes,²⁹¹ a surrogate of muscle strength,¹⁴⁵ whereas dynamometry is reliable to directly measure hallux and toe flexion strength.^{139, 166} Although previous research demonstrated reduced ankle strength in multiple directions among individuals with CAI,⁷⁸ few studies examined muscle sizes and strength more distally, at the foot level.^{62, 65} Feger et al. reported smaller IFM sizes in individuals with CAI compared to controls, among a sample of 9 participants.⁶² Fraser et al. observed deficits in hallux and toe flexion strength in CAI relative to both copers and uninjured controls, but also found that abductor hallucis, flexor digitorum brevis, quadratus plantae and flexor hallucis brevis cross-sectional areas were more strongly associated with sex and body mass index (BMI) than with CAI itself.⁶⁵ Clarifying whether reduced IFM sizes and foot strength are characteristics of CAI is crucial for rehabilitation, as these parameters are modifiable through targeted interventions.²⁵¹

The aim of this study was to address specific domains highlighted in the model of Hertel and Corbett, and particularly movement patterns and muscle sizes.³⁹ The primary objective of this study was to assess ankle and foot joint kinematics and kinetics during side-step cutting among individuals with CAI, copers and uninjured controls. We hypothesized that participants with CAI would exhibit lower ankle sagittal plane range of motion (ROM) in early stance compared to the two other groups. We also hypothesized lower ankle and Chopart joint power absorption and generation in CAI, and differences in joint moments. The secondary, exploratory objective was to compare abductor hallucis and flexor digitorum brevis muscle sizes across the three groups of participants, while controlling for BMI, and to examine their correlations with patient-reported outcomes and physical examination outcomes.

Materials and methods

1. Study design

This cross-sectional study followed STROBE guidelines.²⁹² The trial procedures were prospectively registered with ClinicalTrials.gov (NCT05815576).

2. Participants

Physically active males aged 18-44 were recruited between January 2023 and February 2025 from universities and high schools (students and staff members) in Belgium. Physically active was defined as ≥ 1 hour per week of self-reported physical exertion. The study followed the Declaration of Helsinki and was approved by the Ethics Committee of the University Hospitals Leuven. Written informed consent was obtained before the study start.

A priori power analysis (GPower 3.1)²⁹³ indicated 21 participants per group for the main hypothesis of this study, with power = 0.85 and alpha = 0.05. Therefore, we used previously published data comparing ankle joint ROM between males with CAI and controls captured while running in the same laboratory, resulting in Cohen's $f = 0.43$.⁸⁶

Inclusion criteria for CAI were (1) a first substantial LAS that occurred ≥ 1 year ago, (2) the last substantial LAS occurred ≥ 3 months ago, (3) two or more recurrent sprains, giving way episodes, or feelings of instability in the last six months, and (4) perceived ankle instability confirmed by a Cumberland Ankle Instability Tool (CAIT) score ≤ 24 .^{31, 39, 41} Copers experienced their most recent substantial LAS ≥ 1 year before enrolment. Copers and controls had CAIT ≥ 28 ,³⁹⁻⁴⁰ and Foot and Ankle Ability Measure (FAAM) subscales $\geq 95\%$.^{39, 294} A “substantial LAS” was defined as severe enough to interrupt weight-bearing physical activities for at least one day.³¹

Exclusion criteria: Any systemic, orthopaedic or neurological diseases, important lower limb deformities, a bilateral leg length difference > 3 cm, ankle surgery, and BMI $> 30\text{kg}\cdot\text{m}^{-2}$ (to ensure accurate marker tracking during side-step cutting).

3. Measurement set-up

All analyses occurred at the Clinical Motion Analysis Laboratory (Leuven, Belgium). Passive reflective markers (9.5/12 mm) were placed on anatomical landmarks according to the *IOR-4segment-model-1* (i.e. Rizzoli foot model and Plug-in Gait lower limb model; Figure 24).^{103, 109} Markers were tracked by 10 T-10 infrared cameras (100 Hz, Vicon, Oxford Metrics, UK). A pressure plate (2.8 sensors·cm⁻², 200 Hz, RsScan International, Belgium) was positioned on top of a force plate (1000 Hz, Advanced Mechanical Technology, USA), both embedded within the walkway. A 3D box (Materialise Motion, Paal, Belgium) ensured signal synchronization.



Figure 24. Marker placement according to Rizzoli multi-segment foot model. Lateral view (left image) and medial view (right image).

4. Measurement protocol and processing

Demographics, sprain history,³¹ sport habits, 6-items Foot Posture Index (FPI-6)^{242, 278} and sit-to-stand navicular drop were collected. Navicular drop was measured as the difference in navicular tuberosity height between sitting (leg verticalized) and standing (feet hip-width apart), using a calliper (resolution 0.02 mm). The navicular tuberosity was first marked with a pen to ensure consistent measurements.²⁹⁵ Dynamic balance was assessed with the modified Star Excursion Balance Test in the anterior, posterolateral and posteromedial directions, with three trials per direction. The maximal distance per direction was normalized to leg length, and computed into a composite score.²⁹⁶

Hallux and toe flexion strengths were measured in long sitting, the ankle maximally plantarflexed, with a hand-held dynamometer (microFET®2, Biometrics, France).¹³⁹ Participants pressed maximally for 5 seconds, repeating trials until there was less than 20% variability across the last three trials, and the mean of the two best values was analysed (Video file 4 and Video file 5).

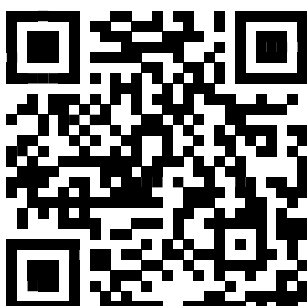
Video file 4: Assessment of hallux flexion strength in the long-sitting position.



OR

<https://www.dropbox.com/scl/fi/1wx8i8erno6uefhhwmdmqo/Hallux-flexion-strength-long-sitting-JL-Peters-Dickie.mp4?rlkey=5pkmmulhzcuwnxvkc6e40e5q7&st=haovrfx1&dl=0>

Video file 5: Assessment of toe flexion strength in the long-sitting position

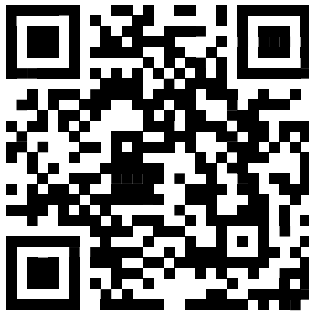


OR

<https://www.dropbox.com/scl/fi/va1j4jw7r0y3brm08e9kl/Toe-flexion-strength-JL-Peters-Dickie.mp4?rlkey=y7f24nq70coeo1loj903te5wd&st=s82h7k7l&dl=0>

Participants were instructed to perform barefoot 45° side-step cutting at $3.3 \text{ m}\cdot\text{s}^{-1}$ ($\pm 10\%$) (Video file 6). After two practice trials and static calibration, five valid trials were collected. Validity included speed and side-step cutting angle. Foot-strike and toe off events were manually identified on Nexus 2.12 software (Vicon, Oxford Metrics, UK) and trials were processed according to the *IOR-4segment-model-1* that distinguishes the shank, hindfoot, midfoot, metatarsus and hallux segments, separated by the ankle, Chopart, Lisfranc, and hallux metatarsophalangeal joints.^{103, 109-110} Data were then handled using an in-house MATLAB program (ACEP_Manager, Leuven, Belgium).

Video file 6: Demonstration of the side-step cutting task



OR

<https://www.dropbox.com/scl/fi/va9821qzhdhcmbhvu2hry/Side-step-cutting-45deg-JL-Peters-Dickie.mp4?rlkey=fagjsucmkvww658rzikn6cear&st=6zzxqoe1&dl=0>

Joint angular position and velocities were determined in the sagittal, frontal and transverse planes. Subsequently, internal joint moments were calculated by an inverse dynamic approach (Newton-Euler equations), with vertical ground reaction force (GRF) distributed according to foot pressure distribution beneath foot segments.¹⁰⁹ Joint power was computed as *moment* $\times$ *angular velocity*. Data were normalized to 100% of the stance phase and the mean curve of each participant was further used in the analyses. Foot strike index,²⁹⁷ spatiotemporal parameters, joint ROM, positive and negative work, and the peaks for joint velocity, moment, power and GRF were computed. ROM were more precisely calculated within the peak impact (0-6% of stance), active peak (7-45%) and propulsive (46-100%) subphases of stance.⁸⁶ Kinetic outcomes were normalized to body weight.

Ultrasound imaging (Telemed Echoblaster, 10 MHz linear transducer) was used to obtain images for abductor hallucis and flexor digitorum brevis (Figure 25 and Figure 26).²⁹¹ Muscle cross-sectional area and thickness were measured with ImageJ software.



Figure 25. Ultrasound imaging of abductor hallucis, transversal view (left) and longitudinal view (right).



Figure 26. Ultrasound imaging of flexor digitorum brevis, transversal view (left) and longitudinal view (right).

5. Statistics

Alpha was set to 0.05. Zero-dimensional data were analysed using SPSS v27 (IBM, USA). Categorical data were compared with Chi-squared tests. Quantitative data normality was verified using Shapiro-Wilk test, and homogeneity of variances was assessed through Levene's test. Normally-distributed data were analysed using One-Way ANOVA (with or without Welch correction), followed by post-hoc t-tests. Non-normally distributed data were compared using Kruskal-Wallis tests, followed by Mann-Whitney post-hoc tests. Post-hoc p-values and analyses related to ROM were corrected using Bonferroni corrections. Mean differences and median differences were used as effect size estimates. Cross-correlation joint coupling coefficients were interpreted as weak (< 0.3), moderate (0.3 to 0.69), and strong (≥ 0.7).²⁹⁸

Kinematic and kinetic curves were analysed using Statistical Parametric or non-Parametric Mapping (SPM or SnPM),¹²² through an open source code for MATLAB.¹²³ Steps included a test for normality followed by SPM- or SnPM-ANOVA and its corresponding post-hoc tests with Bonferroni correction.

IFM sizes were analysed exploratively using general linear models (BMI in covariate), followed by correlation analyses between IFM sizes, questionnaire scores, balance, foot posture, foot mobility, and strength, reporting Pearson (r) or Spearman (ρ) coefficients depending on normality. Coefficients were interpreted as low (0.3 to 0.49), moderate (< 0.7), high (< 0.9), and very high (≥ 0.9).²⁹⁹

Results

1. Participants

A total of 22 participants with CAI, 24 copers and 23 controls took part in the study (Figure 27). Demographics are displayed in Table 11, while details related to ankle sprain history and rehabilitation are available in Appendix 4.A. Participants with CAI reported lower ankle function compared to the two other groups, with median differences of 5% for Activities of Daily Living subscale and 12.5% for Sports subscale ($p < 0.001$). Post-hoc tests are detailed in Appendix 4.B.

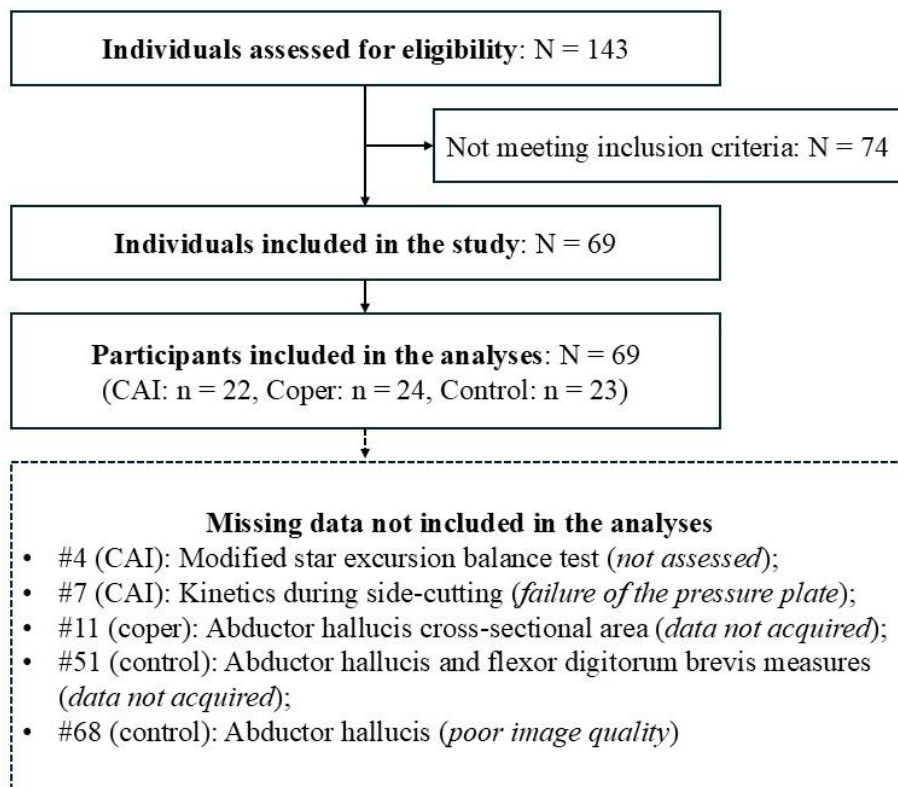


Figure 27. Flow chart of the participants included in the study. CAI (participants with chronic ankle instability), N (number of participants), n (number of participants in a subgroup), # (number of the participant). Missing data are detailed in the lower part of the figure, including the reason for missing data.

Table 11: Demographics and self-reported outcome measures

Outcome	CAI (n = 22) ^a	Coper (n = 24) ^b	Control (n = 23) ^c	p-value
Age (years) †	25.8 (5.4)	25.9 (3)	23.9 (4)	0.091
Height (cm) ‡	180.4 (6.9)	180.7 (6)	179.3 (6.9)	0.747
Weight (kg) †	77.8 (9.7)	77.5 (8.7)	72 (6.7)	0.039*
BMI (kg·m ⁻²) ‡	23.9 (2.5)	23.8 (2.9)	22.5 (2.3)	0.095
Shoe size (European) ‡	43.3 (1.6)	43.5 (1.4)	43 (1.6)	0.516
Dominant side (n)	3 left / 19 right	4 left / 20 right	4 left / 19 right	0.936
Affected side (n)	13 dominant / 9 non-dominant	17 dominant / 7 non-dominant	/	0.30
Weekly physical activity (hours·week ⁻¹) ‡	5.5 (4.4)	5.5 (3.4)	5.1 (2.6)	0.936
CAIT (/30) ‡	17.5 (4.6) ^{b,c}	29.2 (1.2) ^a	29.6 (0.9) ^a	< 0.001*
FAAM ADL subscale (%) ‡	90.3 (9.8) ^{b,c}	99.7 (0.7) ^a	99.9 (0.3) ^a	< 0.001*
FAAM Sports subscale (%) ‡	81.3 (16.5) ^{b,c}	99.3 (2.1) ^a	99.3 (1.9) ^a	< 0.001*
FABQ physical activity (/24) ‡	7.4 (5.6)	5.4 (5.5)	4.3 (4.4)	0.127
FABQ work (/42) ‡	3.3 (5.1)	1.1 (2.7)	1.4 (3)	0.102
FABQ total (/66) ‡	10.7 (9.5)	6.5 (6.9)	5.7 (5.7)	0.089

Legend: Data are presented as mean (standard deviation). ADL (activities of daily living), CAI (chronic ankle instability), CAIT (Cumberland ankle instability tool), FAAM (foot and ankle ability measure), FABQ (fear avoidance belief questionnaire), n (number of participants), FPI-6 (6-items Foot Posture Index), † (normality not rejected), ‡ (normality rejected), * (statistically significant); superscript letters indicate significant group differences during post-hoc tests ($p < 0.001$).

2. Physical examination

Individuals with CAI and copers had a lower normalized hallux flexion strength compared to the control group, with mean differences of $0.25 \text{ N}\cdot\text{kg}^{-1}$ ($p = 0.050$) and $0.28 \text{ N}\cdot\text{kg}^{-1}$ ($p = 0.02$), respectively (Table 12).

Table 12: Outcomes from the physical examination

Outcomes	CAI ^a	Coper ^b	Control ^c	p-value
mSEBT anterior (% leg length) ‡	0.9 (0.06)	0.91 (0.08)	0.91 (0.07)	0.589
mSEBT posteromedial (% leg length) †	0.91 (0.1)	0.96 (0.09)	0.96 (0.08)	0.135
mSEBT posterolateral (% leg length) †	0.87 (0.11)	0.93 (0.08)	0.93 (0.08)	0.073
mSEBT composite (% leg length) †	0.9 (0.08)	0.93 (0.06)	0.93 (0.06)	0.105
Hallux flexion strength ($\text{N}\cdot\text{kg}^{-1}$) †	1.09 (0.32) ^c	1.07 (0.34) ^c	1.35 (0.37) ^{a,b}	0.014*
Other toes flexion strength ($\text{N}\cdot\text{kg}^{-1}$) †	1.04 (0.31)	1.13 (0.29)	1.22 (0.33)	0.178
FPI-6 (score -12 to +12) ‡	4.5 (2.8)	3.2 (3.5)	3.5 (4.3)	0.455
Navicular drop (cm) ‡	0.63 (0.29)	0.67 (0.27)	0.64 (0.27)	0.623

Legend: Data are presented as mean (standard deviation). CAI (chronic ankle instability), n (number of participants), mSEBT (modified star excursion balance test), † (normality not rejected), ‡ (normality rejected), * (statistically significant); superscript letters indicate the groups that differed significantly during post-hoc tests.

3. Side-step cutting task

Participants performed side-step cutting with a mean approach speed of $3.45 \text{ m}\cdot\text{s}^{-1}$, close to the target speed of $3.3 \text{ m}\cdot\text{s}^{-1}$ ($\pm 10\%$). Ankle and foot ROM in the sagittal plane, peak moments in the sagittal plane, peak power and work are displayed in Table 13. There was no between-group difference in these outcomes, and our hypotheses related to ankle ROM were thus not verified.

Participants with CAI had lower posterior peak GRFs compared to controls, with a mean difference of 6% of body weight ($p = 0.03$; Table 14). SPM revealed that participants with CAI had lower posterior GRFs from 22 to 46% of stance compared to controls ($p < 0.05$, Figure 28).

Spatiotemporal parameters, joint ROM and moments in the transverse and frontal planes, and cross-correlations are displayed in Appendices (Appendix 4.C, 4.D, 4.E). There was no significant between-group difference in these secondary outcome measures, and cross-correlation coefficients were weak ($r < 0.3$).

Table 13: Ankle and foot joint kinematics and kinetics during side-step cutting

Outcomes	CAI	Coper	Control	p-value
ANKLE				
PF/DF ROM: P 1 (°) ‡	4.7 (1.8)	5 (2.1)	4.3 (1.8)	1
PF/DF ROM: P 2 (°) †	20.4 (3.1)	21.3 (3.4)	20.6 (2.9)	1
PF/DF ROM: P 3 (°) †	38.8 (7.5)	41 (4.1)	40.9 (4.7)	1
Peak PF moment (Nm·kg ⁻¹) †	2.48 (0.34)	2.57 (0.42)	2.65 (0.4)	0.33
Peak gen. power (W·kg ⁻¹) †	12.3 (3.25)	13.66 (2.35)	13.93 (2.5)	0.117
Peak abs. power (W·kg ⁻¹) ‡	4.71 (2.15)	5.53 (1.89)	5.88 (3.15)	0.245
Pos. work (J·kg ⁻¹) ‡	1.04 (0.25)	1.11 (0.19)	1.17 (0.18)	0.45
Neg. work (J·kg ⁻¹) †	0.42 (0.17)	0.51 (0.19)	0.49 (0.2)	0.228
CHOPART				
PF/DF ROM: P 1 (°) †	2.9 (1.8)	3.2 (2)	2.7 (2)	1
PF/DF ROM: P 2 (°) ‡	6.2 (1.3)	6.4 (2.4)	6.9 (2.5)	1
PF/DF ROM: P 3 (°) †	17 (3.8)	17.5 (5.4)	18.3 (4.2)	1
Peak PF moment (Nm·kg ⁻¹) †	1.83 (0.29)	1.91 (0.31)	1.95 (0.3)	0.45
Peak gen. power (W·kg ⁻¹) †	4.24 (1.22)	4.44 (1.97)	4.74 (1.09)	0.538
Peak abs. power (W·kg ⁻¹) ‡	2.14 (1)	2.23 (1.03)	2.49 (1.16)	0.527
Pos. work (J·kg ⁻¹) †	0.23 (0.05)	0.23 (0.09)	0.26 (0.07)	0.354
Neg. work (J·kg ⁻¹) ‡	0.17 (0.05)	0.18 (0.06)	0.19 (0.05)	0.623
LISFRANC				
PF/DF ROM: P 1 (°) ‡	1.3 (1)	1.4 (1.1)	1.9 (1.1)	0.46
PF/DF ROM: P 2 (°) ‡	2.7 (1.8)	2.2 (1.4)	3.1 (1.5)	0.19
PF/DF ROM: P 3 (°) ‡	8 (3.1)	7.8 (2.6)	7.5 (2.6)	1
Peak PF moment (Nm·kg ⁻¹) ‡	0.78 (0.22)	0.8 (0.22)	0.83 (0.18)	0.71
Peak gen. power (W·kg ⁻¹) ‡	2.4 (1.16)	2.31 (0.68)	1.99 (0.64)	0.395
Peak abs. power (W·kg ⁻¹) ‡	0.4 (0.29)	0.31 (0.16)	0.46 (0.31)	0.189
Pos. work (J·kg ⁻¹) †	0.15 (0.06)	0.12 (0.04)	0.12 (0.03)	0.137
Neg. work (J·kg ⁻¹) ‡	0.025 (0.015)	0.021 (0.011)	0.027 (0.014)	0.318
HALLUX				
PF/DF ROM: P 1 (°) ‡	6.9 (5.3)	5.9 (3.6)	5.4 (2.7)	1
PF/DF ROM: P 2 (°) ‡	7.1 (3.7)	6.4 (2.4)	7.5 (4.2)	1
PF/DF ROM: P 3 (°) †	39.1 (7.6)	42.2 (5.6)	41.3 (6.5)	0.83
Peak PF moment (Nm·kg ⁻¹) ‡	0.26 (0.08)	0.24 (0.08)	0.28 (0.09)	0.36
Peak gen. power (W·kg ⁻¹) ‡	0.41 (0.22)	0.41 (0.32)	0.45 (0.21)	0.579
Peak abs. power (W·kg ⁻¹) ‡	2.56 (1)	2.58 (0.89)	2.69 (0.84)	0.609
Pos. work (J·kg ⁻¹) ‡	0.014 (0.007)	0.013 (0.011)	0.016 (0.007)	0.114
Neg. work (J·kg ⁻¹) †	0.15 (0.05)	0.14 (0.06)	0.16 (0.05)	0.527

Legend: Data are presented mean and standard deviation. Abs. (absorption), DF (dorsiflexion), gen. (generation), neg. (negative), P 1 (impact peak phase), P 2 (active peak phase), P 3 (propulsive phase), PF (Plantarflexion), pos. (positive), ROM (range of motion), † (normality not rejected), ‡ (normality rejected). P-values were corrected to account for the number of subphases for ROM, by multiplying the value by 3.

Table 14: Peaks in ground reaction forces during side-step cutting

Outcomes	CAI ¹	Coper ²	Control ³	p-value
Vertical (% body weight)	216.4 (23.5)	222.5 (29.2)	230.9 (26.1)	0.20
Medial (% body weight)	49 (9.1)	51.4 (5.7)	51.2 (8)	0.53
Posterior (% body weight)	32 (7.9) ³	33 (6.9)	38.2 (8.4) ¹	0.021*
Anterior (% body weight)	16.8 (4.1)	18.8 (3.6)	16.7 (4.8)	0.18

Legend: Data are presented as mean and standard deviation. CAI (chronic ankle instability), † (normality not rejected), ‡ (normality rejected), * (statistically significant); superscript numbers indicate the groups that differed significantly during post-hoc tests.

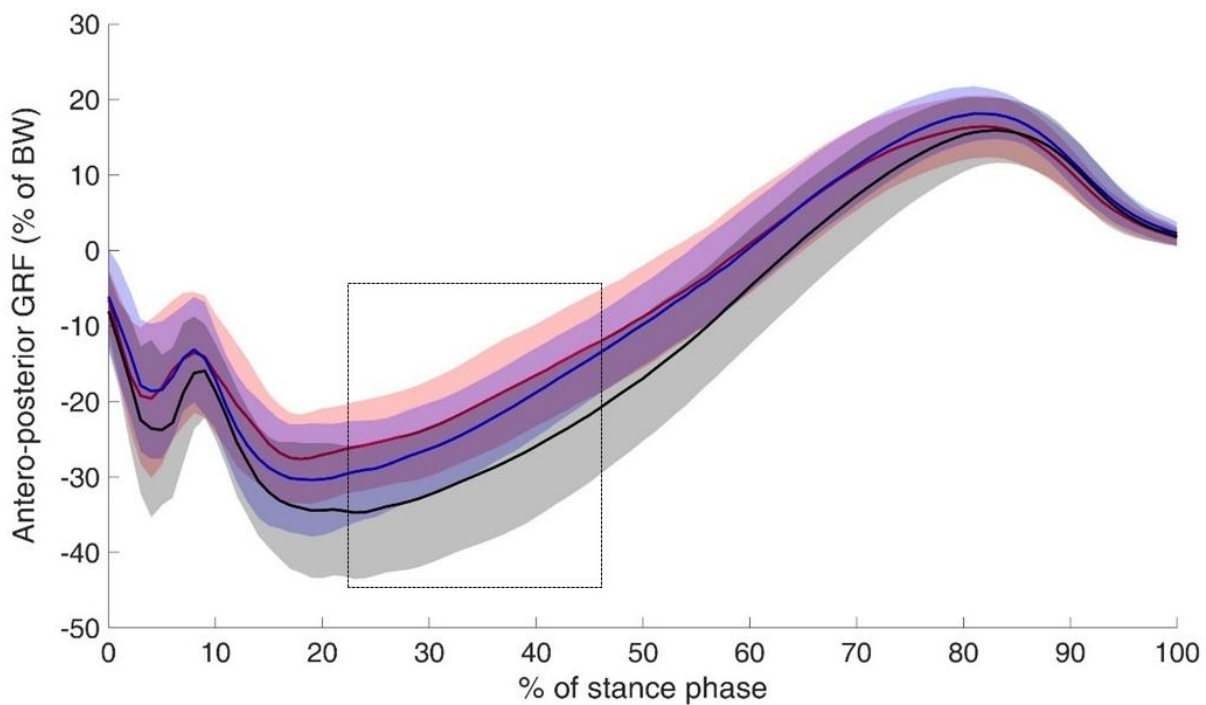


Figure 28. Mean and standard deviation (cloud) of the antero-posterior ground reaction forces during the stance phase among individuals with chronic ankle instability (red), copers (blue) and controls (black). The critical F^* of the SPM ANOVA was exceeded from 22 to 56% of stance ($F^* = 6.55$, $p < 0.001$). During post-hoc tests, individuals with CAI and controls differed significantly from 22 to 46% of stance (indicated by the dotted rectangle), with critical $t^* = \pm 3.61$ ($p < 0.05$). BW (body weight), GRF (ground reaction force).

4. IFM sizes

IFM sizes were explored through general linear models that accounted for BMI and groups (Table 15). Detailed IFM measures are displayed in Appendix 4.F. The variance in flexor digitorum brevis cross-sectional area could be explained based on BMI (p model = 0.03, adjusted $R^2 = 0.094$, p BMI = 0.02). The model for abductor hallucis thickness seemed to be valid (p model = 0.018, adjusted $R^2 = 0.104$),

although neither BMI or group significantly predicted the variance in thickness when considered alone. Scatter plots related to these models are available in Appendix 4.G.

Table 15: General linear models related to foot muscle sizes

Model features	Model parameters	B	Partial η^2	p
Abductor hallucis CSA $R^2 = 0.07$ Adjusted $R^2 = 0.03$ $p = 0.19$	Intercept	1.6	/	/
	BMI	0.03	0.02	0.23
	Group		0.04	0.26
	- CAI	0.24		
	- Coper	0.02		
Abductor hallucis thickness $R^2 = 0.15$ Adjusted $R^2 = 0.10$ $p = 0.018^*$	Intercept	0.75	/	/
	BMI	0.02	0.05	0.08
	Group		0.08	0.08
	- CAI	0.15		
	- Coper	0.07	/	/
Flexor digitorum brevis CSA $R^2 = 0.14$ Adjusted $R^2 = 0.09$ $p = 0.025^*$	Intercept	1.23		
	BMI	0.05	0.08	0.020*
	Group		0.04	0.3
	- CAI	0.18		
	- Coper	0.04		
Flexor digitorum brevis thickness $R^2 = 0.07$ Adjusted $R^2 = 0.03$ $p = 0.18$	Intercept	0.74	/	/
	BMI	0.02	0.06	0.049*
	Group		0.04	0.88
	- CAI	0.02		
	- Coper	0.01		

Legend: B (coefficient of regression), BMI (body mass index), CAI (chronic ankle instability), CSA (cross-sectional area), p (p-value), R^2 (goodness of fit of the model), η^2 (eta-squared), * (statistically significant). A model was interpreted if the p-value of the model was statistically significant. Adjusted R^2 was interpreted as the percentage of variance in muscle size explained by the model.

Abductor hallucis cross-sectional area and thickness were highly positively correlated ($r = 0.72$). This correlation was moderate for flexor digitorum brevis ($r = 0.51$). Multiple low yet significant correlations were found between abductor hallucis measures and the CAIT, the FAAM, and foot posture, as well as between flexor digitorum brevis cross-sectional area and the CAIT (Appendix 4.H, Appendix 4.I). The aforementioned correlations with questionnaires were negative, with bigger IFM associated with lower CAIT (more perceived instability) and FAAM (lower self-reported function).

Discussion

This was the first study to compare multi-segment foot joint kinematics and kinetics during side-step cutting among individuals with CAI, copers, and uninjured controls. Contrary to our hypotheses, no significant sagittal plane kinematic differences and only limited kinetic discrepancies were found, suggesting that foot joint biomechanics during this task may not clearly differentiate the groups.

CAI is a multifactorial condition resulting from the interaction of various impairments.³⁹ In our study, individuals with CAI reported higher levels of perceived instability and self-reported disability compared to the two other groups. In their systematic review, Houston et al. reported Hedges' *g* effect sizes ranging from 1.8 to 3.3 when comparing CAIT scores between individuals with CAI and controls, and from 1 to 3.3 for the FAAM subscales⁷¹. Although we used median differences due to skewed data distributions, our results would correspond to Hedges' *g* values of 3.6 for CAIT and 1.4 to 1.6 for FAAM subscales, when comparing CAI to controls. Our data would turn into comparable effect sizes when comparing CAI to copers. Despite these sensory-perceptual impairments, individuals with CAI showed few differences in motor-behavioural factors (cutting biomechanics, balance, strength) or structural measures (IFM sizes). This highlights the need for clinicians to assess and target multiple CAI domains in rehabilitation.^{47, 300}

A qualitative comparison of kinematic curves confirmed that the three groups exhibited similar movement patterns. To our knowledge, only one previous study has compared cutting kinematics between individuals with CAI and controls – specifically in female football players – and reported no between-group differences.⁸³ During running, although some evidence suggests altered foot joint movement patterns in individuals with CAI,⁸⁴⁻⁸⁵ others have reported no significant differences compared to uninjured controls.³⁰¹⁻³⁰² Qualitatively, we observed non-significant trends towards kinetic differences across groups. Specifically, individuals with CAI tended to exhibit less ankle power generation in late stance and lower Chopart plantarflexion moment compared to the two other groups. Additionally, both CAI and coper participants appeared to show less Chopart power generation in late stance, as well as lower hallux power absorption and hallux dorsiflexion moment compared to controls. As this was the first study to

investigate foot joint kinetics across these groups, direct comparison with existing literature is limited.

SPM analysis identified lower posterior GRFs during the first half of stance in CAI compared to controls, along with lower peak posterior GRFs in CAI. To our knowledge, these results of individuals with CAI exhibiting less deceleration during change of direction are unique. Previous studies either did not report antero-posterior

GRFs,^{125, 127, 130} or found no differences between CAI and controls.⁸³ Studies should thus further investigate deceleration capacities in individuals with CAI, for example by including deceleration and acceleration as covariates when comparing foot joint biomechanics between groups.

These differences in posterior GRFs may reflect compensatory strategies, redistributing energy normally absorbed at the lower leg and foot to more proximal segments. Although knee and hip biomechanics were not assessed in this study, a systematic review reported frequent proximal alterations in CAI during cutting tasks.³⁰³ These adaptations include higher hip flexion during the active peak phase,¹³² and lower knee abduction moment in the second half of stance.²⁸⁹ Lower posterior GRFs also minimize shear forces in the antero-posterior direction, potentially increasing stability in the transverse and frontal planes. Clinically, such adaptations may result in abnormal loading patterns and increased tissular demand that could produce long-term joint degeneration if tissue tolerance is chronically exceeded. From a performance perspective, such adaptations and reduced braking forces may impair directional change efficiency. Given the importance of deceleration as a component of agility in many sports, clinicians should be aware that individuals with CAI may experience issues with deceleration when performing changes of direction.

We measured IFM sizes using ultrasound imaging, a method shown to be valid compared to magnetic resonance imaging and to have good reproducibility.¹⁴⁹ We explained approximately 10% of the variances in abductor hallucis thickness and flexor digitorum brevis cross-sectional area with models that involved BMI and groups, and the effect of groups was not significant. These findings are consistent with recent literature, where the effect of participants being CAI, copers or controls had only little influence on IFM sizes.^{65, 151} Fraser et al. explained 13 to 24% of IFM sizes variance using models with sex, BMI, group (CAI, coper, recent LAS and

control), foot posture, and foot mobility, yet a substantial proportion remained unexplained – as in our study.⁶⁵ Interestingly, we observed low negative correlations ($p = -0.27$ to -0.36) between IFM sizes and CAIT/FAAM scores, suggesting a possible link between muscle morphology and perceived instability or self-reported function that warrants further investigation.

We also investigated foot strength using dynamometry and found lower normalized hallux flexion strength in CAI and copers versus controls, and relatively similar toe flexion strength across groups. These results are partially in agreement with those of Fraser et al., who reported lower normalized hallux and toe flexion strengths in CAI compared to copers and controls.⁶⁵ Hallux flexion and toe flexion strengths seem to influence functional abilities, such as static and dynamic balance performance.³⁰⁴⁻³⁰⁵ Only a couple of studies investigated correlations between foot strength and change of direction performance, resulting in opposite conclusions.^{232, 306} The concomitant existence of lower foot strength despite relatively similar IFM sizes across groups may be explained by altered neuromuscular control in CAI, driven by neuromuscular inhibition.³⁹ IFM function should thus further be explored in CAI, providing a key rationale for implementing interventions that target IFM strengthening to address sensory-perceptual and motor-behavioural impairments associated with CAI.²⁵¹

Personal and task-related factors influence kinematics, kinetics, and GRFs during a change of direction. First, individuals with CAI may present with heterogeneous alterations.³⁹ In our study, perceived instability served as criterion to distinguish between CAI and the two other groups. Mechanical ankle joint laxity is an important confounding factor that influences side-step cutting biomechanics, as consistently reported in two previous studies.^{125, 127} Besides, our participants with CAI reported few alterations in self-reported function, many practiced sports involving changes of direction, and the majority had pursued rehabilitation. These factors could explain the absence of biomechanical differences between groups. Second, task-related factors such as approach speed, cutting angle, being barefoot and anticipation may further have influenced the outcomes. For example, Takeuchi & Anan found a longer stance time and differences in horizontal GRFs in CAI compared to controls during side-step cutting at 90° and at 135°, but no differences during a 45° cutting task. Importantly, these tasks were performed at a self-selected comfortable speed, and they did not observe any foot joint kinematic

difference between groups.⁸³ Lastly, real-world settings also involve chaos in the environment and fatigue, which could further impact motor behaviour.³⁰⁷

The limitations of this study should be considered when interpreting our findings. First, the external validity of our results is restricted to physically active young male adults performing anticipated side-step cutting at moderate speeds. Second, successful return to sport could have been part of the inclusion criteria for copers.⁴⁰ It is possible that some participants included in the coper group adjusted their sport type or sport level as a consequence of LAS. Besides, since FAAM did not serve as an inclusion criterion for CAI, self-reported function was only moderately affected in the CAI group. In addition, most participants with CAI, and half of copers pursued rehabilitation, which could further have decreased the potential differences between the three groups. Third, part of the inter-individual variability in kinematics and kinetics may be explained by small differences in marker placement.³⁰⁸ Lastly, we used a low-end ultrasound device to measure IFM sizes, typically resulting in a lower image resolution and that could affect measure reliability.²⁹¹

Conclusion

In conclusion, young adults with perceived ankle instability may exhibit no or minimal differences in foot joint kinematics and kinetics during a submaximal, anticipated side-step cutting task compared to copers and uninjured controls. Clinicians should be aware that people with ankle instability may encounter altered horizontal deceleration when changing direction. Future research should further explore these braking abilities, as well as the roles of the intrinsic and extrinsic foot muscles.

Acknowledgements

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Summary

Context: Movement patterns are part of the motor-behavioural impairments suspected to contribute to CAI, but they are understudied at the foot level. This was the first study to explore foot joint kinetics among individuals with CAI, and the first to compare side-cutting foot joint kinematics between CAI and copers.

Besides, intrinsic foot muscle sizes and foot strength could also be part of deficits associated to CAI. It is unclear if individuals with CAI exhibit foot muscle atrophy and weakness, while impairments could be targeted in rehabilitation.

Findings: Twenty-two individuals with CAI, 24 copers and 23 controls performed anticipated side-step cutting at $3.5 \text{ m}\cdot\text{s}^{-1}$. Foot joint ranges of motion and kinematic patterns were similar across groups. Participants with CAI exhibited trends for lower Chopart plantarflexion moment and lower ankle power generation compared to the two other groups. Chopart power generation and hallux power absorption appeared to be lower in CAI and copers compared to healthy controls.

Posterior ground reaction forces (i.e. braking forces) were lower in CAI compared to controls during the 45° side-step cutting task. This may be the consequence of energy absorption redistribution from the ankle and foot complex to more proximal joints.

Intrinsic foot muscle sizes did not seem to be altered among individuals with CAI, although hallux flexion strength (normalized to body mass) was lower in individuals with CAI and copers compared to healthy controls. These results suggest that some individuals with CAI could exhibit altered foot strength.

Implications for clinical practice and future research: Clinicians should assess athletes with CAI during sport-specific actions, and be aware that braking abilities may be altered during changes of direction. Future studies should explore lower-limb and foot joint dynamics while performing sporting tasks in more ecological settings, by varying speed, cutting angle, environmental conditions and cognitive demands.

Clinicians could include foot strength-related outcomes in the comprehensive assessment of individuals with CAI, to target specific impairments that may contribute to their clinical presentation.

Chapter 5. Effects of repeated foot muscle contractions on neuromuscular fatigue and side-step cutting foot joint biomechanics in people with chronic ankle instability

Jean-Louis Peters-Dickie, Virginie Otlet, Christine Detrembleur, Sébastien Lobet, Kevin Deschamps

Abstract

Movement patterns are frequently impaired in people with chronic ankle instability (CAI). The intrinsic foot muscles (IFMs) contribute to foot stability and propulsion during sporting tasks. Their sizes and strength could be altered in CAI, potentially modulating movement patterns. This study examined the immediate effects of repeated isometric toe flexor contractions on neuromuscular fatigue and cutting biomechanics in CAI. Twenty-one recreationally active males with CAI performed anticipated 45° side-step cutting ($\sim 3.3 \text{ m}\cdot\text{s}^{-1}$) before and after 16 minutes of repeated isometric toe flexor contractions with electrical stimulation of the medial longitudinal arch. Three-dimensional foot joint kinematics and kinetics were analysed using the IOR-4Segment model. Contrary to hypotheses, frontal-plane ankle range of motion and peak moments remained unchanged. However, significant alterations emerged in transverse-plane ankle moments, with reduced adduction and increased abduction moments in late stance, accompanied by increased posterior and medial ground reaction forces. Chopart joint was more dorsiflexed and Lisfranc mobility decreased after the intervention. Hallux ranges of motion increased during propulsion, suggesting altered stiffness. Strength declined moderately, reflecting inconsistent neuromuscular fatigue, that may be explained by the low current intensity sustained by participants. These findings imply that short-duration IFM activation with electrical stimulation acutely modifies intersegmental foot mechanics without compromising frontal-plane ankle control, supporting the safety of incorporating such interventions into CAI rehabilitation, and encouraging further studies using longer protocols and more demanding, sport-specific tasks.

Introduction

Chronic ankle instability (CAI) is characterized by recurrent sprains, episodes of the ankle giving way, pain and perceived instability.³³ It is highly prevalent among team sport athletes who have already sustained an ankle sprain.³⁰⁹ Changes of direction are key components of performance in many sports and place the ankle in a vulnerable supinated position.^{32, 125, 127}

Altered movement patterns contribute to the clinical presentation of individuals with CAI.³⁹ During side-cutting tasks, individuals with CAI had a more inverted position of the ankle,¹³² and lower net internal ankle eversion moments compared with healthy controls.²⁸⁹ These studies modelled the foot as one rigid segment, which neglects more distal foot joints and overestimates ankle dynamics.^{104, 109}

Using multi-segment kinematic foot models during running, researchers found altered sagittal plane ankle range of motion (ROM) in individuals with CAI. More precisely, they had less ROM in early stance and more ROM until mid-stance than healthy controls.⁸⁶ Curve analyses with statistical parametric mapping (SPM) revealed more ankle eversion and more medial forefoot inversion during the second half of running stance in individuals with CAI compared with controls.⁸⁴ Foot joint kinetics – moment and power – and foot joint work have not yet been investigated in CAI, but could provide valuable insights into altered joint loading patterns resulting from articular remodelling, pathologic laxity and kinematic restrictions associated with CAI.^{39, 109}

The neuromuscular control of the ankle and foot complex is ensured by the extrinsic and intrinsic foot muscles (IFMs), which are differentiated by their anatomical origin in the lower leg or within the foot.^{10, 12} The IFMs support the medial longitudinal arch, attenuate foot impact and support metatarsophalangeal joint stiffness during locomotion.^{22, 153} They are also rich in sensory receptors,²⁷ and could thereby contribute to postural control by transmitting proprioceptive information to the central nervous system.¹⁰ Individuals with CAI have been shown to exhibit smaller abductor hallucis and flexor digitorum brevis muscles compared with their healthy counterparts,⁶⁶ suggesting altered IFM strength.¹⁴⁹ Impaired IFM function is further supported by lower hallux and toe flexor strength measured in the long-sitting position in CAI than in controls.⁶⁵

Meta-analyses showed that interventions targeting IFM activation improved foot strength, may increase IFM sizes, and seem to modify lower-limb kinematics and kinetics during locomotion among healthy and symptomatic subjects.²⁵¹ Emerging evidence supports the use of IFM exercises among individuals with CAI.³¹⁰ Only one study investigated the acute effects of 6-10 minutes of repeated toe flexions on foot joint kinematics during walking among healthy participants, and found increased hindfoot eversion after the intervention.¹⁶² Repeated muscular contractions are also likely to result in decreased foot strength, a marker of neuromuscular fatigue,¹⁵⁷ offering the unique opportunity to simulate impairment in IFM function. Neuromuscular electrical stimulation is sometimes used to promote IFM activation during exercises,¹⁶⁶ and could acutely result in muscle fatigue.^{158, 168} However, the acute effects of IFM exercises combined with electrical stimulation on foot joint biomechanics of individuals with CAI remain to be explored.

Therefore, this study aimed to evaluate the immediate biomechanical effects of a single session of isometric toe flexion exercises combined with medial longitudinal arch neuromuscular electrical stimulation in individuals with CAI. We primarily hypothesized that the intervention would increase frontal plane ankle ROM. Additionally, we expected an increase in frontal plane ankle peak moments and ankle peak power. Finally, we anticipated a reduction in hallux flexion strength following the intervention, indicative of neuromuscular fatigue.

Methods

1. Study design

This was a single-group, within-subject pre–post intervention study. The study was conducted in accordance with the Declaration of Helsinki, and all participants provided written informed consent prior to participation. Ethical approval was obtained from the University Hospitals Leuven Ethics Committee and the study was prospectively registered on ClinicalTrials.gov (NCT05815576).

A sample size of 21 participants was calculated as part of a broader investigation, using existing foot joint kinematic data captured during running in males with CAI and healthy controls.⁸⁶ G*Power (version 3.1)²⁹³ was used to compute the required sample size given the expected Cohen's f effect size = 0.6, α = 0.05 and power = 0.85.

2. Participants

Twenty-one males with CAI were recruited between January 2023 and February 2025 from university and high school campuses across Belgium (Table 16). Inclusion criteria were to be aged 18-44 years and to engage in at least 60 minutes of moderate-to-vigorous physical activity per week. CAI was diagnosed according to the International Ankle Consortium guidelines:³¹ (1) a first substantial lateral ankle sprain more than 1 year prior to enrolment, (2) last substantial lateral ankle sprain more than 3 months ago, (3) at least 2 episodes of ankle giving way, recurrent sprains, or feelings of instability in the past 6 months, and (4) a Cumberland Ankle Instability Tool score ≤ 24 . A substantial ankle sprain was defined as one that interrupted weight-bearing activities for at least 1 day.^{31, 39, 41} Exclusion criteria included systemic, orthopaedic, or neurological conditions; significant lower limb deformities; bilateral leg length discrepancy of more than 3 cm; prior ankle surgery; or BMI higher than $30 \text{ kg}\cdot\text{m}^{-2}$ to ensure accurate motion capture.

Participants remained barefoot during testing. Clinical assessments included the Cumberland Ankle Instability Tool, Foot and Ankle Ability Measure, 6-item Foot Posture Index,²⁴² and sit-to-stand navicular drop test using a calliper (resolution 0.02 mm).²⁹⁵

Table 16: Demographic characteristics of the participants included in the study

Characteristics	Participants (n = 21)
Age (years) ‡	23 [22 – 29]
Height (cm) †	180.5 (7.0)
Weight (kg) †	80.0 (10.0)
BMI (kg/m ²) ‡	23.4 [22.4 – 27.5]
Shoe size (European) †	43.6 (1.4)
Dominant side (n)	3 Left / 18 Right
Affected side (n)	13 Dominant / 8 Non-dominant
Weekly physical activity (hours) ‡	5 [2.5 – 7]
CAIT (/30) †	16.8 (4.7)
FAAM ADL subscale (%) ‡	94.0 [82.1 – 97.6]
FAAM Sports subscale (%) ‡	87.5 [68.8 – 93.8]
FABQ total (/66) ‡	7 [5 – 12.3]
Toe strength (N) †	81.4 (29.6)
6-items Foot posture index (-12 to +12) ‡	5 [3 – 7]

Legend: Data are presented as mean (standard deviation) when normality was not rejected (†), and as median [quartile 1 – quartile 3] when normality was rejected (‡). ADL (activities of daily living), CAIT (Cumberland ankle instability tool), FAAM (foot and ankle ability measure), FABQ (fear avoidance belief questionnaire), n (number of participants).

3. Instrumentation

For the side-step cutting task, ten T-10 infrared cameras (VICON, Oxford Metrics, Oxford, UK) tracked the position of reflective markers at a sampling rate of 100 Hz. Markers were placed on anatomical landmarks following the IOR-4Segment model, which divides the lower limb into the shank, hindfoot, midfoot, metatarsus, and hallux, separated by the ankle, Chopart, Lisfranc, and metatarsophalangeal joints. Segment definitions and joint centres were described in Davis et al. (1991) and Leardini et al. (2007).¹¹⁰ Ground reaction forces were recorded via a custom-built force plate (Advanced Mechanical Technology, Watertown, USA), overlaid with a pressure plate (Footscan, RSscan International, Olen, Belgium; spatial resolution: 2.8 sensors·cm⁻²). Force data were sampled at 1000 Hz and pressure data at 200 Hz. A 3D synchronisation unit (Materialise Motion, Paal, Belgium)

ensured temporal alignment across all systems. This setup is consistent with previously published protocols.^{86, 112}

A hand-held dynamometer (microFET®2, Biometrics, Gometz-le-Chatel, France) was used for strength assessments.

4. Procedures

Strength and side-step cutting biomechanics were assessed before and immediately after the neuromuscular electrical stimulation intervention (Figure 29).

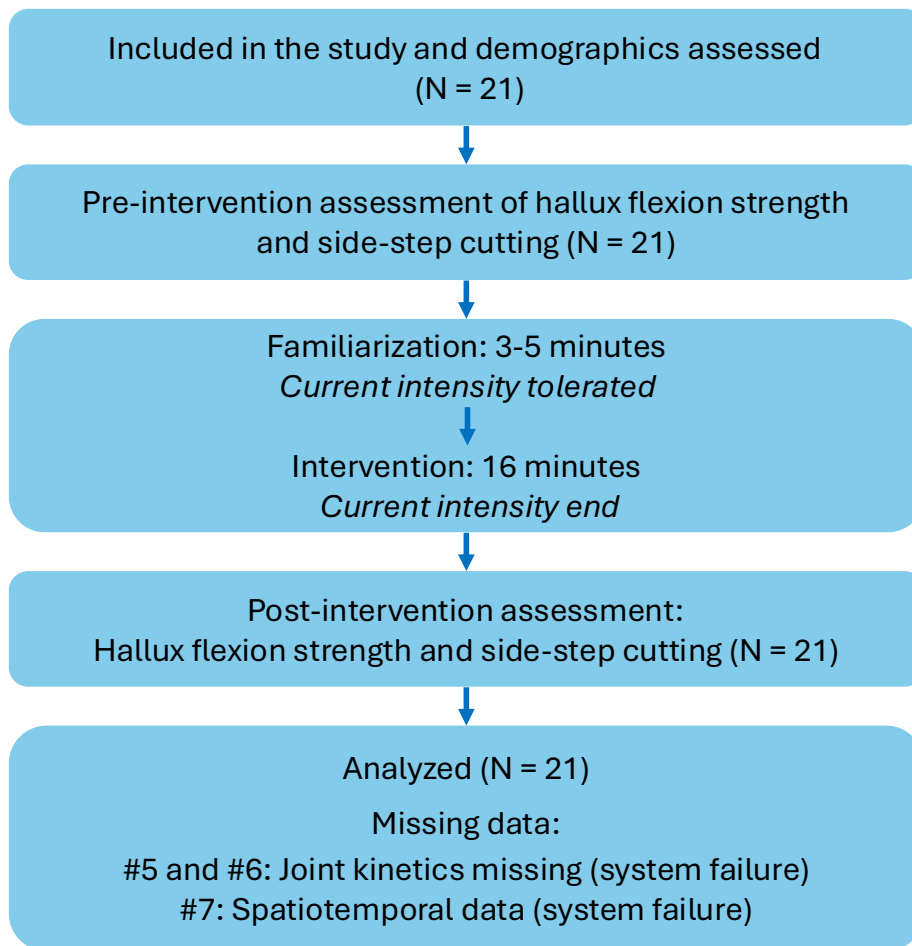


Figure 29. Study flow chart. N (number), # (participant identifier)

4.1. Interventions

Participants underwent a 16-minute protocol of isometric toe flexor contractions with concomitant neuromuscular electrical stimulation, while standing on flat ground. Two 5 × 5 cm electrodes (EN-Trode, Enraf-Nonius®, Malle, Belgium) were placed between the medial process of the calcaneus and the head of the first metatarsal (Figure 30).²⁴⁵ Therefore, the reflective markers placed on the navicular tuberosity and first metatarsal head were removed. A biphasic current was delivered using a Tensmed S84 device (Arseus Medical®, Bornem, Belgium) at 30 Hz, with a pulse width of 400 µs, a ramp-up time of 1.5 seconds, and a ramp-down time of 0.5 seconds, alternating between 8 seconds of stimulation and 4 seconds of rest. To apply external load on the feet, participants wore a weighted vest (20 kg) and a bar across the shoulders (15 kg), with weight removed from the vest so that the total load did not exceed 50% of the participant's body weight. A physiotherapist trained in neuromuscular electrical stimulation supervised the intervention.

Participants first completed a 3–5 minute familiarisation session to practice pressing their toes downward during stimulation and relaxing during rest. This session also helped determine the highest tolerable current intensity. The intervention began at this intensity, with participants encouraged to increase it progressively while maintaining active toe flexion throughout. Breaks were permitted if participants experienced significant discomfort.



Figure 30. Placement of the adhesive electrodes used to deliver the electrical stimulation. Note that the retroreflective marker of the navicular drop was removed to allow the placement of the posterior electrode.

4.2. *Strength assessment*

Hallux flexion strength was measured in a long-sitting position. To promote IFM activation over extrinsic foot muscle involvement, participants were positioned supine on an examination table, supported on their elbows, with the tested foot placed at the edge, the knee bent, and the ankle in maximal plantarflexion without discomfort.¹⁴² The hand-held dynamometer was stabilized on the assessor's knee, with the transducer placed under the second phalanx of the hallux. The evaluator manually stabilized the ankle. Participants were instructed to press their hallux downward with maximal effort for five seconds while maintaining contact between their back and the table.¹³⁹ Trials were repeated until peak force plateaued, with a variability of less than 20% across the final three attempts, and the average of the two highest values was used for analysis.

4.3. Side-step cutting task

Assessments were conducted in the Clinical Motion Analysis Laboratory at University Hospitals Leuven. Participants performed anticipated side-step cutting at a 45° angle while being barefoot (Figure 31 and Video file 6 in *Chapter 4*). After two practice trials, a static calibration was recorded, followed by five valid trials. A trial was considered valid if the participant approached the plate at $3.3 \text{ m}\cdot\text{s}^{-1} \pm 10\%$, landed fully on the pressure plate, and executed a contralateral 45° change of direction.

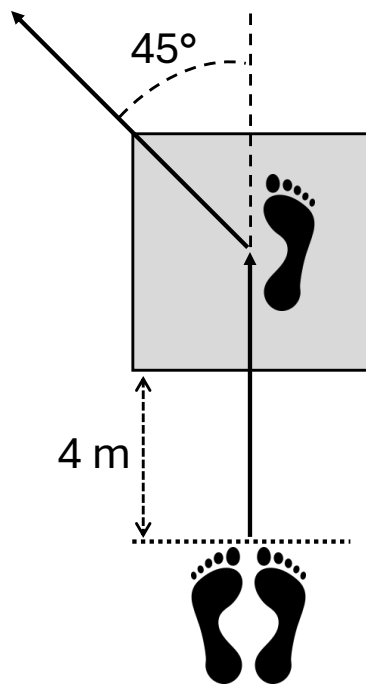


Figure 31. Side-step cutting. Participants started 4 meters away from the pressure plate (grey square), ran in a straight line at a target speed of $3.3 \text{ m}\cdot\text{s}^{-1} (\pm 10\%)$ and performed a 45° change of direction as indicated by white tape on the ground.

5. Data processing

Foot-strike and toe-off events were manually identified, and data were processed using the IOR-4Segment model on Nexus 2.12 software (VICON, Oxford Metrics, Oxford, UK). Data were then analysed using a custom MATLAB-based program (ACEP_Manager, Leuven, Belgium). Joint angles and angular velocities were computed in all three anatomical planes. Internal joint moments were estimated using inverse dynamics based on the Newton-Euler equations, assuming a mass distribution of 30/30/30/10% for proximal to distal foot segment. At each time frame, vertical ground reaction forces were allocated to each segment proportionally to the corresponding plantar pressure distribution.¹⁰⁹ Joint power was calculated as the scalar product of joint moment and angular velocity. Each trial was time-normalised to 100% of the stance phase, and the participant's mean curve was used for further analysis.

6. Outcome measures

6.1. *Primary outcome measure*

The primary outcomes included frontal plane ankle ROM during side-step cutting, frontal plane ankle peak moment, and ankle peak power across the three planes. ROM values were calculated within three subphases of stance, divided by vertical ground reaction force events: impact peak phase (0-6% of stance), active peak phase (7-45%), propulsive phase (46-100%).⁸⁶

6.2. *Secondary outcome measures*

Secondary outcomes included hallux flexion strength, and side-step cutting spatiotemporal outcomes and the multi-segment foot joint kinematic and kinetic outcomes not detailed as primary outcomes.

Foot strike pattern was identified using the strike index, which quantifies the region of the foot that first contacts the ground during initial stance.²⁹⁷ Additional spatiotemporal variables included approach speed, cadence, step time, and step length. The orientation of the foot during the stance phase was also computed as the mean foot progression angle during the active peak phase.

Secondary kinematic and kinetic outcomes included sagittal and transverse planes ankle ROM and peak moments; Chopart and Lisfranc ROM and peak moment in all three planes; sagittal plane hallux ROM and peak moment; and Chopart, Lisfranc and hallux peak power and work (sum of three planes). Joint

moment, power and work were normalised to body weight. Ground reaction forces in the vertical, anteroposterior and mediolateral directions were also obtained and normalized to body weight.

7. Statistical analysis

Zero-dimensional data were analysed using R (version 4.5.1), with the significance level set at $\alpha = 0.05$ and categorical variables were reported as counts. Shapiro-Wilk test was used to assess normality. Normally distributed variables were analysed using paired t-tests and Cohen's d was used as standardized effect sizes, interpreted as <0.2 very small, <0.5 small, <0.8 moderate, <1.2 large, <2 very large, ≥ 2 huge.²⁰¹ Non-normally distributed data were analysed using Wilcoxon signed-rank tests and r standardized effect size was reported ($r = Z/\sqrt{N}$), interpreted as <0.3 small, <0.5 moderate ≥ 0.5 large.³¹¹ To account for multiple comparisons on joint ROM values across subphases of stance, p-values were adjusted by multiplying them by three.

Time-series of kinematic and kinetic data were compared between pre- and post-intervention using Statistical Parametric Mapping (SPM) or its non-parametric equivalent (SnPM) implemented via open-source MATLAB code.¹²² The analysis included a normality check followed by a two-tailed SPM or SnPM t-test, with $\alpha = 0.05$.

Results

All participants performed the repeated contractions per protocol, with a mean electrical stimulation intensity of 15.4 (4.5) mA after the familiarization phase and 19.6 (7.0) mA at the end of the intervention. Joint kinetic results (joint moment, power, work) were missing in two participants, and spatiotemporal results were missing in one other participant, due to failure with the capture system (Figure 29).

1. Primary outcome measures

No significant changes were observed for frontal plane ankle ROM and peak moment, and ankle peak power (Table 17).

Table 17: Primary outcomes, including frontal plane ankle range of motion and peak moment, and ankle peak power

Variables	Pre	Post	p-value	Effect size (95% CI)
Inv./Ev. ROM: P1 (°) ‡	1.2 [0.7 – 1.9]	1.1 [0.6 – 2.2]	1	0.09 (-0.49 – 0.32)
Inv./Ev. ROM: P2 (°) †	3.5 (1.7)	3.2 (1.7)	1	0.16 (-0.27 – 0.59)
Inv./Ev. ROM: P3 (°) ‡	3.6 [3.1 – 6.3]	3.2 [3.0 – 3.9]	0.14	0.44 (0.03 – 1.79)
Peak Inv. moment (Nm·kg ⁻¹) ‡	0.17 [0.12 – 0.32]	0.20 [0.11 – 0.35]	1	0 (-0.04 – 0.03)
Peak Ev. moment (Nm·kg ⁻¹) †	-0.13 (0.10)	-0.15 (0.11)	0.19	0.32 (-0.15 – 0.77)
Peak gen. power (W·kg ⁻¹) †	12.5 (3.3)	12.1 (3.0)	0.36	0.21 (-0.24 – 0.67)
Peak abs. power (W·kg ⁻¹) †	-5.1 (2.4)	-5.4 (2.3)	0.56	0.14 (-0.32 – 0.59)

Legend: Data are presented as mean (standard deviation) when normality was not rejected (†), and as median [quartile 1 – quartile 3] when normality was rejected (‡). Results of the paired tests are reported as the p-value and the standardized effect size, i.e., Cohen's d or Wilcoxon r, depending on normality. CI (confidence interval), Ev. (eversion), gen. (generation), inv. (inversion), neg. (negative), P1 (impact peak phase), P2 (active peak phase), P3 (propulsive phase), pos. (positive), ROM (range of motion).

2. Secondary outcome measures

2.1. *Hallux flexion strength*

Hallux flexion strength showed a significant, moderate decline from 87.6 (24.7) N before the intervention to 76.7 (27.9) N after the intervention ($p = 0.01$, Cohen's $d = 0.6$, confidence interval = 0.13 – 1.06).

2.2. *Spatiotemporal outcomes*

Participants performed side-step cutting per protocol with a mean approach speed of 3.5 (0.4) m·s⁻¹ before and 3.3 (0.4) m·s⁻¹ after the intervention (Appendix 5.A). Before the intervention, 6 out of 21 participants exhibited a rearfoot strike pattern and 15 out of 21 a midfoot strike pattern. After the intervention, 8 out of 21 participants exhibited a rearfoot strike pattern and 13 out of 21 a midfoot strike pattern. Cadence had a small decrease after the intervention ($p = 0.04$), whereas step time showed a moderate increase ($p = 0.03$).

2.3. *Foot joint kinematics and kinetics*

Sagittal plane Lisfranc ROM moderately decreased during the active peak phase (pre = 2.6 (1.7), post = 1.8 (1.0)°, $p = 0.02$), and sagittal plane hallux ROM moderately increased during the propulsive phase (pre = 40.2 (7.3), post = 42.8 (8.0)°, $p = 0.02$). Transverse plane ankle peak moments differed, with moderately lower adduction peak moment (pre = 0.15 (0.06), post = 0.12 (0.07) Nm·kg⁻¹, $p = 0.003$) and moderately increased abduction peak moment (pre = -0.17 (0.09), post = -0.22 (0.12) Nm·kg⁻¹, $p = 0.02$). Posterior ground reaction peak forces moderately increased after the intervention (pre = 30.0 (5.6), post = 32.6 (5.0) % of body weight, $p = 0.03$). Other secondary outcome measures are presented in Appendix 5.B.

2.4. Statistical parametric mapping

Significant findings from the SPM and SnPM analyses are detailed in Table 18. Most notable findings were a more plantarflexed Chopart joint during the impact peak phase (21-26% of stance, Figure 32) and ankle transverse plane moment in the propulsive phase being in abduction instead of adduction (65–97 % of stance, Figure 33). Other differences were identified in ankle, Chopart and Lisfranc joint moments at the end of the propulsive phase (Figure 33). Foot joint power did not differ significantly after the intervention (Figure 34), but medial ground reaction forces increased at the end of the propulsive phase (Figure 35).

Table 18: Significant findings from statistical parametric or non-parametric mapping analyses

Outcome	Normality	Critical t*	Critical t* exceeded	p-value
Sagittal plane Chopart angular position	Satisfied	± 3.23	21–36 %	0.004*
Frontal plane ankle moment	Satisfied	± 3.42	95 %	0.050*
Transverse plane ankle moment	Violated	± 3.07	65–97 %	0.001*
Transverse plane Chopart moment	Satisfied	± 3.40	93–96 %	0.038*
Frontal plane Lisfranc moment	Violated	± 3.10	91–95 %	0.007*
Mediolateral ground reaction forces	Violated	± 3.14	80–100 %	0.001*

Legend: t* (critical value for SPM(t) or SnPM(t)), * (significantly different at $p < 0.05$ level).

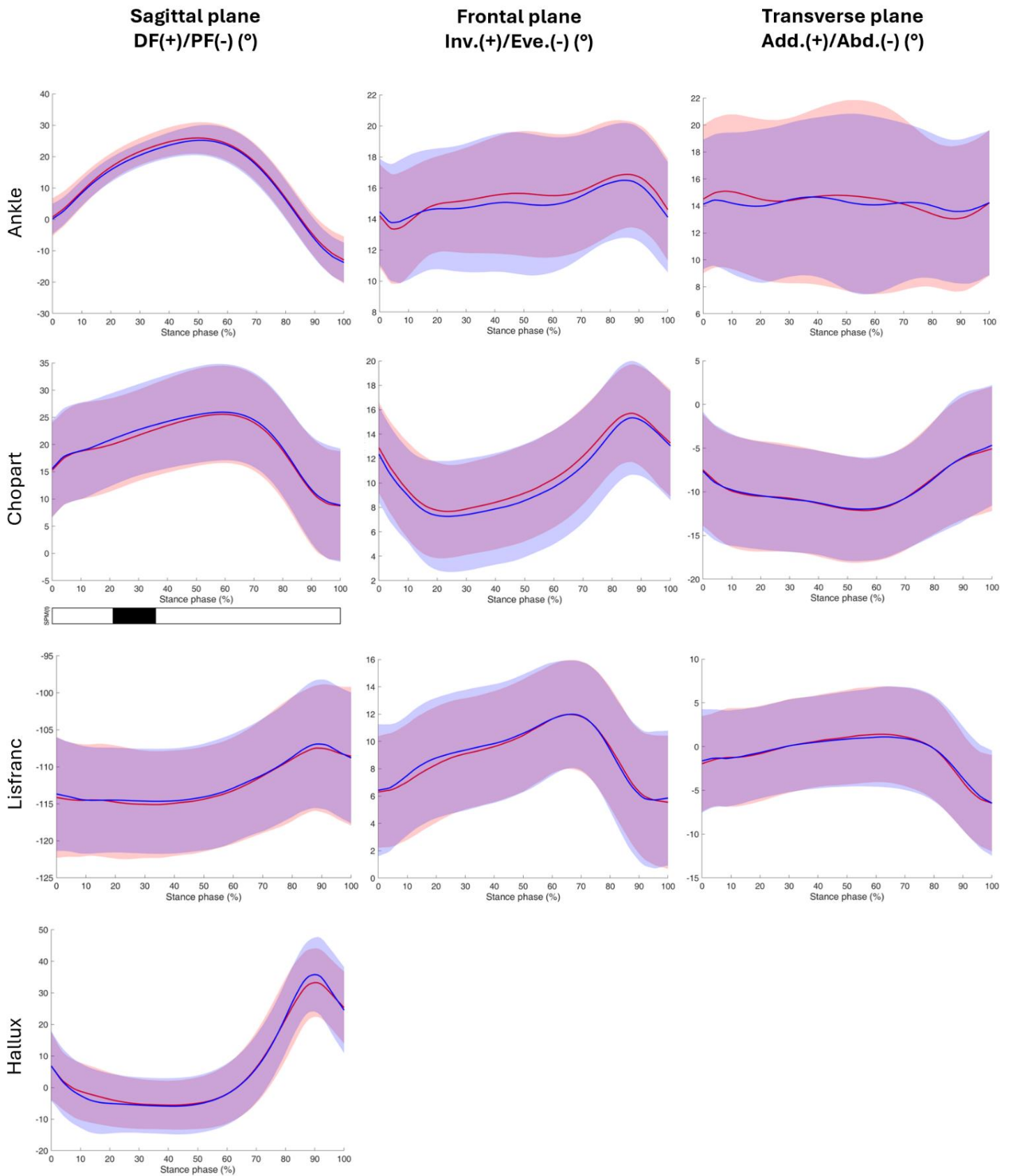


Figure 32. Foot joint angular positions in all three planes. Mean (line) and standard deviation (shaded area) before (red) and after (blue) the intervention.

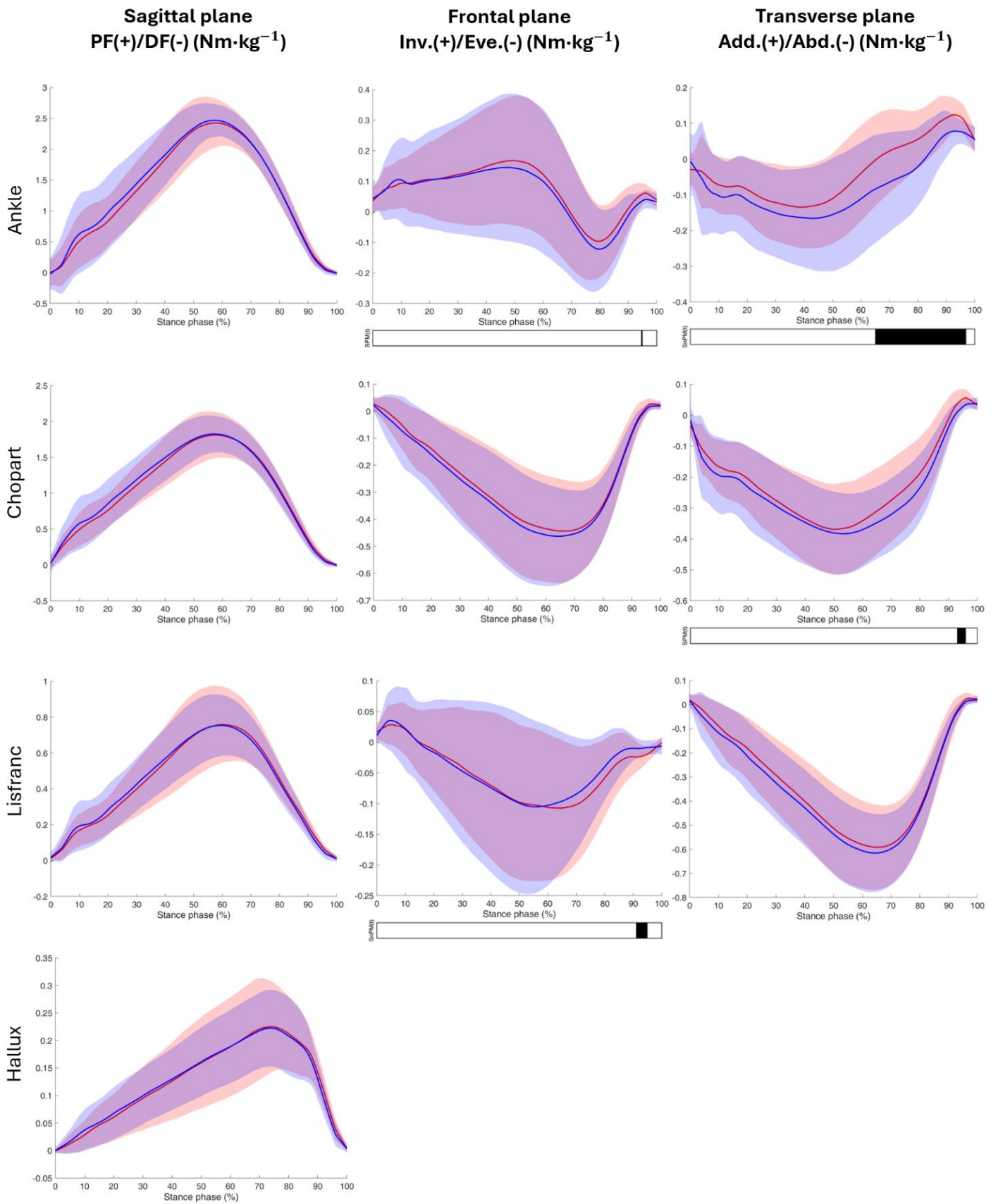


Figure 33. Foot joint moments in all three planes. Mean (line) and standard deviation (shaded area) before (red) and after (blue) the intervention.

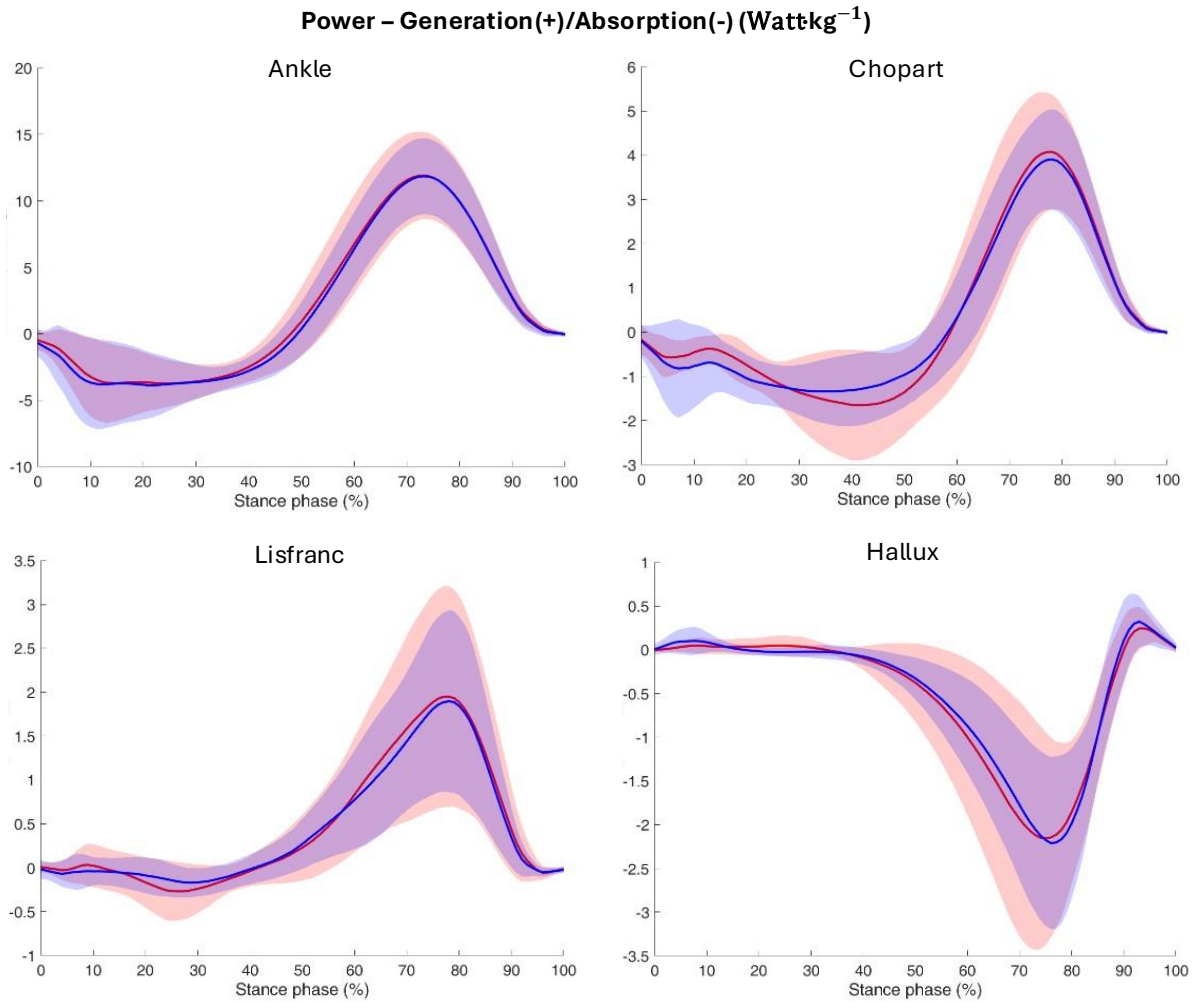


Figure 34. Foot joint powers. Mean (line) and standard deviation (shaded area) before (red) and after (blue) the intervention.

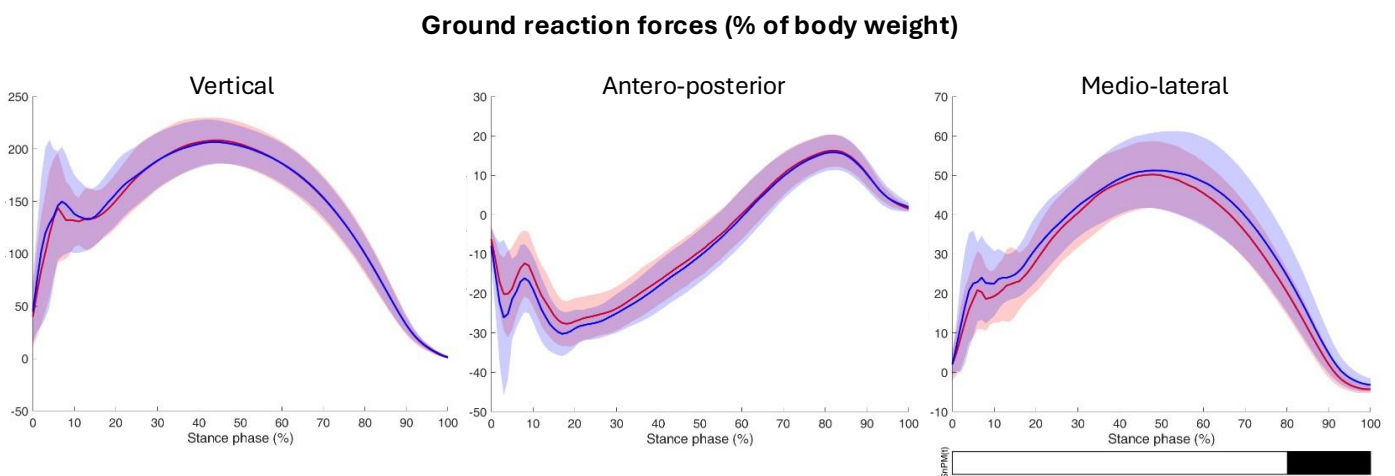


Figure 35. Ground reaction forces in all three planes. Mean (line) and standard deviation (shaded area) before (red) and after (blue) the intervention.

Discussion

This study is the first to examine the immediate effects of isometric toe flexions combined with neuromuscular electrical stimulation on side-step cutting foot joint kinematics and kinetics in individuals with CAI. Contrary to our hypothesis, the intervention did not alter frontal plane ankle ROM and peak moment, nor ankle power.

Still, the intervention elicited acute adjustments in various foot joint ROM, peak moments, and curve analyses of foot joint moments. First, ankle peak adduction moment was reduced, confirmed by the decreased net adduction moment at 65-95% of stance, along with a subtle decrease in ankle inversion moment in late stance (95%), and non-significant trends for more everted positions of ankle and Chopart joints during stance. These alterations may originate from an increased activity of fibularis longus muscle and decreased activity of the medial IFMs. Indeed, fibularis longus inserts on the first metatarsal head and medial cuneiform bone, and is responsible for eversion of the Ankle, Chopart and Lisfranc joints, thus maintaining the medial longitudinal arch close to the ground during propulsion.²⁴⁰ This interpretation needs further validation, by recording the electromyographic activity of extrinsic foot muscles and IFMs. Second, Chopart joint exhibited a more dorsiflexed position (21-36%) and sagittal plane Lisfranc ROM was moderately lower during the active peak phase. These changes advocate for redistribution of motion across the joints responsible for medial longitudinal arch motion. Third, hallux ROM increased during the propulsive phase, but joint moment did not change. Therefore, hallux joint stiffness likely decreased after the intervention, as a result of altered IFM activity. This is consistent with the decreased metatarsophalangeal joint stiffness observed during running propulsion under tibial nerve block.¹⁹ This increased metatarsophalangeal mobility decreases the length of the lever arm of the ground reaction force to the ankle, while that of the ankle extensors is kept constant, resulting in a decreased gear ratio, which is regarded as suboptimal for propulsion.^{23, 312} Lastly, posterior ground reaction peak force increased, reflecting increased braking in early stance. This may be explained by the 3% cadence reduction,³¹³ although the slightly lower approach speed should lead to decreased vertical and posterior ground reaction forces, not increased.³¹⁴ Besides, medial ground reaction forces increased from 80-100% of stance, and may partly be the result of increased step time. Together, these

changes in ground reaction forces suggest that an alteration in IFM activity led to modified interactions between the foot and the ground during braking in early stance and finishing the change of direction in late stance. Overall, the subtle differences in foot joint movements and loading patterns may point towards altered forefoot and hindfoot biomechanics in participants with CAI following a single session of isometric toe flexor contractions with IFM electrical stimulation.

This is, to our knowledge, the first study to assess acute changes in foot strength after repeated isometric toe flexions combined with electrical stimulation. The rationale to implement that precise protocol was the following: sustained isometric hallux flexion and abduction until failure resulted in detectable fatigue,¹⁵⁹ but such a sustained contraction can only be performed for a few minutes before experiencing muscle cramp or transient fatigue.¹⁵⁹⁻¹⁶⁰ Since we aimed to conduct that protocol over 16 minutes, we selected an intermittent isometric protocol. Electrical stimulation was concomitantly administered to assist voluntary contraction¹⁶⁷ and because a previous study found foot muscle fatigue after electrical stimulation used as a standalone intervention.¹⁵⁸ The precise current characteristics that we selected previously induced fatigue at the shoulder flexors,¹⁶⁸ and the duration of the 'On' periods was doubled to obtain a more demanding duty cycle. The protocol was also conducted while standing and wearing additional load to favour spontaneous IFM activation when the foot is loaded.¹⁶⁵

Despite this rationale, we found a moderate, 12% mean decrease in hallux flexion strength, below the 26-46% minimal detectable change,^{139, 141} indicating inconsistent neuromuscular fatigue. About one-third of participants showed no change or even gains, while most exhibited $\geq 15\%$ declines, a variability consistent with previous reports of non-responders to electrical stimulation.²⁴⁵ Mean current intensity was considerably lower among our participants compared with previous studies,^{245, 315} which we mainly explain by participants being instructed to experience discomfort – but no pain – as specified in the protocol. Besides, electrical stimulation frequency and contraction modalities may explain the low current tolerated and the inconsistent fatigue. Low frequencies are more likely to result in decreased muscular contractility without central nervous system involvement compared with higher frequencies.¹⁶⁹ However, low frequency currents also involve more discomfort and pain compared to high-frequency

currents set at the same absolute intensity.²⁵² Besides, evidence from knee extensors suggests that concentric contractions may induce greater fatigue than isometric ones of similar duration.^{249, 316} However, caution is needed because we used a longer duty cycle (8 s on/4 s off) compared with those typically used at the knee (2-3 s on/2 s off), and sustained isometric knee extensor contractions previously resulted in similar or even higher fatigue compared with repeated concentric contractions.^{250, 317} Therefore, the impact of contraction type and duty cycle on IFM fatigue remains unclear and warrants direct comparison in future trials.

Few studies have investigated the effects of interventions targeting the IFMs conducted over several weeks on foot joint dynamics.^{216, 226} For example, Matias, et al.²¹⁶ found increased ankle peak inversion and decreased medial longitudinal arch ROM during running following 8 weeks of a multimodal intervention targeting the IFMs, among healthy participants. The ‘rehabilitation-oriented assessment’ paradigm provides a valuable framework for understanding how IFM-targeted exercises may contribute to CAI management. To date, assessment and treatment addressing IFM function are not part of the standard rehabilitation of patients with CAI,⁴⁷ although IFM function may be impaired in those individuals.^{65-66, 79} In the context of emerging evidence of enhanced self-reported function following rehabilitation incorporating IFM exercises in CAI,³¹⁰ the current study provides unique clinical evidence regarding the acute effects of an exercise targeting the IFM performed in a loaded condition on foot joint dynamics. Frontal plane ankle kinematics – particularly threatened in an acute ankle sprain –³² were not altered during the sporting task, suggesting it can be safely incorporated into rehabilitation. Future studies could investigate the effects of IFM exercises on ankle and foot joint dynamics during more demanding sporting tasks, such as running or changing direction at higher speeds, to better understand the potential of these muscles in rehabilitation.

Several limitations need to be considered. First, the absence of a control group or counterbalancing limits internal validity, as the observed changes may arise from within-subject variability or test-retest effects, rather than true intervention effects. Still, strength test-retest reliability is good to excellent,¹³⁹ and a majority of reflective markers were left in place during the intervention to maximise the reliability of kinematic data. The task could also be optimized in order to

consistently achieve neuromuscular fatigue, by allowing pain during electrical stimulation. Second, findings apply to recreationally active young males performing anticipated barefoot side-step cutting at moderate speed, limiting generalisability. Participants also anticipated the task to land on the pressure plate with the correct foot, which also influences side-step cutting biomechanics.³¹⁸ Finally, the study was powered for a broader observational, between-group design. Yet our paired approach enhances sensitivity compared with a between-group design, supporting the robustness of conclusions from the 21 participants.

Future research should investigate side-step cutting foot joint kinematics and kinetics following interventions conducted over several weeks, along with investigating foot biomechanics during demanding sporting tasks relevant to athletes with CAI, such as changes of direction at maximal speed, sprinting and jumping. Direct comparisons of isometric versus concentric contractions under varying duty cycles are also needed, ideally in randomized trials, and incorporating real-time biofeedback could optimize effort and allow objective monitoring of the provided efforts.

Conclusion

Side-step cutting frontal-plane ankle ROM and peak moments were not affected by a sixteen-minute intrinsic foot muscle exercise combined with electrical stimulation of the medial longitudinal arch, among individuals with CAI. However, this intervention elicited ankle kinematic and kinetic changes, potentially associated with modified intrinsic and extrinsic muscle activity. In early-stance to mid-stance, Chopart joint was more plantarflexed and Lisfranc joint mobility was lower, potentially indicating modified medial longitudinal arch mobility. Mobility of the first metatarsophalangeal joint increased in late stance, suggesting altered stiffness. These findings suggest that exercises targeting the intrinsic foot muscles may acutely influence foot joint biomechanics without compromising frontal-plane ankle control during changes of direction. Incorporating such exercises into rehabilitation may therefore be safe, while future studies should investigate more demanding sporting tasks and compare contraction modalities to clarify how intrinsic foot muscles contribute to rehabilitation among individuals with CAI.

Acknowledgements

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Summary

Context

The acute effects of interventions targeting the foot muscles on side-cutting foot joint dynamics of individuals with CAI were unknown. Besides, no previous study tracked foot muscle fatigue after such an intervention among individuals with CAI.

Findings and interpretation

Twenty-one recreationally active young males with CAI performed repeated isometric toe flexor contractions combined with electrical stimulation. This was done during 16 minutes while standing and wearing 35 kg of additional load. Hallux flexion strength had a moderate decline (10 N) after the intervention. Participants achieved a relatively low current intensity – probably due to the instruction to not experience pain, current characteristics, and non-responding to electrical stimulation.

After the intervention, frontal plane ankle kinematics remained unchanged during the 45° side-step cutting. Therefore, we suggest that changes of direction are safe among individuals with CAI. Still, various differences in foot joint dynamics were identified:

- We observed a decreased ankle adduction moment, a decreased ankle inversion moment, and trends for more everted ankle and Chopart joint positions. We hypothesise that these changes in hindfoot and midfoot joint dynamics originated from an increased activity of fibularis longus and decreased activity of the medial intrinsic foot muscles.
- Chopart joint was more dorsiflexed in early stance, Lisfranc joint ranges of motion decreased in early-stance to mid-stance, while hallux metatarsophalangeal joint ranges of motion increased in late stance. These changes suggest changes in joint stiffness.
- Posterior ground reaction forces increased after the intervention, which echoes the results of *Chapter 4*, where individuals with CAI had lower posterior forces compared to healthy controls. After the isometric protocol, individuals with CAI also exhibited reduced medial ground reaction forces in late stance.

Implications for clinical practice and future research

The aforementioned low current intensity likely contributed to the lack of detectable fatigue. Strength decline was lower than expected, which is why we conducted a cross-over study comparing the isometric protocol to a concentric protocol (see *Chapter 2*).

When investigating fatigue, future studies should include a control group. Researchers should also record electromyographic activity of the foot muscles during various sporting tasks in order to better understand their functions.

General discussion

I. Summary of findings

Ankle sprains are highly prevalent in the physically active general population and among athletes.^{28, 33-34} Approximately 40% of the cases develop chronic ankle instability (CAI) while those who recover are called ‘copers’.³³ CAI is associated with altered lower-limb movement patterns during locomotion, and altered sizes and strength of the extrinsic foot muscles.³⁹ However, foot joint movement patterns are mostly unexplored during sporting tasks, and it is unclear whether alterations in intrinsic foot muscle (IFM) functions are common in individuals with CAI.

Therefore, the clinical objectives of this PhD project were to evaluate foot joint kinematics and kinetics during side-step cutting among individuals with CAI compared to ankle sprain copers and healthy controls, and to study IFM functions in terms of sizes, strength and biomechanical effects of neuromuscular fatigue. Methodological objectives were formulated to achieve these clinical objectives. First, we reviewed the effects of interventions aiming to strengthen the IFMs on IFM sizes, foot strength and locomotion biomechanics. Second, we aimed to validate a low-end ultrasound device to measure selected IFM sizes. Third, we compared the effects of two fatigue protocols on foot strength and ground reaction force-related outcomes during running.

In Chapter 1, pooled data through meta-analysis showed that IFM exercises and minimalist running shoes (MRS) were associated with increased IFM sizes and foot strength, with trivial to important effect sizes. Besides, IFM exercises resulted in reduced mobility of the medial longitudinal foot arch during running, while MRS lead to a more forefoot-oriented running strike pattern. Besides, the large majority of randomized controlled trials that implemented these interventions had a high risk of bias.²⁵¹ *Chapter 2* was undertaken because preliminary analyses in *Chapter 5* revealed inconsistent fatigue following repeated isometric toe flexor contractions. This cross-over randomized trial suggested the superiority of a concentric- over the isometric protocol to induce fatigue. *Chapter 3* highlighted the clinical applicability of a low-end ultrasound device to measure IFM sizes,

especially in test-retest designs. Measurement-based inter-rater reliability was also moderate to good, but global inter-rater reliability was poor to excellent.²⁹¹

The studies related to *Chapters 4 and 5* were conducted in Clinical Motion Analysis Laboratory (CMAL, KU Leuven). During anticipated side-step cutting, individuals with CAI exhibited subtle differences in foot joint kinetics, compared to healthy controls and copers. We also found lower posterior ground reaction forces in the first half of stance in CAI, suggesting altered deceleration abilities. Flexor digitorum brevis muscle sizes were partly explained by body mass index, and none of abductor hallucis and flexor digitorum brevis sizes were significantly explained by being classified as CAI, coper or control. However, hallux flexion strength was lower in CAI and copers compared to healthy controls. Among individuals with CAI, repeated isometric toe flexor contractions combined with electrical stimulation did not result in altered frontal plane hindfoot dynamics, challenging our hypotheses. Still, it did modify Chopart and Lisfranc sagittal plane mobility. Hallux motion in late stance and posterior ground reaction forces increased following the intervention, while hallux flexion strength moderately decreased.

II. Roadmap of the discussion

While each chapter provided insights into specific aspects of IFM function and ankle/foot biomechanics, the overall picture suggests that CAI cannot be attributed to a single deficit. Instead, the interplay of structural, perceptual and motor factors must be considered. Accordingly, the discussion will argue the central thesis that the clinical presentation of an individual with CAI arises from the complex interactions between multi-modal impairments, under the influence of personal and environmental factors.

In the following sections, we first extend altered ankle and foot movement patterns to the entire lower-limb kinetic chain, along with the influence of personal and task-related confounding factors. We also balance the benefits and drawbacks of using a multi-segment foot model (MSFM) in clinical research, and perspectives of technological and methodological developments. Next, we broaden the applicability of ultrasound imaging in ankle research, we elaborate on the current understanding of foot muscle function, and we discuss altered IFM function in CAI. Then, we briefly examine muscle fatigue and electrical stimulation. Lastly, we contextualize our findings within the multi-factorial origin of CAI, before addressing limitations and perspectives for future research.

III. Lower-limb kinematics and kinetics

Movement patterns during sporting tasks in CAI

Movement patterns of the ankle and foot joints among people with CAI

Changes of direction are part of many sports and place important stress on ankle tissues. In *Chapter 4*, individuals with CAI, copers and healthy controls performed anticipated 45° side-step cutting with an approach speed of 3.5 m·s⁻¹. Foot joint kinematic patterns were highly similar between groups, but we observed subtle alterations in moment and power at various foot joints, associated to reduced posterior ground reaction forces in individuals with CAI. To our knowledge, this was the first study to use a kinetic MSFM in individuals with CAI, providing evidence of subtle kinetic differences that were not captured by foot joint kinematics.

While our study focused on side-step cutting, a few studies have investigated ankle and foot joint kinematics during running at moderate speeds.^{84-86, 302} Their results were detailed in the general introduction (*cfr.* p. 51). Shortly, particular alterations were found at the hindfoot in all three planes, and other subtle alterations were found at Chopart and Lisfranc joints, with a general trend towards higher ranges of movements or amplitudes in CAI compared to controls.⁸⁴⁻⁸⁵ Cutting tasks impose different mechanical demands than steady-state running, and studies using cutting protocols therefore provide additional perspectives on ankle and foot biomechanics in CAI.

A single study, to our knowledge, involved a similar task to ours and reported reduced maximum ankle inversion in CAI without joint laxity compared to controls.¹³⁰ In other studies, participants were instructed to hop or jump forward while standing one step away from the force plate, and to perform the cutting task. One of these studies used a kinematic MSFM among female soccer players with and without CAI during cutting at 45, 90 and 135°. It reported no between-group differences in ankle/foot kinematics, though CAI had longer stance times.⁸³ During forward jumping followed by 45° side-step cutting and running as fast as possible, individuals with CAI had higher plantarflexion moment in early stance, lower ankle eversion moment at mid-stance,²⁸⁹ and a more inverted ankle position in late stance.¹³² Similarly, during 90° jump/cutting performed as fast as possible, Kim et al. found less ankle dorsiflexion during most of the stance phase, and more ankle inversion during early and late periods of stance, in a study comparing 100

individuals with CAI to 100 healthy controls.³¹⁹ Two other studies consistently found less positive and negative ankle work in individuals with CAI than in controls.³²⁰⁻³²¹ Overall, individuals with CAI seem to exhibit alterations in sagittal- and frontal-plane ankle kinematics and kinetics during cutting tasks. However, single-segment foot models overestimate ankle dynamics,¹⁰⁴⁻¹⁰⁵ underscoring the need for more research using MSFMs to refine our understanding of ankle and foot biomechanics in CAI.

Altered lower-limb neuromuscular control strategy among individuals with CAI

Beyond the ankle and foot, evidence suggests that CAI affects the entire kinetic chain, with impairments observed in both knee and hip joint kinematics and kinetics. During 45° jump/cutting, the hip was more flexed and abducted in early to mid-stance in CAI compared to controls.^{132, 322} During 90° jump/cutting, Kim et al.'s study reported more knee and hip flexion from foot strike to late stance and more hip abduction in mid-stance in CAI.³¹⁹ Regarding lower-limb kinetics, individuals with CAI showed lower ankle power, but higher knee and hip powers during initial landing of 90° jump/cutting compared to copers and healthy controls,³²¹ and they also showed higher knee and hip negative work, reflecting higher energy dissipation.³²⁰ Neuromuscular control is likely also impaired during the preparation phase of cutting tasks. In the 200ms preceding foot strike of 90° jump/cutting, consistent evidence shows greater ankle inversion,¹³² knee flexion, and hip flexion in CAI than in controls.³²³

These joint dynamic adaptations also present with differences in muscular activity during sporting tasks.³²³⁻³²⁴ During 90° jump/cutting, participants with CAI had lower activities of peroneus longus, medial gastrocnemius, vastus lateralis, adductor longus, gluteus medius and gluteus maximus, compared to healthy controls. Copers showed increased activity of some of these muscles, that was interpreted as an efficient protective strategy that individuals with CAI failed to develop.³²³

Systematic reviews reported that dynamics of the pelvis and trunk are rarely investigated in CAI.³²⁵⁻³²⁷ Still, a restricted number of studies found increased trunk and pelvis rotation during a dynamic balance test,³²⁸ increased lateral trunk

flexion,³²⁶ and more variability in trunk movements during single leg landing in individuals with CAI compared to healthy controls.³²⁹ No study, to our knowledge, assessed motor control of these proximal body parts during side-step cutting tasks, which should thus be examined in future research.

Overall, these changes clearly show that compensatory neuromuscular control strategies redistribute constraints from the ankle to more proximal joints during sporting tasks, especially in the first part of the stance phase. During that period, negative work is needed to absorb energy. This echoes our findings related to posterior ground reaction forces in CAI in *Chapter 4* and *Chapter 5*. The increased posterior ground reaction forces found in *Chapter 5* may result from changes in lower-leg and foot muscle activity after repeated foot muscle contractions, that warrant further research. Besides, the reduced posterior ground reaction forces in CAI compared to controls (*Chapter 4*) may be due to a protective neuromuscular strategy, limiting energy absorption at the ankle. This may argue for including tasks that require repeated phases of acceleration and braking when assessing functional the status of individuals with CAI, such as the agility t-test.³³⁰

Clinical applicability and research perspectives

Personal and task-related factors influence these biomechanical patterns, which help explain the variability observed across previous research. Gender influences ankle, knee and hip biomechanics during side-step cutting,³³¹⁻³³² and females showed higher hindfoot and midfoot ranges of motion than males during running.³³³ Age also influences lower-limb biomechanics while running, especially at the ankle level.³³⁴ Whether gender and age influence foot joint biomechanics during side-step cutting still needs to be confirmed using MSFMs. Since most studies investigating foot joint dynamics in CAI included young adults, results among older participants are difficult to anticipate. Next, the unique alterations of each individual with CAI likely interact together and result in the observed movement patterns (see section VII. of the general discussion).^{39, 335-336}

Cutting tasks differ in starting position, execution speed, and cutting angle (*cfr.* p. 48, Figure 11 in *General introduction*). These parameters vary across studies and should be considered when interpreting results. Notably, most evidence on foot joint dynamics during running in individuals with CAI has been collected at

moderate speeds.^{84-86, 302} Our cutting task was performed at a similar submaximal speed, as *Chapter 4* and *Chapter 5* used a target speed of $3.3 \text{ m}\cdot\text{s}^{-1}$. This speed reflects many ecological, submaximal conditions typical of recreational athletes. Besides, this predetermined speed was already challenging for some participants, requiring multiple attempts to achieve a valid trial. We expect between-group differences in approach speeds if participants were asked to sprint, that would have biased lower-limb biomechanics.³³⁷ Besides, achieving higher speeds in CMAL laboratory could be difficult, given the 10-meter walkway and the sensitivity of the infrared cameras to vibration caused by abrupt braking against the walls. Cutting angle also influences lower-limb biomechanics.³³⁷⁻³³⁸ Future studies should therefore compare foot joint dynamics among CAI, copers and controls during cutting at higher speeds – with varying cutting angles and in unanticipated conditions – to better understand movement patterns and joint constraints in such settings.

Additionally, footwear condition (shod versus barefoot)³³⁹⁻³⁴⁰ and anticipation^{318, 341} further modulate lower limb biomechanics during running or cutting tasks. On-field situations also involve a high degree of chaos and low level of control.³⁰⁷ The external validity of controlled laboratory studies therefore remains limited, and future studies should aim to enhance ecological validity by introducing task-relevant visual and auditory cues, thereby increasing perceptual and decision-making demands during cutting tasks.

Limitations and perspectives of multi-segment foot models

General applications

MSFMs are used to describe foot joint movements and loading patterns by modelling the foot into 2-26 rigid segments.^{108, 290} They were developed because single-segment foot models overestimate ankle joint dynamics,¹⁰⁴ with ankle power inflated by 25% during running.³⁴² MSFMs have already been used during running at moderate speed,^{84, 112, 302, 343} side-step cutting at moderate speed (*Chapters 4-5*), side-step cutting at maximal speed,^{125, 127} jump/cutting⁸³ and jump landing.³⁴⁴ This task variety demonstrates the applicability of MSFMs during various sporting tasks. Despite their widespread use, MSFMs present practical challenges in research and clinical settings.

Practical limitations

MSFMs require a specific and expensive equipment (i.e. at least reflective markers, infrared cameras, and a dedicated software). Data reliability depends on extensive training from specialized clinicians,³⁴⁵ as well as time-consuming marker placement, data acquisition and data processing. For example, in our clinical studies occurring at CMAL (*Chapters 4-5*), approximately one hour was needed to place the markers according to '*IOR-4segment-model-1*'¹⁰⁹ and to obtain five valid side-step cutting and running trials per side.

Validity and reliability

The literature includes more than 40 kinematic MSFMs¹⁰⁸ and about 5 kinetic MSFMs,^{109, 113, 346-347} yet different MSFMs yield different kinematic and kinetic results.³⁴⁸ Clinicians may thus be lost when facing this variety of models. Validating MSFMs is difficult because gold-standard assessments of 'true' bony motion involve irradiation (i.e. biplane fluoroscopy) or invasive procedures (i.e. intracortical bone pins).^{108, 349}

Rizzoli MSFM¹¹⁰ appears as the most commonly used MSFM in CAI research. It is supported by good to excellent reliability for sagittal plane outcomes,^{104, 350} and moderate to excellent reliability in frontal and transverse planes.³⁴⁵ Its repeatability may be superior to that of other MSFMs,¹⁰⁸ although contradicting evidence also

exists.³⁵¹ Clinicians should be aware that repeatability is better when using ranges of motion (1.5-3.5° variability across walking trials) than when using absolute angular positions (2.2-6.5° variability), because the former minimizes the effect of marker placement error.³⁴⁵

Besides, we believe that reliability is also influenced by performance bias: investigators will theoretically achieve better reliability with the MSFM they are most familiar with, and research teams have an interest in promoting the model they developed or commonly use. Beyond general reliability concerns, additional limitations arise from the assumptions underlying kinetic MSFMs.

Kinetic modelling assumptions

Kinetic MSFMs compute net internal joint moment, power and work from joint kinematics and ground reaction forces. Their results rely on definitions of inertial properties of the geometric forms that approximate true anatomical segments. For example, the mass of foot segments were arbitrarily determined as 30/30/30/10% of total foot mass in the 'IOR-4segment-model-1' (Rizzoli MSFM).¹⁰⁹ These assumptions are likely close, yet imperfect representations of the reality,³⁵² and better approximations are difficult since anthropometric tables only distinguish one or two foot segments.³⁵³⁻³⁵⁶ Studies reporting foot segment masses *in vivo* or in cadavers would help refine the premises of kinetic MSFMs.

Aforementioned kinetic MSFMs apply inverse dynamics by dividing ground reaction forces according to the proportion of foot pressure below each segment.^{109, 113, 346-347, 357} For each foot segment, this 'proportionality scheme' computes frontal- and transverse-plane shear forces as a proportion of vertical forces, with the latest being directly measured by the pressure plate.^{109, 346} Although antero-posterior and medio-lateral shear forces do not exactly distribute in the same way as vertical forces,³⁵⁸ they only suffer from an error of 1-2% of body weight during walking, compared to using two adjacent force plates.³⁵⁷ As a result, the calculated foot joint peak moments and powers were close to those of the gold standard method.^{357, 359} These studies thus validated the proportionality scheme during walking using a three-segment foot model (hindfoot, midfoot including metatarsus, and hallux).^{346, 357} Validation is needed for four-segment kinetic foot

models that separate the metatarsal bones from the midfoot, and validation is also needed during running and multi-planar tasks such as cutting.

Recent advancements integrate direct measurements of antero-posterior and medio-lateral shear forces, overcoming the limitations of the ‘proportionality scheme’ to estimate anteroposterior and medio-lateral shear forces. These methods rely on the deformation of a stress sensitive film, captured by a camera positioned below a transparent glass. This technique also reveals ‘spreading’ phenomena, where opposing shear forces act between and within segments, that cannot be captured using the proportionality scheme.³⁵⁸ Finite element modelling may provide an even deeper understanding of constraints at the tissue level – within bones, joints and soft tissues. It integrates subject-specific anatomical geometries from medical imaging with experimentally acquired ground reaction forces, plantar pressures or tendon forces, to simulate stress and strain distributions.³⁶⁰⁻³⁶¹

Kinematic MSFMs rely on reflective markers placed on anatomical landmarks. Alternative methods such as biplane fluoroscopy capture in vivo bone-to-bone kinematics using dual X-ray imaging synchronized with 3D bone models.^{14, 362} It can be considered the gold standard for validating ankle and foot motion. However, radiation, high costs, complex calibration, and limited accessibility limit its applicability in research and clinical settings.³⁶³

IV. Ultrasound imaging as a useful clinical tool in CAI research

Ultrasound imaging is known from the general audience for its ability to capture foetal morphology during pregnancy (Figure 36), but it is also a very powerful tool for musculoskeletal research, with applications to clinical practice.



Figure 36. Longitudinal view of the whole body (top) and transverse view of the feet (lower part) of a foetus at 18 weeks of pregnancy. She is nicknamed ‘Crevette’ by her loving parents, desperately waiting to finally meet her in a few months.

Reliability of intrinsic foot muscle sizes

Ultrasound imaging is a non-invasive, widely available, and less costly alternative to magnetic resonance imaging.¹⁴⁷⁻¹⁴⁸ Most studies that assessed the reliability of ultrasound imaging to measure IFM sizes used mid- to high-end devices (i.e. \$10,000 – \$100,000).^{148, 258, 261, 265-266, 270} Such high-end devices typically offer better spatial resolution compared to cheaper devices (Figure 37).²⁸⁸

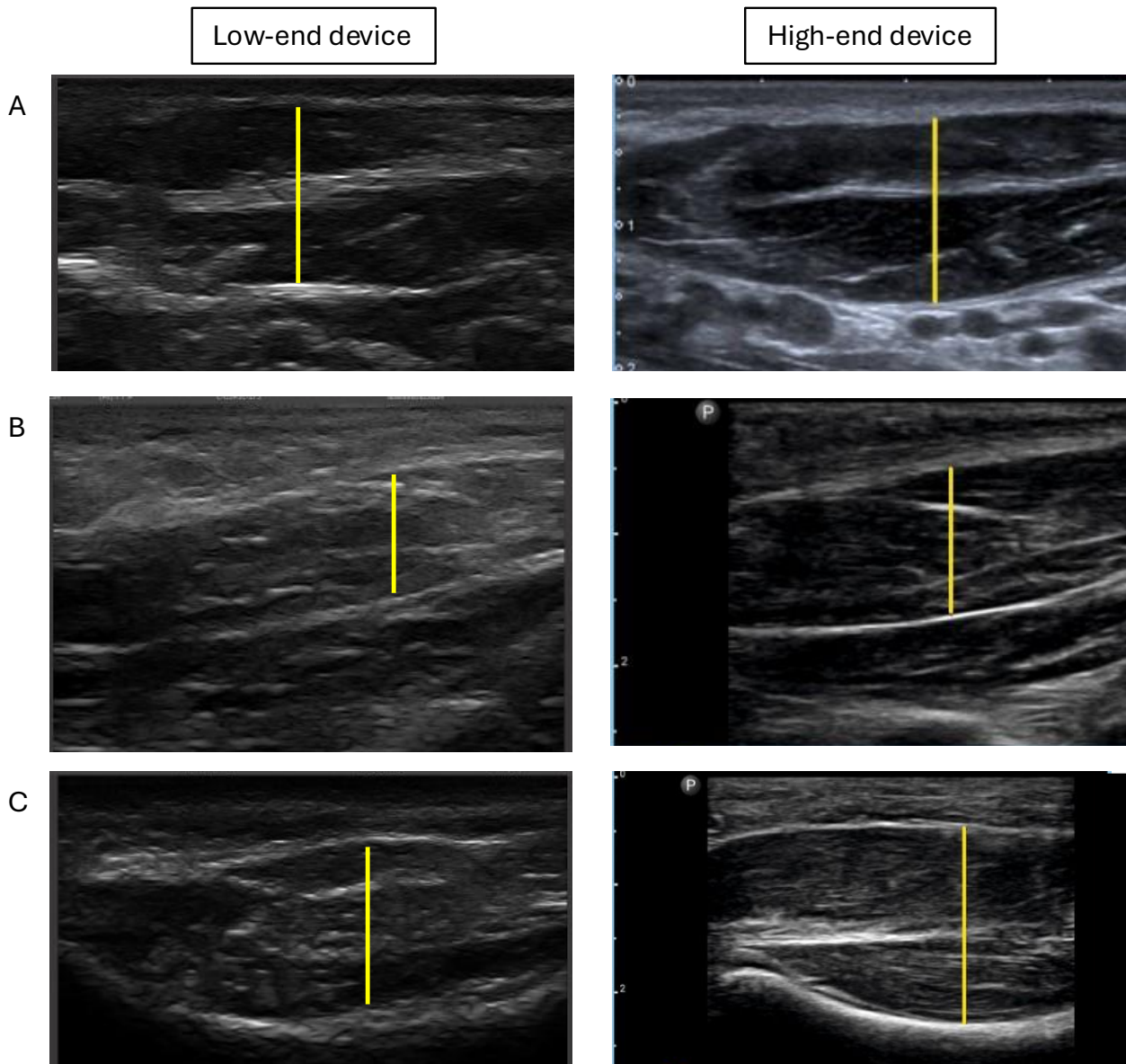


Figure 37. Comparison of representative images obtained with a low-end device (left column) versus those obtained with a high-end device (right column). Abductor hallucis thickness (A), flexor digitorum thickness (B) and flexor hallucis brevis thickness (C) are identified in yellow. Note the difference in image resolution, with more intramuscular and peri-muscular details visible in the right column. Images of the right column were reported by Willemse et al. (2022).²⁶⁰

Low-end devices – such as the *MicrUs EXT-1H* (Telemed UAB, Lithuania) used in *Chapter 3* – may constitute relevant alternatives to high-end devices.^{260, 268} Such low-end devices are already available in many laboratories and care centres worldwide, but the poorer image resolution could impact reliability. Besides, previous studies rarely conducted sample size calculation and often included a limited number of participants. Investigators typically drew landmarks on participants' skin during inter-rater acquisitions and did not enforce time constraints for data collection.¹⁴⁹ Consequently, these methodological choices may have led to an overestimation of ultrasound reliability in more ecological settings. Therefore, we aimed to assess ultrasound reliability using a low-end device with three raters (as recommended in guidelines)²⁶⁹ with a predefined acquisition schedule.

We found a majority of structures with moderate to good test-retest and measurement-based inter-rater reliability, while global inter-rater reliability was poor for many structures.²⁹¹ These results are worse than those found in our systematic review.¹⁴⁹ It is unknown to what extent our results are attributable to investigator experience, acquisition procedures, the use of a low-end device, or a combination of these factors. The study could have been strengthened by including a comparator condition, such as measurements obtained using a high-end device, without time constraints, or by an examiner with extensive ultrasound experience, allowing these sources of variability to be more clearly disentangled.

Previous studies found that novice investigators could obtain nearly as good or excellent test-retest reliability as experts if they are properly trained,^{260, 264} although one of these studies may overestimate reliability since investigators removed images that exceeded 'reasonable' variability.²⁶⁰ Johnson obtained ICCs > 0.9 for a novice investigator who only received six sessions of two hours over three weeks,²⁶⁴ suggesting that intensive training received over a short period of time could be enough. Besides, novice investigators could enhance their reliability by recording a 'cine-loop' (i.e. a short video sequence of scanned images) instead of a still image.²⁶⁴ Building on this evidence, in *Chapter 4* we recorded 3-5 s cine-loops that were later used during measurements, to ensure proper identification of muscle borders despite relatively low image resolution. Using low-end ultrasound devices, two studies reported good to excellent test-retest reliability for

IFM sizes.^{260, 268} Future studies still need to confirm inter-rater reliability of IFM sizes using low-end devices.

Despite these limitations, our results and those reported in previous pieces of research suggest that ultrasound imaging reliably capture the sizes of certain IFMs, especially in test-retest settings. Investigators, and especially less experienced operators, should be informed that measurement reliability relies on appropriate training and consistency throughout the scanning process, including the recording of a cine-loop.

Clinicians who have access to ultrasound imaging could benefit from specialized training courses to confidently monitor morphological changes during the treatment of ankle and foot musculoskeletal disorders. This approach could complement symptom-based and performance-based outcome measures.

Ultrasound-based outcomes for ankle/foot research

Beyond muscle size measurements, ultrasound imaging can provide a range of other outcome measures relevant to ankle and foot research.

Foot muscle width can be measured with good to excellent test-retest reliability, in addition to cross-sectional area and thickness, for a comprehensive assessment of macroscopic muscle morphology.²⁵⁷ Advanced techniques allow to characterize muscle volume by performing 2D-ultrasound imaging at regular intervals throughout the muscle long axis.³⁶⁴⁻³⁶⁶ Such ‘3D ultrasound imaging’ has already been performed at the IFMs with excellent reproducibility (ICC > 0.98).³⁶⁷ Interestingly, combining 2D ultrasound imaging with tracking of markers attached to the probe facilitates muscle volume calculation (i.e. ‘freehand 3D ultrasound imaging’).³⁶⁸⁻³⁶⁹ Muscle sizes can also reliably be obtained during active contractions to provide a more functional measure than purely static measures.^{65,}

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While the above-mentioned measures capture macroscopic muscle morphology, ultrasound also enables quantification of intramuscular features. Notably, pennation angle represents the angle between muscle fibre orientation and the longitudinal orientation of the tendon or aponeurosis. The force generated in the tendon is negatively influenced by that pennation angle, because only the component of the force vector parallel to the line of pull contributes to muscle power. Pennation angle can be measured reliably in the extrinsic foot muscles,^{257, 259, 370} but reliability for the IFMs is unknown.

Pennation angle has often been used to model muscle action at a joint. For example, physiological cross-sectional area – which uses pennation angle (θ) in its formula (Equation 9) – is the ‘theoretical area that would be occupied by a cylinder with a length equal to that of [muscle] fibres’.³⁷¹

Equation 9: Physiological cross-sectional area (cm²) =

$$\frac{\text{Muscle mass}(g) \times \cos(\theta)}{\text{Muscle density}(g/cm^3) \times \text{fibre length}(cm)}$$

However, Lieber (2022) recommends to abandon pennation angle because it does not meaningfully reduce tendon force, it omits fibre rotation during contraction, and it primarily results from a packing strategy of muscle fibres rather than presenting any functional significance.³⁷¹ In addition, Ledoux showed that the IFMs have limited pennation angles ($< 20^\circ$), resulting from their ability to generate force in the absence of important joint excursion (i.e. IFMs do not need to shorten in a great extent compared to knee or elbow flexors). This limited pennation angle means that $\cos(\theta)$ will be close to 1 and will not significantly affect physiological CSA.^{185, 372} Further, some IFMs include plural muscle bellies and others are bipennate, complicating the calculation of their pennation angle.

Muscle quality can be roughly estimated through echogenicity, which is the ability of tissues to return ultrasound waves. In simple terms, echogenicity is highest in liquids and lowest in air. Modified-Heckmatt scale provides a qualitative visual grading from 1 to 4, reflecting structural changes such as fatty infiltration and/or fibrosis.³⁷³⁻³⁷⁵ Among individuals with CAI, worse modified-Heckmatt score was associated with lower eversion strength and fibularis muscle cross-sectional area.³⁷⁶ However, magnetic resonance imaging provides more detailed and reliable information on tissue composition.¹³

Ultrasound imaging may also be useful to quantify soft-tissue stiffness, using shear-wave ultrasonography. This technique measures the speed of propagation of a focused acoustic pulse produced by the transducer to estimate the shear modulus (G , Equation 10) and Young's modulus (E , Equation 11) of the tissue, knowing the density of the selected tissue.³⁷⁷

Equation 10: $G \approx \rho C^2$; with ρ being density and C being shear wave speed.

Equation 11: $E \approx 3 \times G$.

In Equation 10 and Equation 11, E and G are usually expressed in kPa, ρ is expressed in $\text{g}\cdot\text{cm}^{-3}$, and C is expressed in $\text{m}\cdot\text{s}^{-1}$. Density ρ can be estimated based on published values, or approximated to be close to $1 \text{ g}\cdot\text{cm}^{-3}$ (i.e. density of water). Equation 11 is an approximation because it is only true for materials that have the same physical properties in all directions (i.e. isotropic materials with linear elasticity), while muscles are anisotropic and viscoelastic.³⁷⁸

V. Foot muscle function

Evolution of paradigms: Morphology and versatility

The mobile adaptor–rigid lever paradigm emerged in the early 1900s to explain foot function during locomotion. It proposes that the foot behaves as a compliant structure during early stance to accommodate the ground and attenuate impact forces, and transitions toward a functionally stiffer configuration during late stance to facilitate propulsion.⁹ Within this framework, changes in joint orientation—particularly at the subtalar and midfoot joints—are thought to enhance the effective transmission of ankle plantarflexor moments.⁹ Muscles intrinsic and extrinsic to the foot have been proposed to contribute to this functional stability and force transmission,¹⁰ although the paradigm did not originally specify the underlying neuromuscular mechanisms.⁹ We were guided by this paradigm in the conceptualization of the current PhD project, particularly when formulating hypotheses of altered ground reaction forces during running as a result of intrinsic foot muscle fatigue (Chapter 2).

Behling et al. critically reviewed the scientific validity of the ‘mobile adaptor-rigid lever paradigm’, concluding that many of its central assumptions have not been directly validated *in vivo* under dynamic conditions. More particularly, the notion that the foot stiffens or becomes quasi-rigid prior to or during propulsion needs to be confirmed. Based on contemporary evidence, they argue that foot function is better described as mechanically versatile, with substantial joint mobility persisting throughout stance and behaviour adaptively controlled by the central nervous system rather than by passive structural stiffening.⁹ Supporting this perspective, recent *in vivo* work demonstrated that medial longitudinal arch mobility plays an important role during push-off in running, with arch recoil occurring predominantly at the cuneonavicular joint. Rather than primarily serving to propel the body forward through elastic energy return, this recoil contributes to reorientation of the talus relative to the tibia, helping maintain an upright limb posture and influencing the timing of ground contact during propulsion.³⁶²

Other studies have shown that commonly used static morphological measures (e.g. medial longitudinal arch height) only weakly predict foot function during dynamic tasks, while more complex assessments of foot morphology may better perform.^{8, 379-381} Using a combination of *in vivo* joint mechanics during walking and

running and static three-dimensional foot scans in 100 adults, Schuster et al. (2024) applied principal component analysis to relate foot morphology to foot mechanics. They found that combinations of medial longitudinal and transverse arch posture, relative foot proportions, and toe shape explained variance in foot mechanics with moderate to excellent predictive strength. These findings support the view that foot function emerges from the integrated contribution of multiple morphological characteristics rather than from any single structural determinant.⁸

Contribution of the intrinsic foot muscles

Studies examining lower-limb biomechanics following IFM-fatigue protocols can provide insights into IFM roles. A single study observed walking biomechanics under IFM-fatigue,¹⁶² and none involved running or other sporting tasks. To address this shortcoming, *Chapter 2* details the changes in running ground reaction forces following concentric and isometric protocols designed to produce IFM fatigue among healthy participants. We observed a 4-5% increase in leg stiffness, and a 2-3% decrease in propulsive and vertical impulses. In *Chapter 5*, individuals with CAI performed anticipated side-step cutting before and after an isometric intervention targeting the IFMs. Participants exhibited a more dorsiflexed Chopart joint, reduced Lisfranc sagittal plane mobility during the active peak phase, and increased hallux motion during the propulsive phase. Greater hallux mobility may indicate decreased forefoot stiffness in late stance, shifting the ground reaction force application proximally and shortening the lever arm to the ankle centre of rotation. This interpretation aligns with the role of IFMs as active stabilizers of the metatarsophalangeal joint.²³ Our results may suggest that moderate toe flexor fatigue – affecting both IFMs and extrinsic muscles – compromised their ability to stiffen the forefoot. According to the ‘ankle gear ratio’ (Figure 38), this would create less favourable conditions for the ankle plantar flexors to generate power.²³ Still, these results should be confirmed under higher cutting speeds to strengthen our conclusions.

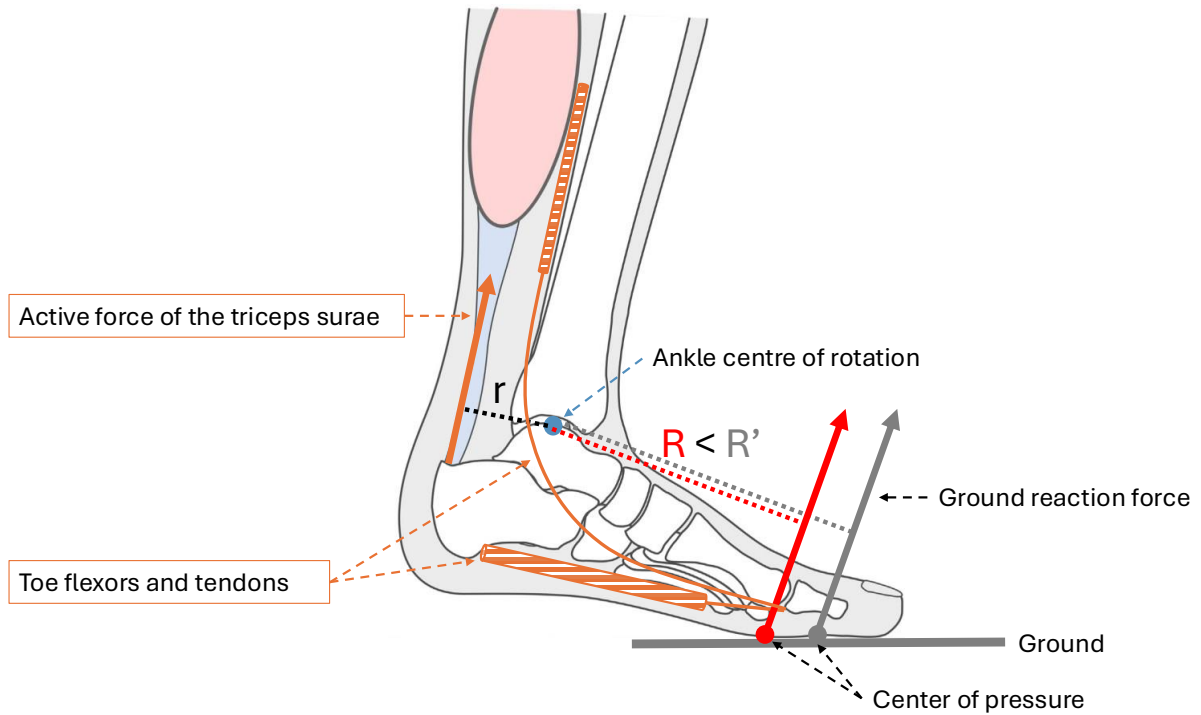


Figure 38. The ankle gear ratio is the ratio of the lever arm of the ground reaction force (R) and that of the ankle plantar flexors (r) at the ankle centre of rotation.³¹² Increased forefoot stiffness, for example by increased torque produced by the toe flexors (intrinsic and/or extrinsic muscles), results in a more distal point of application of the ground reaction force (grey vector), and thus a longer lever arm to the centre of rotation (R').^{19, 23} Figure adapted from Iaquinio (2023).¹⁰¹

IFM contributions to foot function have been investigated recently using nerve blocks, electromyography and simulation. Nerve blocks ensure that there is virtually no muscular contraction, allowing to discern muscle roles through the changes in joint dynamics. During jumping and landing, Smith et al. found that the IFMs contributed only a small fraction of total work output ($\approx 3\%$), but with large functional consequences: IFM activity stiffened the forefoot, shifted the centre of pressure forward and enhanced ankle leverage, enabling more effective energy generation and dissipation.¹⁵⁴ These findings were confirmed in a second study, where blocking IFM activity reduced metatarsophalangeal joint stiffness during jumping, shortened the ankle external moment arm, lowered ankle torque, and decreased jump height performance. IFMs thus actively optimize the lever function of the foot by shifting the centre of pressure forward, thereby enhancing ankle gearing and torque during propulsion.²³

Smith et al. (2022) combined motion capture, force plates, electromyography and ultrasound imaging to examine the flexor digitorum brevis fascicles and muscle-tendon unit during dynamic tasks. During propulsion, the tendon of this IFM amplified power output, and during landings it attenuated power input to protect muscle fascicles. These findings reveal that the IFMs contribute actively to both energy generation and absorption at the foot.²⁰ Williams et al. (2022) compared observed ankle/foot joint mechanics during walking to a prediction generated by a purely passive windlass model. They showed that the windlass mechanism accounts for a small portion of arch rise and distal foot coupling, with most effects arising from muscular activation and soft tissue contributions.³⁷⁹ Further, the activity of the IFMs was found to be tuned to surface stiffness and movement requirements, minimizing muscular effort.³⁸²

These findings position IFMs as key modulators of foot stiffness, arch mobility, and push-off efficiency, adapting their role to task demands rather than serving a single fixed function. This highlights the importance of considering IFM function in understanding the mechanisms underlying CAI.

Altered foot muscle function in chronic ankle instability

Altered extrinsic foot muscle function is a common feature of CAI:³⁹ individuals with CAI have been found to have smaller fibularis longus and brevis muscles,²⁷³ altered fibularis muscle quality,⁶⁴ and lower ankle strength in multiple directions.^{64, 78-79, 383} During single leg landing, fibularis longus latency was longer and its activity was lower in CAI compared to controls.^{341, 384}

Despite the functional importance of the IFMs, only a couple of studies had examined their sizes and strength in CAI at the outset of this project, yielding conflicting results.^{62, 65} Therefore, we advocated to further explore these outcomes,³⁸⁵ revealing that IFM size variance was partly explained by body mass index and clinical status (CAI, copers or control), and that hallux flexion strength was 20% lower in CAI and copers than in healthy controls (*Chapter 4*).

Ten years after the first study that investigated IFM sizes in CAI,⁶² considerable uncertainty remains. Our findings (*Chapter 4*) are consistent with those published by Fraser et al. (2021),⁶⁵ suggesting that altered hallux and toe strength may exist in the absence of systematic muscle size alterations among individuals with CAI. Two studies published in 2023 – both including a limited number of CAI participants – found no differences in the sizes of abductor hallucis, flexor hallucis brevis, flexor digitorum brevis, abductor digitorum brevis and quadratus plantae between CAI, copers and healthy controls. They reported (non-significant) trends of higher hallux and toe flexion strength in CAI, with effect sizes ranging from $d = 0.6-0.9$.^{151, 386} However, these two studies did not report perceived instability and self-reported function, complicating comparison and interpretation. Very recently, Jaffri et al. found large deficits in abductor hallucis ($d = 1.5$) and flexor digitorum brevis ($d = 1.4$) cross-sectional areas in CAI compared to controls, along with higher echogenicity values of both muscles, in a sample characterized by important perceived instability and self-reported disability.⁶⁶

These inconsistencies may reflect methodological differences in data acquisition and normalization: ultrasound imaging of the IFMs can be performed in non-weight bearing or standing positions, and measurements can be normalized to body mass. Overall, it seems that CAI is associated with lower IFM sizes normalized to body mass when captured in a weight bearing standing position,⁶⁶ but no difference was seen when images were captured unloaded.^{65, 151, 386} Similarly, toe strength is likely to be lower in CAI than in controls when accounting for body mass

and the length of the lever arm,⁶⁵ but not consistently when strength is not normalized.¹⁵¹ Besides, few reference values exist for IFM sizes and foot strength among the general population, providing opportunities for future research.

Altogether, the evidence points to the presence of IFM impairments in CAI, that are likely influenced by personal factors such as age and body composition. For those who present with impaired IFM function, this could favour other impairments related to CAI, such as dynamic balance,¹⁶⁴ the ability of the IFMs to absorb and generate power during dynamic tasks,²² and possibly somatosensation.^{27, 387} Impaired IFM function could thus indirectly affect the outcome on the continuum between CAI and coper. Besides, clinicians should be aware that the detection of these deficits seem to depend on measurement methodology, and future studies should validate standardisation procedures and publish normative values to clarify their roles and to inform targeted rehabilitation strategies.

Foot muscle assessment and strengthening as part of rehabilitation for ankle instability

Current guidelines for the treatment of ankle sprain and CAI recommend tailoring rehabilitation based on the specific deficits of each individual.^{47, 94-95, 300} Some individuals with CAI are likely to present with impairments in IFM function, which offers an opportunity to integrate the IFMs in the ‘rehabilitation-oriented assessment’.^{310, 388} This assessment could include strength, endurance, and selective motor control.

Clinicians with access to a hand-held press dynamometer can reliably evaluate maximal isometric flexion strength of the hallux and toe flexors with the patient in long sitting position and ankle in plantarflexion.¹³⁹ Alternatively, they may build a set-up to record doming strength, as proposed by Ridge et al. (2017).¹⁴⁰ Those equipped with a pull dynamometer can assess maximal toe-flexor strength using the enhanced-paper grip test. In this test, the patient or athlete sits with the hallux placed on a plastic card attached to the dynamometer and presses downward with maximal flexion effort while the clinician applies a horizontal pull. The peak force needed to remove the card provides an objective estimate of toe-flexor strength.³⁸⁹ Kekliceck et al. (2024) adapted this protocol to assess IFM endurance by recording the time the participant could hold the card against a pre-defined pulling force,¹⁶⁰

although validity and reliability remain unknown. Finally, the ability to achieve specific toe positions has also been suggested as an indicator of motor control and performance.¹³⁹

In our studies, maximal isometric hallux and toe flexion strength was assessed with participants in the long sitting position and the ankle plantarflexed. This position theoretically increases the contribution of the IFMs relative to the extrinsic foot muscles.¹³⁹ Alternatively, ‘forefoot strength’ can be considered as the result of the complementary actions of the IFMs and extrinsic foot muscles, rather than aiming to study them in isolation. For instance, Roma et al. validated the use of a fixated dynamometer to record metatarsophalangeal joint flexion strength with these joints at 30° dorsiflexion.³⁹⁰

In *chapter 1*, we showed that specific exercises and the use of minimalist footwear could improve foot strength and the sizes of some IFMs.²⁵¹ Our systematic review included any kind of participants, but we identified no randomized controlled trial assessing our outcomes among individuals with CAI. There is growing literature surrounding the use of IFM exercises in that population.^{310, 391-392} For example, in a pre-post study design, adding several foot muscle exercises as part of a four-week comprehensive rehabilitation lead to increased abductor hallucis and flexor digitorum brevis muscle cross-sectional areas, along with improved self-reported function and kinesiophobia.³¹⁰ This evidence suggests that interventions aiming to increase foot strength could be included in the rehabilitation of people with CAI.

The interventions implemented in the randomized controlled trials included in our systematic review were highly analytical (e.g. the short foot exercise) and delivered at low-dose, characterized by few sets, few repetitions and minimal resistance.²⁵¹ In contrast, a recent randomized controlled trial by Tourillon et al. showed that two months of more functional, high-load exercises produced substantial improvements in strength among highly trained individuals.³⁹³ These findings suggest that athletes with CAI may benefit more from high-load foot strengthening exercises than from isolated IFM exercises to enhance foot strength.

Furthermore, a paradigm shift is probably needed in the field of foot strengthening: while analytical IFM exercises can help patients or athletes develop awareness of IFM activation in early rehabilitation,¹⁶⁶ clinicians should apply fundamental conditioning principles by progressively increasing challenge, load, and dosage in prescribed interventions.³⁹³ Large randomized controlled trials should be

conducted among individuals with CAI, and pragmatic clinical trials could better tailor the content of rehabilitation to observed deficits in balance, ankle and foot strength, ranges of movement, and functional movements.

Minimalist footwear could increase foot strength,²⁵¹ but no evidence directly supports its use among individuals with CAI. Future studies could assess the safety of using such shoes during daily life or running among individuals who sustained ankle sprains, in addition with effects on impairments related to CAI.

The foot is also rich in sensory receptors that contribute to maintaining postural control.²⁷ Cutaneous sensation may be impaired among individuals with CAI,⁷⁶ providing support to interventions that involve sensorial aspects. Notably, McKeon and Wikstrom introduced the ‘sensory-targeted ankle rehabilitation strategies’, which combine ankle joint mobilization, plantar massage and triceps surae stretching.³⁹⁴ These interventions improved single-leg balance in a randomized controlled trial,³⁹⁵ and a systematic review reported that plantar massage produced small-to-moderate improvements in static balance.³⁹⁶ Other sensory-targeted rehabilitation strategies include whole-body vibration and exercising on textured surface,³⁹⁶ although further research is needed to establish their efficiency.

VI. Neuromuscular fatigue and electrical stimulation

Terminology

Chapters 2 and 5 implemented interventions aiming to affect toe flexor strength. There is discussion surrounding the choice of ‘fatigue’ versus ‘fatigability’ to describe a decline in force resulting from muscular activity.^{157, 397-399} Depending on the context, ‘fatigue’ could also refer to a self-reported, disabling symptom characterized by tiredness, lack of energy or exhaustion that limits physical and/or cognitive function.³⁹⁸⁻⁴⁰⁰ ‘Fatigability’ is defined as the rate and magnitude of change in performance capacity and fatigue-related sensations that occur in response to a defined physical or cognitive task, with a distinction between performance fatigability (e.g. decline in strength) and perceived fatigability (e.g. self-reported change in effort).^{398, 401}

To our knowledge, all the articles related to interventions aiming to acutely decrease hallux or toe flexion strength referred to ‘fatigue’.¹⁵⁸⁻¹⁶³ More generally, the vast majority of the literature related to muscles of the thigh, leg and foot preferred ‘fatigue’, even when reviewing recently published literature. PubMed search results confirm that ‘fatigue’ has been far more commonly used than ‘fatigability’, even in recent years (Figure 39). Therefore, we believe that using ‘fatigue’ instead of ‘fatigability’ would help readers find our articles on databases, and that a comprehensive definition of the selected term ensures research reproducibility.

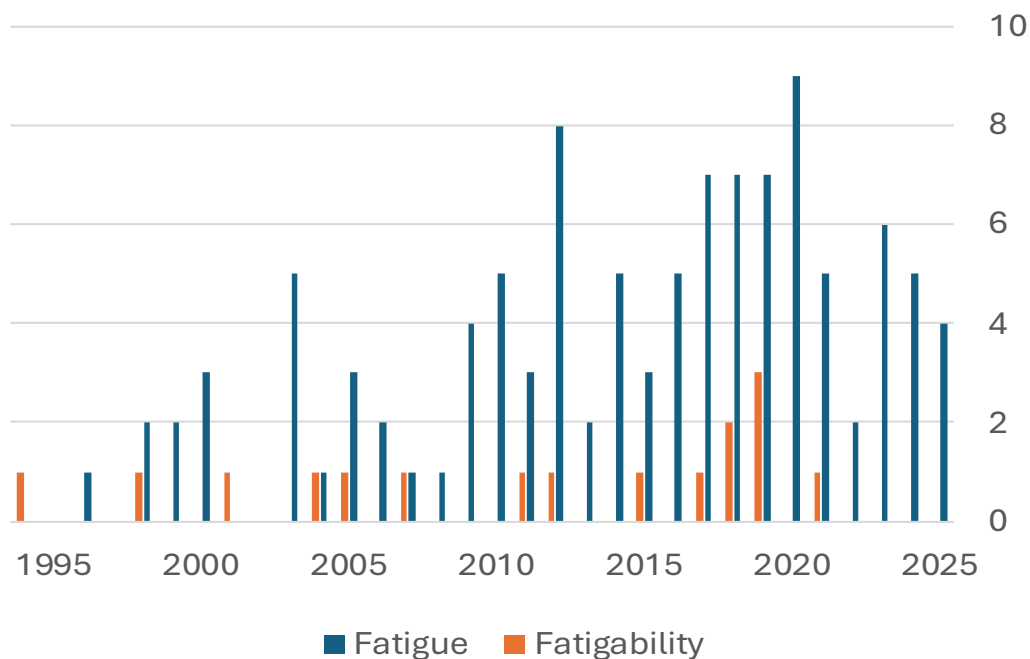


Figure 39. Evolution of the number of hits on PubMed using ‘Fatigue’ (blue) versus ‘Fatigability’ (orange), for a search related to the lower limbs. The search queries were “(fatigue[Title]) AND (flexor[Title] OR flexion[Title] OR extension[Title] OR extensor[Title]) AND (hip[Title] OR knee[Title] OR ankle[Title])” and its strict equivalent with ‘fatigability’.

Isometric versus concentric toe flexor contractions

At the outset of this PhD project, our goal was to investigate the effects of toe flexor fatigue on side-step cutting biomechanics among individuals with CAI (*Chapter 5*). Midway through recruitment, preliminary analyses revealed minimal fatigue following the isometric protocol. Around the same time, data collection was interrupted due to equipment failure at CMAL (UZ Leuven Pellenberg). This unexpected pause provided an opportunity to initiate a comparative study at the NMSK laboratory (Brussels), testing our isometric protocol against a concentric one, among healthy participants (*Chapter 2*). Despite this chronology, *Chapter 2* was placed in the ‘methodological’ part and *Chapter 5* in the ‘clinical’ part of this thesis. This also explains why *Chapter 5* employed the isometric protocol, whereas *Chapter 2* suggested that the concentric protocol would be more effective in inducing fatigue.

Several previous studies aimed to induce toe flexor fatigue, but achieved limited effectiveness.¹⁵⁸⁻¹⁶³ We justified our isometric intervention in the introduction of

Chapter 2. Briefly, the isometric modality was selected because a prior study showed that sustained hallux flexion and abduction at 40% of maximal voluntary contraction resulted in a 25% strength reduction.¹⁵⁹ The 16-minute duration was chosen to address the short intervention times used in earlier studies,¹⁵⁹⁻¹⁶⁰ while the loaded position differed from the seated posture commonly employed previously. We expected that sustained isometric contractions would not be possible for such an extended period,¹⁵⁹⁻¹⁶⁰ and therefore implemented intermittent contractions. Electrical stimulation was concurrently applied to the medial longitudinal arch to facilitate foot muscle activation,¹⁶⁶⁻¹⁶⁷ along with muscle fatigue being observed following such electrical stimulation.^{158, 168}

After having observed little fatigue following the isometric protocol (*Chapter 5*), we aimed to compare that protocol with an ‘enhanced’ protocol, matched for duration, standing position, and loading (*Chapter 2*). Concentric contractions appeared promising based on a previous article reporting fatigue after 4 sets of 75 concentric hallux flexions.¹⁶² This approach – multiple sets of 75 contractions performed at a rate of 40 repetitions per minute – has been commonly adopted in foot muscle fatigue studies.¹⁶¹⁻¹⁶³ Moreover, concentric contractions may be easier to execute than isometric ones, because the former involve visible movements. Therefore, participants performed eight sets of 75 concentric contractions at a rate of 40 repetitions per minute while standing and wearing a weighted vest (*Chapter 2*).

Results from *Chapter 2* showed that the concentric protocol induced a 20% reduction in strength, whereas the isometric contractions did not produce significant fatigue. These findings suggest that repeated concentric contractions performed in standing positions can effectively elicit fatigue; however the current protocol requires optimization to achieve greater strength declines. To this end, maximal contractions should be encouraged and supported by biofeedback and standardized verbal instructions. Biofeedback may rely on pressure sensors, force measurements, or electromyography, thereby allowing investigators to simultaneously monitor performance decrements and muscle recruitment over time. Furthermore, intervention duration could be individualized to each participant, performing the intervention to failure or to a predetermined strength decline. Finally, fatigue recall may be necessary if data acquisition extends several minutes beyond the intervention.¹⁶⁸

Electrical stimulation as part of fatigue protocols

We combined isometric toe flexor contractions with electrical stimulation applied to the medial longitudinal arch through a biphasic symmetric rectangular current. The lack of fatigue observed following this intervention (*Chapter 2* and *Chapter 5*) was likely strongly influenced by the low current intensities tolerated by participants. This limitation appears to stem from the instruction to experience significant discomfort without pain. While higher intensities would presumably induce greater fatigue, the local ethics committee approved only a non-painful intervention. These findings highlight the importance of considering higher, maximally tolerated stimulation intensities in future study designs, subject to appropriate ethical approval. In practice, stimulation intensity may be progressively increased towards each participant's maximal tolerance, while ensuring that participants retain the right to reduce the intensity or withdraw from the study at any time.

In *Chapter 2* and *5*, the electrical stimulation parameters were 30 Hz frequency, 400 μ s pulse width, 8 s 'On' / 4 s 'Off' duty cycle, 1.5 s rise time, and 0.5 s fall time (Figure 40). We selected these precise characteristics because a previous study reported neuromuscular fatigue of the shoulder flexors using the same current.¹⁶⁸ In addition to intensity, these settings likely influenced fatigue output, and may thus also have influenced subsequent biomechanics during side-step cutting.

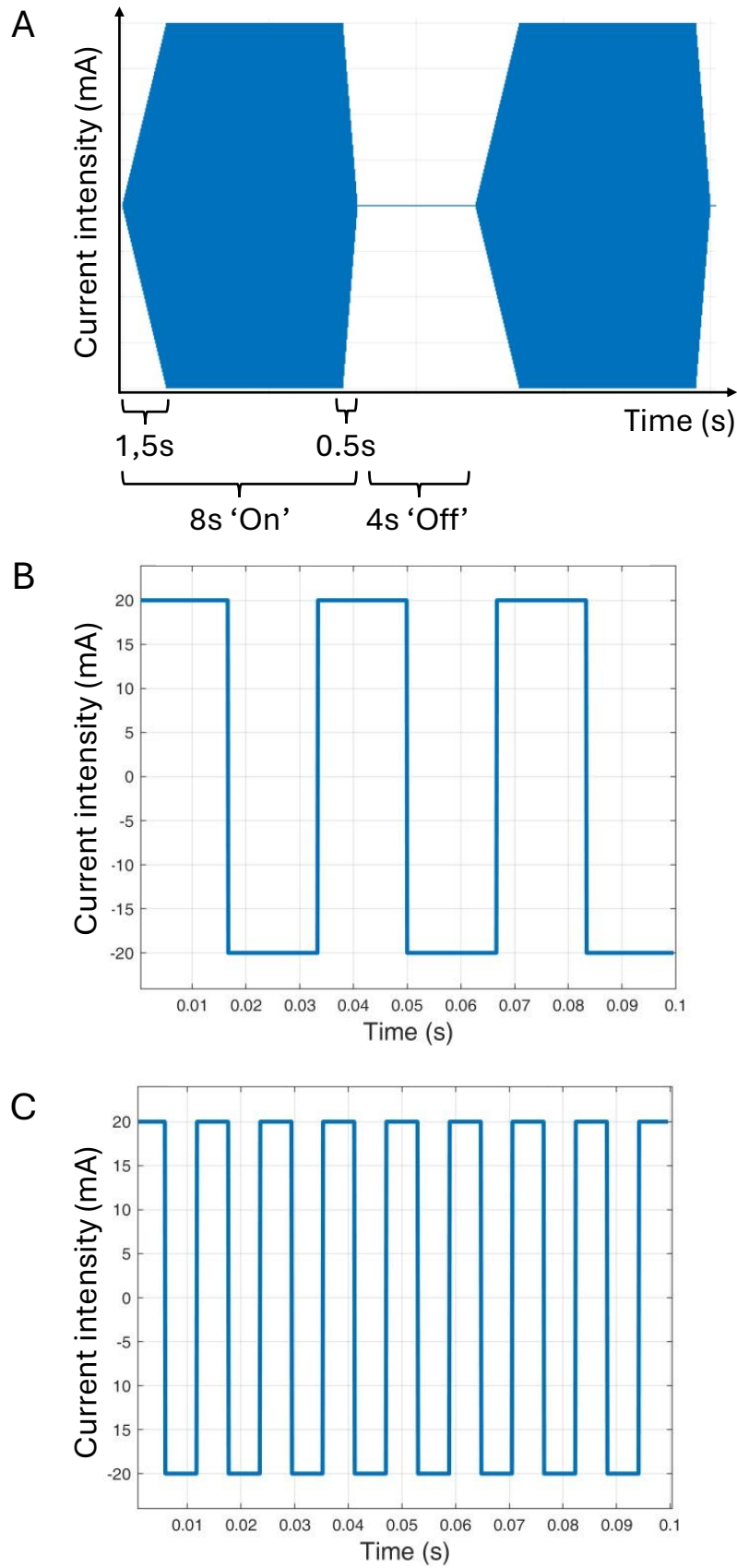


Figure 40. Graphical representations of rectangular biphasic currents. Graph (A) depicts the 'On' and 'Off' periods, with 'On' being composed of a rise ramp, plateau and fall ramp. The two lower graphs represent arbitrary currents of 20 mA at frequencies of 30 Hz (B) and 85 Hz (C) over a period of 0.1 s.

Frequency is the number of cycles per second (Figure 40, parts B and C). Rehabilitation protocols commonly employ values around 85 Hz,^{211, 225, 245} but such settings induce modulations at spinal and supraspinal levels in addition to expected peripheral effects. By contrast, lower frequencies mainly affect muscular contractility with less effects on the central nervous system.¹⁶⁹ This motivated our selection of 30 Hz, together with evidence of fatigue following interventions using rates of 20 and 30 Hz.^{168, 402} However, higher values (80-100 Hz) unequivocally induced higher levels of fatigue in other studies.^{169, 403-405} Moreover, a 30-Hz stimulation rate is likely less tolerable than higher rates at the same intensity, as participants perceive each individual twitch more distinctly. Taken together, these observations suggest that higher frequencies could have induced greater fatigue, driven by alterations occurring at muscular, spinal and supraspinal levels.

Impulse width is the duration of each individual electrical strain, typically lasting 400 μ s in rehabilitation,^{211, 225, 245} How impulse width impacts fatigue is not clear.²⁵² In theory, longer pulse widths generate deeper stimulation into the muscles and more small-diameter motor axons are stimulated, but research showed that at 20 or 80 Hz, increasing impulse width did not modify fatigue.⁴⁰⁵ Wide-pulse (1000 μ s), high-frequency (100 Hz), low-intensity electrical stimulation could also induce fatigue,¹⁵⁸ but a systematic review found no difference in maximal voluntary contraction strength and perceived fatigue compared to conventional electrical stimulation (20 to 50 Hz).⁴⁰⁶

Duty cycle is the time under contraction relative to rest time.²⁴⁸ A continuous stimulation – characterized by the absence of rest – results in high levels of fatigue but also a high risk of muscle cramp.²⁵² We thus used intermittent electrical stimulation with longer contraction than recovery periods (8s ‘On’ and 4s ‘Off’), to maximise fatigue while avoiding symptoms.

Neuromuscular electrical stimulation is also associated with inherent limitations. First, approximately 25% of people are ‘non-responders’, meaning the intervention will not affect outcomes as expected.^{245, 407} Second, electrical stimulation induces non-selective, synchronous motor unit recruitment that differs from the orderly and adaptable recruitment patterns of voluntary contractions, thereby affecting motor control through distinct neural mechanisms.⁴⁰⁷ Lastly, electrical stimulation preferentially activates superficial muscle regions close to the electrodes, leaving

deeper muscle portions relatively unstimulated.⁴⁰⁷ These limitations must be considered when interpreting the observed biomechanical effects in our studies.

To conclude, electrical stimulation demonstrates considerable potential as component of fatigue protocols, although further studies are needed to determine optimal current characteristics. Future research should carefully define and standardize key variables, including frequency, impulse width, duty cycle and criteria for determining adequate intensity.

VII. Building on the ‘updated model of chronic ankle instability’

The ‘updated model of chronic ankle instability’ proposed by Hertel and Corbett (2019) provides a comprehensive framework that integrates sensory-perceptual, pathomechanical, and motor-behavioural impairments to explain the persistent symptoms and functional limitations after a lateral ankle sprain.³⁹ By situating our findings within this model (Figure 41), we aim to better understand how the observed alterations in strength, movement, and self-reported function may interact with other factors contributing to the clinical presentation of CAI.

Our main sensory-perceptual inclusion criterion for CAI in clinical studies was perceived instability. Therefore, individuals with CAI reported important levels of perceived instability compared to copers and healthy controls, and our sample was also characterized by some alterations in self-reported function and trends for kinesiophobia in CAI compared to copers and controls. These sensory-perceptual impairments likely interacted with pathomechanical and motor-behavioural impairments, mediating lower-limb biomechanics during side-step cutting. During single-leg drop, it was found that perceived instability correlated with smaller ankle adduction and plantarflexion angles before landing and less ankle abduction and dorsiflexion after landing.³³⁵ Individuals with CAI generally rely more on visual information than healthy controls, and they have issues with double tasking.⁴⁰⁸⁻⁴⁰⁹ Therefore, we expect that lower-limb joint dynamic alterations would be more pronounced when participants are exposed to a ‘chaotic’ environment, compared to controlled laboratory settings.

Pathomechanical impairments – which relate to structure – also influence side-step cutting biomechanics. There is converging evidence that ankle joint mechanical laxity results in greater hindfoot external rotation during 45° side-step cutting.^{125, 127} Furthermore, reduced weight-bearing ankle dorsiflexion range of motion – a common osteokinematic restriction – may lead to reduced ankle dorsiflexion and inversion angles during jump landing/cutting.⁴¹⁰ Correlations also exist between the time required for cartilage to recover its thickness after loading and lower-limb joint dynamics and spatiotemporal gait parameters during walking.⁴¹¹

Among motor-behavioural factors, we found lower hallux flexion strength in CAI and copers compared to healthy controls, in addition to non-significant trends of alterations in the lesser toe flexion strength and dynamic balance. We did not record other factors that may influence movement patterns, such as muscle activation patterns, altered reflexes and neuromuscular inhibition. IFM function also likely influences ankle/foot biomechanics during sporting tasks, since these muscles attenuate foot impact, support metatarsophalangeal joint stiffness during running push-off,^{22, 153-154} and contribute to ankle joint moment during jumping.²³ Still, these pieces of evidence were obtained among healthy participants, not individuals with CAI. The very limited evidence specific to CAI points towards higher abductor hallucis activity in CAI than copers and controls during running,⁴¹² altered hallux and toe flexion strength,⁶⁵ and we found subtle changes in foot joint moments during late stance of side-step cutting following IFM fatigue in CAI (*Chapter 5*). Another study concomitantly found lower IFM sizes and increased echogenicity (i.e. lower tissue quality) in CAI compared to controls,⁶⁶ but more research is needed to investigate factors contributing to IFM deficits.

As a reminder, most evidence exposed above is related to correlations. Therefore, causality between variables should be confirmed by prospective studies. Caution is also warranted when considering external validity, as findings often apply to populations with specific characteristics, performing specific tasks, and with uncertainty regarding unmeasured factors (Figure 41). Future research should therefore aim for comprehensive, multi-factorial assessments in CAI-related research.

To conclude, clinicians should be guided by the ‘assess-treat-reassess’ principle.⁴¹³ Therefore, they should at least investigate the outcomes advised in recent guidelines, including balance and ankle strength.⁴⁷ Some individuals with CAI appear to exhibit altered IFM function, although assessment guidelines do not include foot strength nor foot motor control. We suggest that there are at least two circumstances when clinicians could implement interventions targeting the IFMs and/or the sensorial aspect of the foot (*cfr.* p. 212). First, if well-conducted physical therapy failed to relieve symptoms and restore function, alternative treatment strategies should be considered. Second, if clinical reasoning points towards deficits in IFM function – at any stage of rehabilitation – clinicians could include specific exercises targeting those deficits and monitor changes in clinically-relevant outcomes such as perceived instability, pain, and dynamic balance.

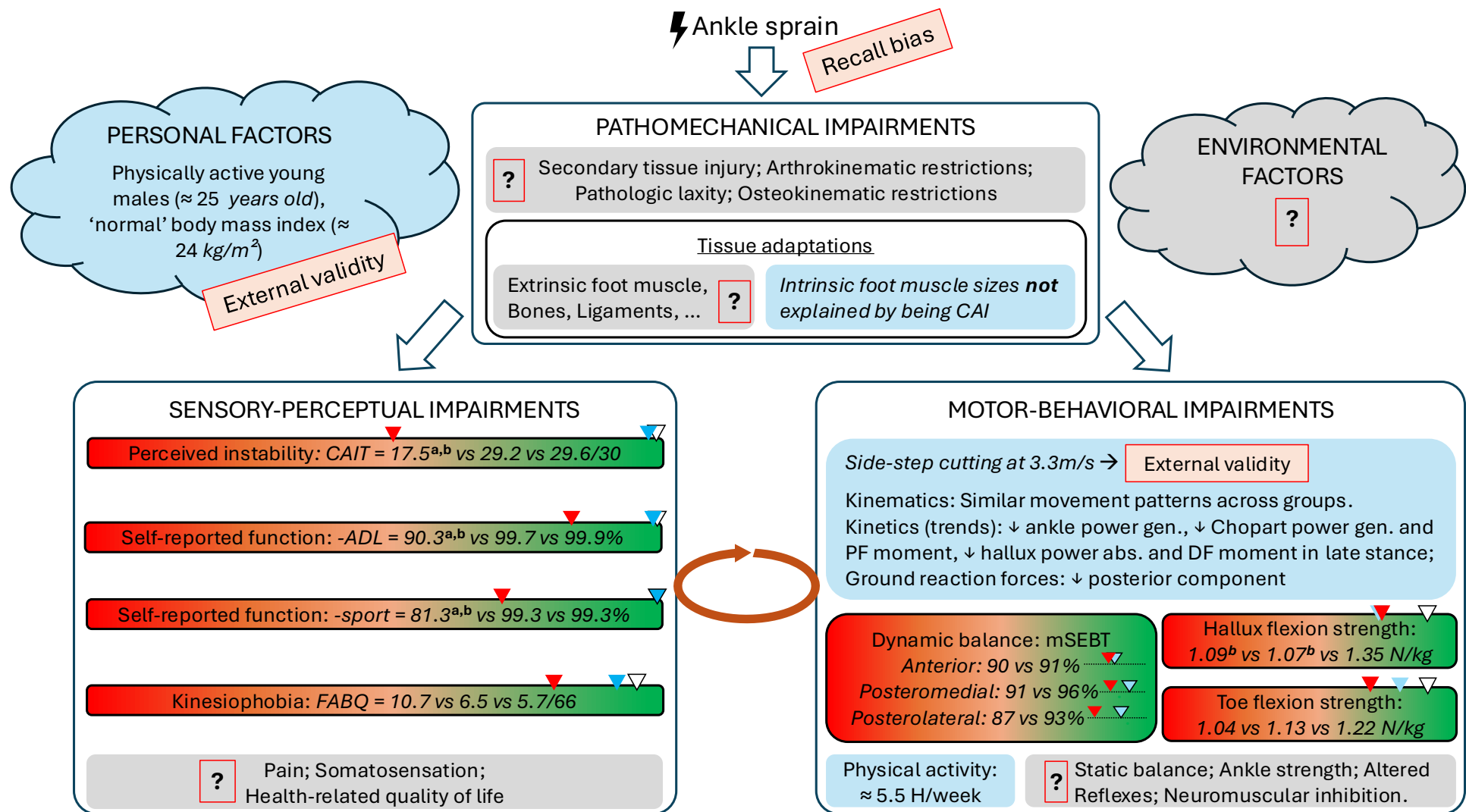


Figure 41. Contextualization of the results gathered in the clinical study (*Chapter 4*) of individuals with CAI (red triangles), copers (blue triangles) and healthy controls (white triangles). Abbreviations: Abs. (absorption), ADL (Foot and ankle ability measure – activities of daily life), CAI (chronic ankle instability), CAIT (Cumberland ankle instability tool), DF (dorsiflexion), FABQ (Fear avoidance belief questionnaire), gen. (generation), mSEBT (modified star excursion balance test), PF (plantarflexion), ? (not investigated in this study). Superscript letters indicate significant between-group differences at $p = 0.05$: ^a (different from copers), ^b (different from control). Figure adapted from Peters-Dickie et al. (2025) and Hertel and Corbett (2019).^{39, 414}

VIII. Limitations

The different studies conducted as part of this PhD project present limitations that have been acknowledged at the end of each chapter. Next section will consider these limitations more transversally.

Internal validity

Search strategy and outcome definitions

In our meta-analysis (*Chapter 1*), we had broad definitions of the interventions of interest. We combined different kinds of IFM exercises and different applications of minimalist footwear, although we kept exercises and minimalist footwear separated. Our results may also not capture differences in efficacy between varying dosages and durations.²⁵¹

Lower limb biomechanics during locomotion were part of the secondary outcomes of this systematic review, but search queries employed in the databases could have been optimized for this purpose. The ‘outcome’ part of the search strings only includes synonyms of ‘intrinsic foot muscles’. To ensure a comprehensive search while minimizing irrelevant results, we could have incorporated other search queries specific to lower limb biomechanics.

IFM function is an umbrella term encompassing several aspects. It may refer to the activation patterns of the IFMs – including amplitude and timing during locomotion and sporting activities. These considerations view the IFMs as part of the active components of the ankle and foot complex (*cfr.* p. 33). IFM function can also denote their capacity to generate maximal strength and sustain endurance. Because strength tests typically involve both IFMs and extrinsic foot muscles, muscle sizes are often used as surrogate measures (*cfr.* p. 52). However, changes in IFM strength or sizes do not necessarily translate into altered function during dynamic tasks, and conversely, abnormal activation patterns may occur without measurable strength or size deficits. Therefore, caution is warranted when interpreting strength as a surrogate of function. Future studies should investigate foot muscle activation using electromyography in individuals with CAI, and examine potential associations between IFM strength and symptoms.

Protocol limitations and measurement validity

In *Chapter 2*, randomization per block was performed without stratification for confounding factors such as sex, which explains why gender is not similarly distributed between sequence orders. Still, the other demographic characteristics were evenly distributed.

The main limitations of the fatigue protocols (*Chapters 2 and 5*) are their inability to produce important declines in hallux flexion strength and the uncertainty associated with the relative contribution of the IFMs and extrinsic foot muscles during interventions and strength assessments. Especially, *Chapter 5* included the isometric protocol, despite it was suggested to be ineffective in *Chapter 2*. Future studies should allow participants to perceive ‘tolerable pain’ during electrical stimulation, to elicit greater levels of fatigue.

The superiority of the concentric over the isometric contractions to produce foot muscle fatigue is difficult to defend since these interventions had different duty cycles and only the isometric protocol was combined with electrical stimulation. Besides, some individuals do not respond to electrical stimulation,^{245, 407} and the associated discomfort may reduce participants’ adherence to maximal contractions.

In the ultrasound reliability study, some muscles exhibited poor reliability, which can partly be attributed to the investigators’ limited experience. Although several days of study piloting were conducted to ensure consistent scanning procedures, additional training could have improved reliability. Beyond investigators’ experience, other contributing factors may include the use of a low-end device, the absence of skin landmarks, and the restricted time for image acquisition. A key limitation of this study is its inability to determine the extent to which these factors influenced the results, which would require controlled studies. Our results suggest that ultrasound imaging should be performed under the most ‘optimal’ conditions, including adequate investigator expertise and sufficient time for acquisition, while being aware of potential limitations of low-end devices.

Marker-based motion analysis is subject to skin motion artifact.⁴¹⁵ Still, alternatives have considerable drawbacks. First, the invasive nature of markers pinned to bones prevents large-scale implementation.⁴¹⁶ Second, markerless motion capture of foot joint kinematics has only been partly validated for walking gait.⁴¹⁷ Furthermore, data from knee and hip suggest that frontal and transverse plane dynamics obtained with markerless technology may not be valid.⁴¹⁸ Lastly, biplane fluoroscopy involves a limited imaging volume, radiation exposure, reduced accuracy during rapid movements, and complexity of data processing.³⁶³ Therefore, marker-based motion capture currently represents the most suitable and scalable technique for accurately quantifying multi-segment foot joint kinematics during dynamic tasks in large participant cohorts.

Sampling bias and recall bias

Our experimental studies may be subject to sampling bias, as participants were primarily recruited from the direct or indirect professional networks of the investigators. Consequently, the sample may exhibit demographic or socio-economic characteristics that differ from those of the broader population of interest.

In addition, data reported in *Chapters 4* and *5* were collected during the same acquisition session conducted at the CMAL laboratory (UZ Leuven, Pellenberg). These chapters are therefore also susceptible to recall bias, as participants were asked to retrospectively report their history of ankle sprains and other injuries.^{40, 45,}
¹²⁹ Any unmeasured sampling bias would thus be expected to similarly influence the findings of both chapters.

Order effect and multiplicity of outcomes

Data acquisition and interventions were performed in the same order for all participants rather than order being randomized (*Chapters 2, 4, and 5*). The sequence of procedures was decided prospectively to promote reliability of primary outcome measures, and was tested during study piloting. Order bias may occur if completing one procedure influences the results of subsequent procedures. Still, if order bias existed in our studies, the effect would be systematic because order was kept constant.

Further, investigating lower-limb biomechanics results in dozens of outcomes related to joint kinematics, joint kinetics and ground reaction forces. For each study, a few outcomes are defined as ‘primary outcome measures’, for which sample size was calculated in order to achieve sufficient statistical power for a pre-determined effect size. However, the study may be underpowered for some secondary outcome measures, and the important number of those secondary outcome measures result in a high risk of false positive, due to the nature of hypothesis testing.

Study design validity

Cross-sectional studies such as *Chapter 4* do not allow interpretation of findings in terms of causality.⁸⁹ Therefore, it is not possible to conclude from our studies that CAI is the cause or the consequence of the observed differences. *Chapter 5* had a pre-post design, without a control group, meaning that observed change cannot be attributed to the intervention with certainty.

External validity

Populations generalizability

We chose to recruit young recreationally active men in our clinical studies, because we intended to remove possible variation due to sedentary lifestyle and sex. Besides, the sample size calculation (*Chapter 4*) was done by using data from male participants. Women represent 40% of study participants in research related to ankle sprain,⁴¹⁹ and some studies related to lower-limb biomechanics during sporting tasks included only females.^{83, 420} Efforts should be made to continue exploring foot and lower limb joint dynamics among females. By contrast, middle-aged adults and elderly are underrepresented in CAI research, despite more frequently presenting ankle structural changes that could influence movement patterns and foot joint energy generation and dissipation.

According to Hertel and Corbett, researchers need to decide if self-reported function is part of their inclusion criteria for individuals with CAI, whether it is relevant for their research question to include only people with altered function.³⁹ We made the choice to recruit participants regardless their FAAM score. We believe that our results were influenced by this choice and that individuals with more perceived instability and more self-reported disability may present with more pronounced alterations in movement patterns during side-step cutting. We did not have enough participants to perform formal subgroup analyses with adequate statistical power.

Interventions generalizability

Current recommendations related to IFM strengthening are biased because most studies used interventions of limited volume and intensity, and short follow-up periods.²⁵¹ More demanding protocols may lead to more important changes.³⁹³ Similarly, transitions to minimalist footwear were often too rapid and not accompanied by injury-prevention exercises, and it is therefore difficult to interpret the high rates of reported injuries.²⁵¹

Ecological validity

Data acquisition with MSFMs is often performed barefoot because markers must be attached directly to the skin. This limits ecological validity for activities and sports that are usually performed with shoes.

In *Chapter 2*, running speeds were predefined at 9 and 12 km·h⁻¹, which represent varying efforts for the individuals relative to their physical abilities. In *Chapters 4-5*, side-step cutting was anticipated, influencing lower-limb biomechanics.^{318, 341} Precautions could be taken to limit anticipation, for example by randomly dictating running straight ahead or cutting, and possibly cutting direction,³⁴¹ but this would multiply the number of trials requested if data collection involves pressure or force plates.

IX. Perspectives

Improving and extending outcome measures

Advancing foot muscle assessment

We need to further investigate the contribution of the IFMs to foot function versatility under varying tasks and constraints. Electromyography will provide crucial information in both healthy participants and individuals with CAI.

Analytical measures of IFM function – such as strength and sizes – provide valuable insights, supporting clinical reasoning. Maximal isometric metatarsophalangeal joint flexion strength, with these joints positioned at 30° dorsiflexion as recommended by Roma et al. and Tourillon et al. ^{–232, 390} should be examined in individuals with CAI compared to copers and healthy controls. Aside from isometric strength, assessments could be enriched by incorporating endurance measures, concentric and excentric strength across the physiological range of motion, and electromyographic data. Furthermore, new tools and methodologies are needed to assess isokinetic hallux and toe flexion strength. To ensure consistency across studies, standardized protocols are essential, including clear instructions on whether patients are permitted to perform toe grasping (interphalangeal joint flexion) during testing.

Ultrasound imaging is a valid method to collect foot muscle sizes, with good to excellent test-retest, ^{148, 257, 260-261, 266-268} and moderate global inter-rater reliability. ^{262, 265} Researchers and clinicians aiming to perform ultrasound imaging should follow intensive and specific training, ²⁶⁰ record a cine-loop while scanning, ²⁶⁴ use both palpable and internal anatomical landmarks, and add labels to images during recording. These precautions will enable better location of muscle borders during measurement and ensure reliability. Future studies should compare the reliability obtained by novice versus experts using a low-end ultrasound device.

Weight-bearing ultrasound imaging provides the opportunity to capture muscle sizes in more functional situations, but it comes with additional challenges, ²⁶⁵⁻²⁶⁶ and reliability needs to be confirmed.

Lower-limb biomechanics during sporting tasks

As described earlier, the ‘proportionality scheme’ associated with foot joint kinetics requires validation during running and other sport-specific tasks, including multi-planar movements such as side-step cutting. Recently developed tools allow to capture ‘real’ medio-lateral and antero-posterior constraints at the level of individual foot segments.³⁵⁸

During side-step cutting, it would be particularly relevant to investigate lower-limb dynamics during the stance phase preceding the change of direction, as this may provide insights into motor control in the preparatory phase. Researchers should also aim to standardize and control step length and the actual angle of directional change, for example by using the position of the reflective markers attached to the feet.

Dynamic balance

The modified star excursion balance tests assesses dynamic balance by measuring the distance reached by the participant, while maintaining single-leg balance.⁴²¹ However, researchers could expand that scope of assessment by incorporating additional outcomes, particularly when advanced facilities such as motion analysis, force plates, and electromyography are available. These devices would provide valuable insights into joint kinematic coordination and motor control strategies,⁴²² execution speed, centre of pressure evolution, and muscle activation patterns.

Refining conceptual models of CAI

Research of the next couple of years should also seek to refine and validate the ‘updated model of CAI’.³⁹ Longitudinal cohort studies are therefore the most suitable to assess prognosis factors and interactions between factors. Alternatively, cross-sectional studies should continue to include a ‘coper’ group and researchers need to anticipate subgroup analyses in their sample size calculation. These stratified analyses should investigate the effects of specific impairments, such as the degree of self-reported function, on other impairments associated with CAI. For example, we explored in a Master thesis how K-means clustering could help classify individuals with CAI and copers based on foot joint ranges of motion during side-step cutting and dynamic balance. We identified two distinct groups, one ‘hyper-mobile’ and one ‘hypo-mobile’, but once more, sample size was probably too low to detect group-differences in other impairments (unpublished data).⁴²³ In another Master thesis, we used multiple linear regressions that aimed to explain the variance in perceived instability based on sensory-perceptual, motor-behavioural, and structural factors commonly affected in CAI (unpublished data).⁴²⁴ We conclude that K-means clustering and multiple linear regressions have promising applications to explore the multi-causal nature of CAI.

Some individuals with CAI likely present with altered foot muscle sizes and strength, but the question of impaired IFM function as a common feature associated with CAI remains open. In total, studies investigating IFM sizes have included 73 participants with CAI, 40 copers and 110 healthy controls. In addition to that limited pool of participants, these studies have come to contrasting conclusions. More research is thus needed, including innovative ways of evaluating foot strength, and symptom severity should be explored as a potential covariate.

Since IFMs contribute to foot function during dynamic tasks, future studies should explore the influence of IFM activation patterns on foot joint kinematics and kinetics in individuals with CAI.

Integrating the intrinsic foot muscles as part of rehabilitation

Ongoing randomized controlled trials are investigating the effects of IFM exercises as part of rehabilitation for CAI.³⁸⁸ We suggest that future trials implement higher-intensity, higher-volume interventions compared to those commonly used in the literature. On a clinical perspective, studies should investigate the effects of such interventions on the deficits and symptoms commonly associated with CAI, or the specific symptoms of each participant in ‘pragmatic’ clinical trials.

The safety of using minimalist footwear also needs to be considered among symptomatic individuals. Besides, we suggest that patients who suffered an ankle sprain perform a substantial part of their supervised rehabilitation barefoot, to encourage IFM activation and stimulate neural receptors of the foot sole.

Sport-specific tasks and neuromechanical adaptations

Side-step cutting remains a particularly relevant task for individuals with ankle sprains in sporting contexts, which is why more studies should investigate ankle and foot joint dynamics during such movements. We observed subtle foot joint kinetic differences between individuals with CAI and the other groups when the approach speed was moderate, despite pronounced sensory-perceptual alterations in those with CAI. Therefore, we suggest that future studies vary speed and cutting angle during unanticipated tasks.

During the acquisition session at CMAL (UZ Leuven Pellenberg), we also collected foot joint dynamics during running, but these data were not formally analysed yet. The scientific community would benefit from their publication, as it would provide the first characterization of foot joint kinetics during running in CAI, while complement existing kinematic data.^{84-86, 302} Another exciting and challenging type of analysis would be to perform 2D-SPM comparisons of plantar pressure data, that we collected during running and side-step cutting.⁴²⁵⁻⁴²⁶

Research should investigate movement patterns during varying sport-specific actions. Among others, braking (deceleration) abilities were not yet assessed in CAI despite their repeated use in sporting contexts. Besides, individuals with CAI make more errors than healthy controls while reacting to visual cues during balance task, likely because they have troubles for accurately perceiving their surroundings and making quick decisions.⁴²⁷ On-field situations involve rapid decision-making in a chaotic environment,³⁰⁷ and it is therefore urgent to integrate neurocognitive aspects in CAI-related research and rehabilitation. Examples include visual and auditory inputs, varying ground surfaces, and dual tasking.

X. Conclusion

This PhD contributes to advancing the understanding of chronic ankle instability (CAI) by examining foot joint kinematics and kinetics in conjunction with intrinsic foot muscle (IFM) functions. Taken together, the results suggest that CAI arises from a complex interplay of structural, sensory-perceptual, and motor-behavioural impairments, all of which are influenced by individual characteristics and task-specific demands.

Methodologically, this thesis supports the clinical applicability of ultrasound imaging to assess IFM sizes as a surrogate of foot muscle strength, with moderate to good test-retest reliability. The systematic review confirmed that IFM exercises and the use of minimalist shoes can increase IFM sizes, enhance foot strength, while also contributing to modifications in locomotion biomechanics. Concentric repeated toe flexor contractions were more effective than isometric ones to decrease hallux flexion strength. During anticipated side-step cutting, individuals with CAI demonstrated subtle kinetic differences at various foot joints and reduced posterior ground reaction forces compared to healthy controls and/or copers. IFM sizes were more influenced by body composition than by clinical status, yet hallux flexion strength was consistently reduced in CAI and copers compared to healthy controls, indicating possible neuromuscular deficits.

IFMs are active contributors to foot joint dynamics, modulating stiffness, mobility, and propulsion according to task demands. Their dysfunction may therefore exacerbate other impairments associated with CAI. However, current evidence inconsistently identifies IFM deficits, partly due to methodological variability and limited sample sizes. Standardized protocols and normative datasets are essential to clarify IFM roles and guide individualized rehabilitation. Integrating IFM assessment and strengthening into multimodal rehabilitation could enhance clinical outcomes, and research should consider cognitive aspects within motor-behavioural assessments.

Beyond the clinical scope, this thesis highlights the promise and challenges of multi-segment foot models for capturing complex foot biomechanics and calls for future developments that increase ecological validity and accessibility. It also emphasizes the need for longitudinal studies to refine the understanding of causal pathways and interactions leading to CAI.

In conclusion, this work contributes novel methodological and clinical insights into ankle-foot function in CAI. The foot is an adaptable sensorimotor system and should be integrated as such into the understanding and rehabilitation of CAI, alongside comprehensive consideration of structural, perceptive, and motor impairments.

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Appendices

Appendices of Chapter 1: Systematic review and meta-analysis

1.A Search equations used in the databases

PubMed search equation

("foot musc*" [Title/Abstract] OR "foot core" [Title/Abstract] OR "intrinsic musc*" [Title/Abstract] OR "intrinsic foot muscle" [Title/Abstract] OR "plantar musc*" [Title/Abstract] OR "abductor hallucis" [Title/Abstract] OR "flexor hallucis brevis" [Title/Abstract] OR "abductor digiti minimi" [Title/Abstract] OR "flexor digitorum brevis" [Title/Abstract] OR "quadratus plantae" [Title/Abstract] OR "lumbrical*" [Title/Abstract] OR "extensor digitorum brevis" [Title/Abstract]) AND ("resistance training" [MeSH Terms] OR "training" [Title/Abstract] OR "strengthening" [Title/Abstract] OR "retraining" [Title/Abstract] OR "exercise*" [Title/Abstract] OR "rehabilitation" [MeSH Terms] OR "rehabilitation" [Title/Abstract] OR "exercise therapy" [MeSH Terms] OR "exercise therapy" [Title/Abstract] OR "barefoot" [Title/Abstract] OR "minimalist" [Title/Abstract] OR "minimalistic" [Title/Abstract] OR "minimal shoes" [Title/Abstract] OR "minimal shoe" [Title/Abstract] OR "minimal footwear" [Title/Abstract] OR "minimal running shoe" [Title/Abstract])

Embase search equation

('foot muscle'/exp OR 'foot muscle':ti,ab,kw OR 'foot core':ti,ab OR 'intrinsic muscle'/exp OR 'intrinsic foot muscle':ti,ab OR 'plantaris muscle'/exp OR 'plantar muscle':ti,ab,kw OR 'plantaris muscle':ti,ab,kw OR 'abductor hallucis muscle'/exp OR 'abductor hallucis muscle':ti,ab,kw OR 'flexor hallucis brevis muscle'/exp OR 'flexor hallucis brevis muscles':ti,ab,kw OR 'flexor hallucis brevis muscle':ti,ab,kw OR 'abductor digiti minimi muscle'/exp OR 'flexor digitorum brevis muscle'/exp OR 'flexor digitorum brevis muscle':ti,ab,kw OR 'quadratus plantae muscle'/exp OR 'lumbrical muscle'/exp OR 'extensor digitorum brevis muscle'/exp OR 'extensor digitorum brevis muscle':ti,ab,kw OR 'musculus extensor digitorum brevis':ti,ab,kw) AND ('resistance training'/exp OR 'strengthening exercise'/exp OR 'retraining'/exp OR 'exercise'/exp OR 'effort':ti,ab,kw OR 'exercise':ti,ab,kw OR 'exertion':ti,ab,kw OR 'fitness workout':ti,ab,kw OR 'physical conditioning, human':ti,ab,kw OR 'physical effort':ti,ab,kw OR 'physical work-out':ti,ab,kw OR 'physical workout':ti,ab,kw OR 'training'/exp OR 'training':ti,ab,kw OR 'rehabilitation'/exp OR 'readaption':ti,ab,kw OR 'readjustment':ti,ab,kw OR 'rehabilitation':ti,ab,kw OR 'rehabilitative treatment':ti,ab,kw OR 'exercise therapy':ti,ab OR 'barefoot'/exp OR minimalist:ti,ab OR minimalistic:ti,ab OR 'minimal shoe':ti,ab OR 'minimal footwear':ti,ab OR 'minimal running shoe':ti,ab OR 'kinesiotherapy'/exp OR 'kinesiotherapy':ti,ab,kw OR 'kinesitherapy':ti,ab,kw)

Cochrane search equation

Search string 1: "foot muscle" OR "foot muscles" OR "foot core" OR "intrinsic muscle" OR "intrinsic muscles" OR "intrinsic foot muscle" OR "intrinsic foot muscles" OR "plantar muscle" OR "plantar muscles" OR "abductor hallucis" OR "flexor hallucis brevis" OR "abductor digiti minimi" OR "flexor digitorum brevis" OR "quadratus plantae" OR "lumbrical" OR "lumbricals" OR "extensor digitorum brevis"

Search string 2: "resistance training" OR "training" OR "strengthening" OR "retraining" OR "exercise" OR "exercises" OR "rehabilitation" OR "exercise therapy" OR "barefoot" OR "minimalist" OR "minimalistic" OR "minimal shoe" OR "minimal shoes" OR "minimal footwear" OR "minimal running shoe" OR "minimal running shoes"

Search string 3: "MeSH descriptor: [Resistance Training] explode all trees"

Search string 4: "MeSH descriptor: [Rehabilitation] explode all trees"

Search string 5: "MeSH descriptor: [Exercise Therapy] explode all trees"

Search string 6: Combine search strings 2, 3, 4 and 5 using Boolean operator "OR".

Final search equation used on Cochrane database: Combine search strings 1 and 6 using Boolean operator "AND".

Sportdiscus search equation

AB ("foot musc*" OR "foot core" OR "intrinsic musc*" OR "intrinsic foot muscle" OR "plantar musc*" OR "abductor hallucis" OR "flexor hallucis brevis" OR "abductor digiti minimi" OR "flexor digitorum brevis" OR "quadratus plantae" OR "lumbrical*" OR "extensor digitorum brevis") AND AB ("resistance training" OR "training" OR "strengthening" OR "retraining" OR "exercise*" OR "rehabilitation" OR "exercise therapy" OR "barefoot" OR "minimalist" OR "minimalistic" OR "minimal shoe*" OR "minimal footwear" OR "minimal running shoe*")

1.B Risk of bias of the included studies assessed with the Cochrane tool for randomized controlled trials (RoB2)

Author (year)	Randomization process	Deviations from intended interventions	Missing outcome data	Measurement of the outcome	Selection of the reported result	Overall risk of bias
Chen et al. (2016) ¹⁸³	⊕	⊖	!	⊕	⊕	⊖
Day et al. (2019) ²¹⁰	!	⊖	!	⊕	⊕	⊖
Ebrecht et al. (2018) ²¹¹	!	⊖	⊖	!	⊕	⊖
Elsayed et al. (2023) ¹⁹⁴	⊕	⊕	⊖	⊕	⊕	⊖
Goldmann et al. (2013) ²⁰²	!	⊖	⊖	⊕	⊕	⊖
Hartmann et al. (2019) ²⁰⁵	⊕	⊖	⊖	⊕	⊕	⊖
Johnson et al. (2016) ²¹²	⊕	⊖	!	⊕	⊕	⊖
Jung et al. (2011) ²¹³	⊕	⊕	⊕	!	⊕	!
Kim et al. (2015) ²²²	⊕	⊕	⊕	⊖	⊕	⊖
Kim et al. (2022) ²¹⁷	⊕	⊕	⊕	⊕	⊕	⊕
Mahmoud et al. (2022) ²¹⁸	⊕	!	!	⊖	⊕	⊖
Matias et al. (2022) ²¹⁶	⊕	⊖	⊖	⊕	⊕	⊖
McCarthy et al. (2013) ²⁰⁶	!	⊖	!	⊕	⊕	⊖
Mickle et al. (2016) ²¹⁹	⊕	⊖	⊖	⊕	⊕	⊖
Miller et al. (2014) ²¹⁴	!	⊖	⊖	⊕	⊕	⊖
Namsawang et al. (2019) ²²⁵	⊕	⊕	⊕	⊕	⊕	⊕
Namsawang et al. (2022) ²²¹	!	⊕	⊕	⊖	⊕	⊖
Okamura et al. (2020) ²²⁰	⊖	⊖	⊖	⊕	⊕	⊖
Okamura et al. (2020b) ²²⁶	!	⊕	⊕	⊕	!	!
Quinlan et al. (2022) ²¹⁵	⊕	⊖	!	⊖	⊕	⊖
Ridge et al. (2019) ¹⁸⁴	!	⊖	⊖	⊕	⊕	⊖
Shen et al. (2022) ²²⁷	⊕	⊖	⊖	⊕	⊕	⊖
Spink et al. (2011) ²⁰³	⊕	⊕	⊕	⊕	⊕	⊕
Suda et al. (2024) ¹⁹¹	⊕	⊖	⊖	⊕	⊕	⊖
Taddei et al. (2020) ²²³	⊕	⊕	⊕	⊕	⊕	⊕
Utsahachant et al. (2023) ¹⁹²	⊕	⊕	⊕	⊕	⊕	⊕
Warne et al. (2017) ²⁰⁴	⊕	!	⊖	⊕	⊕	⊖
Xiao et al. (2022) ²²⁴	!	⊕	⊕	⊕	⊕	!

Legend: ⊕ (low risk of bias), ! (some concerns of risk of bias), ⊖ (high risk of bias). For each article, the risk of bias was determined according to algorithms published with the Cochrane's revised tool spreadsheets. For the five domains (columns 2 to 6 of the table), this is done by answering so-called signalling questions. Overall risk of bias is at high risk of bias if at least one of the domains is scored as "high risk" or if more than 3 domains are scored as "some concerns". Overall risk of bias is at some concerns of risk of bias if one to three domains are scored as "some concerns".^{197, 428}

1.C Supervision, adherence, drop out, retention and adverse events reported in the studies

	Study	Supervision	Adherence: mean (SD)	Drop out (number per group)	Retention	Adverse events IG	Adverse events CG
MRS during running or daily life	Chen et al. (2016) ¹⁸³	Not supervised (deducted)	MRS: 39.2 (27) % Running: 28.0 (15.3) km in IG and 33.9 (22.5) km in CG (target ≥ 20 km/w)	9 drop out: 4 in IG and 5 in CG	81 %	0	0
	Johnson et al. (2016) ²¹²	Not supervised (deducted)	Estimated from graph: Total weekly mileage = 25.1 (5.95) km in IG and 27.0 (11) km in CG. Target = 24 to 48 km/w. Weekly mileage in MRS = 9.14 (9.9) km/w. 7 participants excluded (3 in IG) for lack of adherence	NA	84 %	8 bone marrow oedema (but symptomatic status not reported) Other injuries NA	1 bone marrow oedema
	Miller et al. (2014) ²¹⁴	Not supervised (deducted)	Weekly distance = 41.3 (13.8) km in IG and 42.7 (22.8) km in CG out of a target of 48.3 km/w.	NA	≥ 88 %	NA	NA
	Quinlan et al. (2022) ²¹⁵	Not supervised (deducted)	NA. Protocol: adherence assessed with surveys.	6 drop out: 4 in IG and 2 in CG	91 %	1 Sever's disease after 5 months	1 Sever's disease after 3 months
MRS + PIFM ex.	McCarthy et al. (2013) ²⁰⁶	Not supervised	NA. Similar between group difference.	11 drop out: 6 in IG and 5 in CG. Among the 11 drop out, 8 were injured (4 per group)	NA	≥ 4 injured: 1 or 2 metatarsal stress injuries 1 hip and calf pain 1 metatarsal pain 1 metatarsal stress fracture	≥ 4 injured
	Shen et al. (2022) ²²⁷	≥1d/w, unclear	Mean running distance per week > target distance Training sessions: 2.8 (0.4)/w (target = 3)	3 drop out: 2 in IG and 1 in CG	90 %	0	0
	Warne et al. (2017) ²⁰⁴	Not supervised (deducted)	IG: 87 % [56–96 %] of the injury prevention exercises and 96 % [88–100 %] of MRS CG: 92 % [84–98 %] of the injury prevention programme.	4 drop out: 2 in IG and 2 in CG. The 2 drop out in IG were injured.	86 %	2	NA
MRS or PIFM ex.	Ebrecht and Sichtung (2018) ²¹¹	Probably 100 %	NA	15 drop out: 8 in IG1 (all with overloading symptoms), 4 in IG2 and 3 in CG	80 %	MRS group: 8 with overloading symptoms	0

Exercises targeting the PIFM	Ridge et al. (2019) ¹⁸⁴	1/w	NA. Recorded by online form.	8 drop out in total. Number per group NA	88 %	NA 4 injuries unrelated to the study in total	
	Day and Michael (2019) ²¹⁰	Not supervised	NA. Training logs recorded.	3 drop out: 1 in IG and 2 in CG	87 %	3 injuries unrelated to the study in total	
	Elsayed et al. (2023) ¹⁹⁴	1 every 2w	PIFM ex.: 34.1 (4.8) days per participant (target = 42). Insoles: 83.2 % in IG and 82.2 % in CG	3 drop out (2 in IG)	93 %	NA Probably 0	NA Probably 0
	Goldmann et al. (2013) ²⁰²	100 % supervised	NA	NA	NA	NA	NA
	Hartmann et al. (2009) ²⁰⁵	2/w	NA	11 drop out: 4 in IG and 7 in CG	80 %	0	Rib fracture Pulled muscle
	Jung et al. (2011) ²¹³	1/w	NA	0 drop out	100 %	NA Probably 0	NA Probably 0
	Kim et al. (2022) ²¹⁷	100 % supervised	100 % (deducted)	0 drop out	100 %	0	0
	Kim et al. (2015) ²²²	1/w	NA	NA Probably 0 drop out	100 %	NA Probably 0	NA Probably 0
	Mahmoud (2022) ²¹⁸	≥ 2/w for SFE	NA	3 drop out + 1 excluded due to missed treatment sessions.	92 %	NA Probably 0	NA Probably 0
	Matias et al. (2022) ²¹⁶	IG: 1/w in person + 1/w remote; CG: 1/w remote	Weekly running mileage = 77.78 (57.31) km in IG and 89.09 (60.9) km in CG. Target = maintain pre-study (20 to 100 km/w) In-person sessions: 96.7 % Remote sessions: 83.5 % (at months 3 and 4) and 48.9 % (at months 11 and 12)	15 drop out: 8 in IG (among which 3 injured) and 7 in CG (among which 6 injured).	83 %	≥ 3	≥ 6
	Mickle et al. (2016) ²¹⁹	IG 100 % supervised CG: not supervised	IG: > 80 % of the participants attended at least 75 % of the exercise classes over the 12 weeks CG: 80 % completed > 75 % of the exercise sessions	17 drop out: 7 in IG and 10 in CG	80 %	1 back pain 1 “other”	2 foot pain 3 “other”
	Namsawang et al. (2019) ²²⁵	100 % supervised	100 % (deducted)	0	100 %	NA Probably 0	NA Probably 0

Namsawang et al. (2022) ²²¹	Probably 100 % supervised	100 %	0	100 %	0	0
Okamura et al. (2020) ²²⁰	100 % supervised	81 % in IG and 87 % in CG for PIFM ex.	6 drop out: 2 in IG and 4 in CG	74 %	NA Probably 0	NA 1 fall
Okamura et al. (2020b) ²²⁶	1/w	Home exercises: 24.5 (1.4) sessions per participant (target = 24) In-person sessions: 6.2 (1.0) sessions per participant (target = 8).	0	100 %	NA Probably 0	NA Probably 0
Spink et al. (2011) ²⁰³	Not supervised	IG: 66 % adherence. 52 % completed 75 % or more of the requested three exercise sessions weekly. 55 % wore orthoses most of the time.	28 drop out: 19 in IG and 9 in CG	91 %	NA	NA
Suda et al. (2024) ¹⁹¹	IG: 1/w in person + 1/w remote; CG: 1/w remote	NA N = 87 due to biomechanics not assessed in the other eligible individuals	NA	NA	NA	NA
Taddei et al. (2020) ²²³	IG: 1/w in person + 1/w remote; CG: 1/w remote	80.4 % in IG NA in CG.	0 (deducted)	100 % (deducted)	NA Probably 0	NA Probably 0
Utsahachant et al. (2023) ¹⁹²	3d/w videoconference call	NA	0	100 %	NA Probably 0	NA Probably 0
Xiao et al. (2022) ²²⁴	100 % supervised	100 %	0	100 %	NA Probably 0	NA Probably 0

Legend: /w (per week), [] (minimum and maximum), > (more than), ≥ (at least), CG (control group), d (day), IG (intervention group), MRS (minimalist running shoes), NA (not available), PIFM ex. (plantar intrinsic foot muscle exercises), and SD (standard deviation).

1.D Results related to the biomechanical outcomes

	Study	Outcome domain <i>Instrument</i>	Outcome (units)	Intervention group	Control group	Between group p-value Effect size
				Mean (SD), within-group p-value	Mean (SD), within-group p-value	
MRS	Miller, E et al. (2014) ²¹⁴	Kinematics (running at self-selected speed) <i>Vicon tri-dimensional motion capture system</i>	Foot incidence at foot strike (deg) ^{MA} <i>Greater than 0 indicates forefoot strike.</i> <i>Pre: Habitual shoes</i> <i>Post: MRS for IG and habitual shoes for CG.</i>	Pre : -8.6 (6.6) Post: -2.7 (7.6); p = 0.035*	Pre: -10.6 (8.2) Post: -11.1 (7.6); p = 0.868	p = 0.011* , g = 1.07
	McCarthy et al. (2013) ²⁰⁶	Kinematics (running shod at 3.33 m·s ⁻¹ with conventional shoes approximately 25% index) <i>Tri-dimensional motion capture system</i>	Ankle angle at foot strike (deg) (negative if forefoot strike) ^{MA}	<i>Mean (SEM)</i> Pre: 0.2 (1.4) Post: - 8.4 (2.4); p < 0.001*	<i>Mean (SEM)</i> Pre: 0.4 (1.1) Post: - 0.6 (1.1); p NA ^{NS}	p < 0.01* , g = -1.34
MRS combined with PIFM ex.	McCarthy et al. (2013) ²⁰⁶	Kinematics (running shod at 3.33 m·s ⁻¹ with conventional shoes approximately 25% index) <i>Tri-dimensional motion capture system</i>	Peak ankle dorsiflexion angle (deg)	<i>Mean (SEM)</i> Pre: 22.5 (1.2) Post: 19.3 (1.1); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 21.4 (1.0) Post: 19.7 (0.9); p NA ^{NS}	p NA ^{NS} , g = -0.12
			Ankle ROM, absorption phase (deg)	<i>Mean (SEM)</i> Pre: 22.3 (0.7) Post: 27.7 (1.9); p < 0.01*	<i>Mean (SEM)</i> Pre: 20.9 (1.0) Post: 20.3 (1.0); p NA ^{NS}	p < 0.001* , g = 1.56
			Ankle angle toe off (deg)	<i>Mean (SEM)</i> Pre: - 27.1 (2.0) Post: - 33.2 (1.9); p < 0.01*	<i>Mean (SEM)</i> Pre: - 21.3 (2.8) Post: - 23.1 (3.0); p NA ^{NS}	p < 0.05* , g = -1.22
			Knee angle at foot strike (deg)	<i>Mean (SEM)</i> Pre: 16.1 (1.5) Post: 15.9 (1.3); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 15.7 (1.4) Post: 14.8 (0.7); p NA ^{NS}	p NA ^{NS} , g = 0.34
			Peak knee flexion angle (deg)	<i>Mean (SEM)</i> Pre: 46.7 (1.8) Post: 43.7 (1.5); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 44.5 (1.6) Post: 43.1 (1.3); p NA ^{NS}	p NA ^{NS} , g = 0.13
			Knee ROM, absorption phase (deg)	<i>Mean (SEM)</i> Pre: 30.6 (1.5) Post: 27.7 (1.8); p < 0.05*	<i>Mean (SEM)</i> Pre: 28.8 (0.6) Post: 28.3 (1.0); p NA ^{NS}	p NA ^{NS} , g = -0.13
			Knee angle toe off (deg)	<i>Mean (SEM)</i> Pre: 12.0 (2.4) Post: 9.8 (1.5); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 13.9 (2.5) Post: 13.7 (1.9); p NA ^{NS}	p NA ^{NS} , g = -0.70
		Spatiotemporal parameters (shod at 3.33 m·s ⁻¹ with conventional shoes approximately 25% index)	Stance phase duration (ms)	<i>Mean (SEM)</i> Pre: 251 (8) Post: 243 (8); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 238 (7) Post: 237 (9); p NA ^{NS}	p NA ^{NS} , g = 0.22
			Heel-toe latency (ms) <i>(negative if forefoot strike pattern)</i>	<i>Mean (SEM)</i> Pre: 21.8 (1.8) Post: - 3.6 (7.7); p < 0.001*	<i>Mean (SEM)</i> Pre: 20.8 (1.7) Post: 21.3 (1.6); p NA ^{NS}	p < 0.001* , g = -1.46

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Kinematics (running barefoot at 3.33 m·s ⁻¹)	Ankle angle at foot strike (deg)	<i>Mean (SEM)</i> Pre: – 5.5 (2.5) Post: – 7.1 (2.1); p NA ^{NS}	<i>Mean (SEM)</i> Pre: – 4.2 (2.0) Post: – 4.2 (1.7); p NA ^{NS}	p NA ^{NS} , g = -0.48
	Peak ankle dorsiflexion angle (deg)	<i>Mean (SEM)</i> Pre: 22.0 (1.8) Post: 20.9 (1.8); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 20.0 (1.6) Post: 18.5 (1.0); p NA ^{NS}	p NA ^{NS} , g = 0.53
	Ankle ROM, absorption phase (deg)	<i>Mean (SEM)</i> Pre: 27.5 (2.3) Post: 28.0 (2.4); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 24.2 (1.3) Post: 22.7 (1.6); p NA ^{NS}	p NA ^{NS} , g = 0.82
	Ankle angle toe off (deg)	<i>Mean (SEM)</i> Pre: – 34.3 (2.3) Post: – 38.3 (2.0); p < 0.01*	<i>Mean (SEM)</i> Pre: – 26.8 (3.5) Post: – 28.2 (3.8); p NA ^{NS}	p NA ^{NS} , g = -1
	Knee angle at foot strike (deg)	<i>Mean (SEM)</i> Pre: 19.8 (1.0) Post: 19.8 (1.4); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 17.3 (1.3) Post: 17.7 (1.0); p NA ^{NS}	p NA ^{NS} , g = 0.54
	Peak knee flexion angle (deg)	<i>Mean (SEM)</i> Pre: 42.2 (1.8) Post: 40.2 (1.9); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 40.9 (1.5) Post: 40.1 (1.1); p NA ^{NS}	p NA ^{NS} , g = 0.02
	Knee ROM, absorption phase (deg)	<i>Mean (SEM)</i> Pre: 22.4 (1.4) Post: 20.4 (1.3); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 23.6 (0.6) Post: 22.4 (1.3); p NA ^{NS}	p NA ^{NS} , g = -0.48
	Knee angle toe off (deg)	<i>Mean (SEM)</i> Pre: 11.6 (2.3) Post: 10.0 (1.8); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 14.3 (2.1) Post: 13.8 (2.2); p NA ^{NS}	p NA ^{NS} , g = -0.58
	Spatiotemporal parameters (running barefoot at 3.33 m·s ⁻¹)			
Shen, B and al. (2022) ²²⁷	Stance phase duration (ms)	<i>Mean (SEM)</i> Pre: 232 (8) Post: 230 (7); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 221 (7) Post: 219 (10); p NA ^{NS}	p NA ^{NS} , g = 0.39
	Heel-toe latency (ms) (negative if forefoot strike pattern)	<i>Mean (SEM)</i> Pre: – 7.4 (4.5) Post: – 14.3 (2.9); p NA ^{NS}	<i>Mean (SEM)</i> Pre: 2.8 (3.8) Post: 2.4 (3.6); p NA ^{NS}	p < 0.05* , g = -1.56
	Kinematics (rearfoot strike running 12 km·h ⁻¹)			
	Arch angle ROM (deg)	Pre: 11.29 (1.47) Post: 11.22 (2.26); p = 0.023*	Pre: 9.69 (3.74) Post: 9.6 (2.79); p = 0.62	p NA, g = 0.62
	Maximum arch angle (deg)	Pre: 170.07 (6.51) Post: 161.45 (5.0); p < 0.001*	Pre: 160.58 (5.47) Post: 160.20 (4.63); p = 0.733	p = 0.031* , g = 0.25
	Arch height at touchdown (cm) ^{MA} Greater values indicate forefoot strike	Pre: 1.84 (0.3) Post: 2.43 (0.3); p < 0.001*	Pre: 2.44 (0.44) Post: 2.41 (0.24); p = 0.811	p < 0.001* , g = 0.07
	Arch height ROM (cm)	Pre: 0.44 (0.19) Post: 0.57 (0.17); p = 0.023*	Pre: 0.52 (0.21) Post: 0.54 (0.18); p = 0.62	p NA, g = 0.17
	Kinetics (rearfoot strike running 12 km·h ⁻¹)			
	Arch stiffness (N·cm ⁻¹ ·kg ^{-2/3})	Pre: 266.47 (142.63) Post: 177.68 (64.47); p = 0.021*	Pre: 269.2 (210.21) Post: 203.77 (71.77); p = 0.21	p NA, g = -0.37
Motion capture system	Kinematics (fore-foot strike running 12 km·h ⁻¹)			
	Arch angle ROM (deg)	Pre: 15.47 (2.22) Post: 15.59 (2.05); p>0.05	Pre: 12.14 (3.04) Post: 13.59 (2.73); p>0.05	p NA, g = 0.80

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Exercises targeting the PIFM	Warne et al. (2017) ²⁰⁴	<i>Motion capture system</i>	Maximum arch angle (deg)	Pre: 169.48 (18.27) Post: 161.02 (4.72); p>0.05	Pre: 158.4 (7.41) Post: 159.92 (4.0); p>0.05	p NA, g = 0.24
			Arch height at touchdown (cm)	Pre: 2.47 (0.33) Post: 2.72 (0.3); p = 0.04*	Pre: 2.71 (0.56) Post: 2.68 (0.2); p = 0.845	p NA, g = 0.15
			Arch height ROM (cm)	Pre: 0.81 (0.19) Post: 0.87 (0.19); p>0.05	Pre: 0.76 (0.21) Post: 0.84 (0.19); p>0.05	p NA, g = 0.15
		Kinetics (fore-foot strike running 12 km·h ⁻¹) <i>Motion capture system</i>	Arch stiffness (N·cm ⁻¹ ·kg ^{-2/3})	Pre: 127.51 (30.25) Post: 119.15 (30.10); p>0.05	Pre: 161.74 (76.3) Post: 132.75 (30.47); p>0.05	p NA, g = -0.43
	Warne et al. (2017) ²⁰⁴	Kinetics (running 11 km·h ⁻¹ in conventional shoes) <i>Motion capture system + force plate</i>	Loading rate (BW·s ⁻¹)	Pre: 55.56 (18.84) Post: 47.54 (16.46); p>0.05	Pre: 60.88 (15.51) Post: 49.56 (12.21); p = 0.08	p < 0.001* pre; p = 0.016* post, g = -0.13
			Impact peak force (BW)	Pre: 1.43 (0.36) Post: 1.39 (0.42); p>0.05	Pre: 1.58 (0.3) Post: 1.48 (0.4); p>0.05	p NA, g = -0.21
			Vertical stiffness (N·m ⁻¹)	Pre: 31246.16 (3568.69) Post: 31569.62 (3508.29); p>0.05	Pre: 32548.33 (6038.81) Post: 35024.39 (5052.69); p>0.05	p = 0.07, g = -0.77
			Knee stiffness (N·m ⁻¹)	Pre: 6.2 (1.7) Post: 6.99 (1.94); p>0.05	Pre: 7.56 (1.81) Post: 7.29 (1.76); p>0.05	p NA, g = -0.16
			Ankle stiffness (N·m ⁻¹)	Pre: 10.73 (5.63) Post: 10.91 (5.31); p>0.05	Pre: 9.5 (2.82) Post: “(1) (1.89)” (possible error in reporting); p = 0.03*	p NA, g not calculated due to the possible error in reporting
	Day and Michael (2019) ²¹⁰	Kinematics (running 14 km·h ⁻¹) <i>Motion capture system</i>	MTP joint sagittal plane ROM (deg)	Pre: 19.3 (3.4) Post: 20.0 (5.4); p = 0.514	Pre: 16.4 (4.5) Post: 19.0 (4.0); p < 0.048*	p = 0.027* , g = 0.19
			Peak ankle PF velocity (rad·s ⁻¹)	Pre: 13.9 (1.6) Post: 14.1 (1.5)	Pre: 14.4 (3.9) Post: 13.4 (2.5)	p = 0.459, g = 0.33
		Kinetics (running 14 km·h ⁻¹) <i>Motion capture system + force plate</i>	Peak MTP joint sagittal moment (Nm·kg ⁻¹) ^{MA}	Pre: 0.56 (0.14) Post: 0.59 (0.23)	Pre: 0.47 (0.12) Post: 0.55 (0.12)	p = 0.099, g = 0.20
			MTP joint sagittal moment at peak dorsiflexion (Nm·kg ⁻¹)	Pre: 0.15 (0.06) Post: 0.16 (0.08)	Pre: 0.17 (0.06) Post: 0.22 (0.08)	p = 0.455, g = -0.71
			MTP joint stiffness (Nm·kg ⁻¹ ·deg ⁻¹)	Pre: 0.008 (0.004) Post: 0.010 (0.008)	Pre: 0.010 (0.004) Post: 0.012 (0.007)	p = 0.493, g = -0.25
			Peak ankle sagittal moment (Nm·kg ⁻¹) ^{MA}	Pre: 3.40 (0.39) Post: 3.26 (0.51)	Pre: 3.33 (0.59) Post: 3.54 (0.38)	p = 0.754, g = -0.58
		Spatiotemporal parameters (running 14 km·h ⁻¹) <i>Force plate</i>	Foot contact time (s) ^{MA}	Pre: 0.21 (0.02) Post: 0.21 (0.02)	Pre: 0.20 (0.01) Post: 0.20 (0.01)	p = 0.249, g = 0.57

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Elsayed, W and al. (2023) ¹⁹⁴	Kinematics (running 18 km·h ⁻¹) <i>Motion capture system</i>	MTP joint sagittal plane ROM (deg)	Pre: 21.1 (4.7) Post: 22.7 (5.6)	Pre: 17.1 (5.4) Post: 18.6 (2.7)	p = 0.396, g = 0.85
		Peak ankle PF velocity (rad·s ⁻¹)	Pre: 17.9 (2.1) Post: 18.3 (1.3)	Pre: 16.1 (1.1) Post: 14.7 (1.7)	p = 0.279, g = 2.29
	Kinetics (running 18 km·h ⁻¹) <i>Motion capture system + force plate</i>	Peak MTP joint sagittal moment (Nm·kg ⁻¹)	Pre: 0.77 (0.17) Post: 0.77 (0.20)	Pre: 0.63 (0.15) Post: 0.60 (0.10)	p = 0.582, g = 0.98
		MTP joint sagittal moment at peak dorsiflexion (Nm·kg ⁻¹)	Pre: 0.22 (0.09) Post: 0.23 (0.10)	Pre: 0.20 (0.08) Post: 0.21 (0.08)	p = 0.464, g = 0.21
		MTP joint stiffness (Nm·kg ⁻¹ ·deg ⁻¹)	Pre: 0.011 (0.006) Post: 0.012 (0.009)	Pre: 0.013 (0.007) Post: 0.012 (0.005)	p = 0.404, g = 0
		Peak ankle sagittal moment (Nm·kg ⁻¹)	Pre: 3.60 (0.59) Post: 3.70 (0.50)	Pre: 3.55 (0.55) Post: 3.55 (0.44)	p = 0.537, g = 0.30
	Spatiotemporal parameters (running 18 km·h ⁻¹) <i>Force plate</i>	Foot contact time (s)	Pre: 0.18 (0.01) Post: 0.17 (0.01); p>0.05	Pre: 0.17 (0.01) Post: 0.19 (0.01); p < 0.001*	p < 0.001* , g = -1.89
	Kinematics (running 22 km·h ⁻¹) <i>Motion capture system</i>	MTP joint sagittal plane ROM (deg)	Pre: 25.5 (4.1) Post: 25.1 (4.9)	Pre: 17.9 (6.6) Post: 18.7 (5.4)	p = 0.429, g = 1.17
		Peak ankle PF velocity (rad·s ⁻¹)	Pre: 21.1 (2.4) Post: 21.5 (1.5)	Pre: 17.5 (0.6) Post: 17.0 (2.5)	p = 0.748, g = 2.17
	Kinetics (running 22 km·h ⁻¹) <i>Motion capture system + force plate</i>	Peak MTP joint sagittal moment (Nm·kg ⁻¹)	Pre: 0.99 (0.18) Post: 0.99 (0.26)	Pre: 0.86 (0.28) Post: 0.91 (0.16)	p = 0.780, g = 0.33
		MTP joint sagittal moment at peak dorsiflexion (Nm·kg ⁻¹)	Pre: 0.32 (0.05) Post: 0.34 (0.04)	Pre: 0.32 (0.08) Post: 0.36 (0.12)	p = 0.191, g = -0.24
		MTP joint stiffness (Nm·kg ⁻¹ ·deg ⁻¹)	Pre: 0.014 (0.003) Post: 0.014 (0.004)	Pre: 0.020 (0.006) Post: 0.020 (0.008)	p = 0.083, g = -0.96
		Peak ankle sagittal moment (Nm·kg ⁻¹)	Pre: 4.21 (0.63) Post: 4.51 (0.54)	Pre: 4.29 (0.74) Post: 4.47 (0.55)	p = 0.611, g = 0.07
	Spatiotemporal parameters (running 22 km·h ⁻¹) <i>Force plate</i>	Foot contact time (s)	Pre: 0.15 (0.01) Post: 0.15 (0.01)	Pre: 0.15 (0.01) Post: 0.15 (0.01)	p = 0.583, g = 0
	Kinetics (walking) <i>EMEd-X® pressure plate</i>	Plantar force (N) (<i>probably peak vertical</i>) ^{MA}	Pre: 838.3 (184.3) Post: 877.0 (171.9)	Pre: 816.4 (294.9) Post: 841.1 (308.4)	p = 0.46, g = 0.14
		Plantar pressure (kPa) (<i>probably peak</i>)	Pre: 696.0 (180.8) Post: 882.2 (197.5); p < 0.001*	Pre: 739.6 (189.8) Post: 815.3 (191.8); p < 0.001*	p = 0.13, g = 0.34
Spatiotemporal (walking) <i>EMEd-X® pressure plate</i>		Foot contact area (cm ²) (<i>probably peak</i>)	Pre: 148.9 (18.0) Post: 146.3 (16.3)	Pre: 144.3 (21.2) Post: 143.5 (24.4)	p = 0.16, g = 0.13

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Goldmann et al. (2013) ²⁰²	Kinetics (walking 1.6 m·s ⁻¹) <i>Vicon tri-dimensional motion capture system + force plates</i>	Peak ankle dorsiflexion moment (Nm·kg ⁻¹)	Pre: 1.71 (0.10) Post: 1.72 (0.09); p=0.67	Pre: 1.73 (0.17) Post: 1.74 (0.17); p=0.34	p=0.36, g = -0.15
		Peak metatarsophalangeal dorsiflexion moment (Nm·kg ⁻¹)	Pre: 0.22 (0.07) Post: 0.24 (0.06); p=0.052	Pre: 0.20 (0.06) Post: 0.22 (0.07); p=0.22	p=0.61, g = 0.3
		Mechanical power of the ankle and metatarsophalangeal joints (W·kg ⁻¹)	<i>Described in methods, results NA.</i> p>0.05	<i>Described in methods, results NA.</i> p>0.05	p NA, g NA
	Kinetics (running 3.5 m·s ⁻¹)	Peak ankle dorsiflexion moment (Nm·kg ⁻¹) ^{MA}	Pre: 2.70 (0.17) Post: 2.71 (0.20); p=0.85	Pre: 2.64 (0.20) Post: 2.64 (0.15); p=0.97	p=0.87, g = 0.38
		Peak metatarsophalangeal dorsiflexion moment (Nm·kg ⁻¹) ^{MA}	Pre: 0.25 (0.08) Post: 0.27 (0.09); p=0.50	Pre: 0.24 (0.09) Post: 0.25 (0.07); p=0.64	p=0.42, g = 0.24
		Mechanical power of the ankle and metatarsophalangeal joints (W·kg ⁻¹)	<i>Described in methods, results NA.</i> p>0.05	<i>Described in methods, results NA.</i> p>0.05	p NA, g NA
	Kinetics (vertical jumping)	Peak ankle dorsiflexion moment (Nm·kg ⁻¹)	Pre: 1.68 (0.37) Post: 1.68 (0.35); p=0.98	Pre: 1.68 (0.45) Post: 1.69 (0.45); p=0.89	p=0.37, g = -0.02
		Peak metatarsophalangeal dorsiflexion moment (Nm·kg ⁻¹)	Pre: 0.22 (0.16) Post: 0.23 (0.18); p = 0.66	Pre: 0.15 (0.10) Post: 0.18 (0.10); p = 0.25	p = 0.14, g = 0.32
		Mechanical power of the ankle and metatarsophalangeal joints (W·kg ⁻¹)	<i>Described in methods, results NA.</i> p>0.05	<i>Described in methods, results NA.</i> p>0.05	p NA, g NA
	Kinetics (horizontal jumping)	Peak ankle dorsiflexion moment (Nm·kg ⁻¹)	Pre: 1.70 (0.17) Post: 1.81 (0.19); p = 0.08	Pre: 1.66 (0.32) Post: 1.79 (0.19); p = 0.25	p = 0.83, g = 0.10
		Peak metatarsophalangeal dorsiflexion moment (Nm·kg ⁻¹)	Pre: 0.69 (0.10) Post: 0.75 (0.14); p = 0.012*	Pre: 0.68 (0.15) Post: 0.70 (0.12); p = 0.48	p = 0.40, g = 0.37
		Mechanical power of the ankle and metatarsophalangeal joints (W·kg ⁻¹)	<i>Described in methods, results NA.</i> p>0.05	<i>Described in methods, results NA.</i> p>0.05	p NA, g NA
Hartmann et al. (2009) ²⁰⁵	Spatiotemporal parameters (walking) <i>Accelerometer system</i>	Walking speed (m·s ⁻¹)	Pre: 1.33 (0.19) Post: 1.41 (0.19); p=0.001*	Pre: 1.38 (0.15) Post: 1.48 (0.14); p < 0.001*	p>0.05, g = -0.41
		Cadence (steps·min ⁻¹)	Pre: 115.1 (10.1) Post: 117.8 (10.2); p=0.017 *	Pre: 117.6 (5.8) Post: 120.5 (5.5); p=0.004*	p>0.05, g = -0.32
		Step duration (s) ^{MA}	Pre: 0.53 (0.04) Post: 0.51 (0.05); p=0.057	Pre: 0.51 (0.03) Post: 0.50 (0.02); p=0.004*	p>0.05, g = 0.26
		Step length (m)	Pre: 0.69 (0.07) Post: 0.72 (0.07); p=0.001*	Pre: 0.70 (0.07) Post: 0.74 (0.06); p < 0.001*	p>0.05, g = -0.30
Kim, H and al. (2022) ²¹⁷	Kinematics (single-leg-squat) <i>Phone camera</i>	Dynamic valgus index (deg)	IG1 Pre: 31.94 (9.22); post: 17.37 (5.79) IG2 Pre: 28.95 (5.85); post: 13.66 (8.67)	Pre: 30.35 (8.02) Post: 19.33 (7.62)	p>0.0017 ^{NS} , g = -0.28 (IG1 vs CG) and p>0.017 ^{NS} , g = -0.68 (IG2 vs CG)

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Matias et al. (2022) ²¹⁶	Kinematics (running at self-selected speed) <i>Vicon tri-dimensional motion capture system</i>	Medial longitudinal arch ROM (deg) ^{MA}	Pre: 6.16 (8.14) Post: 0.17 (6.86); p NA	Pre: 3.59 (7.89) Post: 2.88 (5.36); p NA	p = 0.024* , g = -0.44
		Shank-calcaneus inversion peak (deg) (<i>inversion in negatives</i>)	Pre: -0.56 (7.42) Post: -5.74 (6.31); p NA	Pre: -3.30 (8.71) Post: -3.51 (5.70); p NA	p = 0.037* , g = -0.37
		Shank-calcaneus eversion peak (deg) (<i>eversion in positives</i>)	Pre: 6.72 (3.29) Post: 5.90 (2.95)	Pre: 6.89 (2.35) Post: 6.39 (1.88)	p = 0.557, g = -0.20
		Sagittal-plane inclination of the metatarsal bones to the ground and metatarsal bone divergence in the foot transverse plane (deg)	<i>Continuous data - see graphs</i>		For all: p>0.05
		Medial longitudinal arch angle (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05
		1 st MTP joint abduction (deg)	<i>Continuous data - see graph</i> p>0.05	<i>Continuous data - see graph</i> p>0.05	p = 0.003* mid to late stance (IG>CG)
		1 st MTP joint dorsiflexion (deg)	<i>Continuous data - see graph</i> p>0.05	<i>Continuous data - see graph</i> p>0.05	p = 0.003* late stance (IG>CG)
		Lisfranc dorsiflexion (deg)	<i>Continuous data - see graph</i> p>0.05	<i>Continuous data - see graph</i> p>0.05	p = 0.016* late stance (IG<CG)
		Lisfranc adduction (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05
		Lisfranc inversion (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05
		Calcaneus-metatarsals adduction (deg)	<i>Continuous data - see graph</i> p < 0.017* late stance (pre>post)	<i>Continuous data - see graph</i> p>0.05	p = 0.008* mid to late stance (IG>CG)
		Calcaneus-metatarsals PF (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05
		Calcaneus-metatarsals inversion (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05
		Chopart dorsiflexion (deg)	<i>Continuous data - see graph</i> p = 0.005* early to mid-stance (pre<post)	<i>Continuous data - see graph</i> p < 0.001* whole stance (pre<post)	p = 0.014* early stance and p = 0.013* late stance (IG<CG)
		Chopart abduction (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05
		Chopart inversion (deg)	<i>Continuous data - see graph</i> p>0.05	<i>Continuous data - see graph</i> p=0.009* early stance (pre<post)	p=0.014* early stance (IG>CG)
		Shank-calcaneus inversion (deg)	<i>Continuous data - see graph</i> p < 0.05* early to mid-stance (pre>post)	<i>Continuous data - see graph</i> p < 0.05 whole stance (pre>post)	p=0.05* initial contact (IG>CG)
		Shank-calcaneus adduction (deg)	<i>Continuous data - see graph</i> p=0.025* initial contact and p=0.023* toe off (pre>post)	<i>Continuous data - see graph</i> p=0.025* initial contact and p=0.011* mid to late-stance (pre>post)	p < 0.001* whole stance (IG<CG)
		Shank-calcaneus PF (deg)	<i>Continuous data - see graph</i>	<i>Continuous data - see graph</i>	p>0.05

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	Kinetics (running at self-selected speed) <i>Vicon tri-dimensional motion capture system + force plate</i>	Vertical impact peak (BW) ^{MA}	Pre: 1.13 (0.39) Post: 1.14 (0.55)	Pre: 1.21 (0.44) Post: 1.09 (0.58)	p=0.129, g = 0.09
		Loading rate (BW·s ⁻¹)	Pre: 75.19 (46.43) Post: 77.17 (57.52)	Pre: 73.48 (43.17) Post: 72.84 (65.73)	p = 0.537, g = 0.07
		Peak breaking force (BW) ^{MA}	Pre: -0.24 (0.05) Post: -0.24 (0.07)	Pre: -0.24 (0.05) Post: -0.24 (0.05)	p = 0.934, g = 0
Okamura et al. (2020b) ²²⁶	Kinematics (walking) <i>Motion capture system</i>	Medial longitudinal arch ROM (mm) ^{MA}	Pre: 6.2 (1.7) Post: 6.2 (1.5)	Pre: 5.9 (2.6) Post: 5.4 (2.5)	p>0.05, g = 0.37
		Peak vertical ground reaction force (% BW) ^{MA}	Pre: 109.1 (4.5) Post: 108.3 (5.7)	Pre: 107.5 (6.2) Post: 108.4 (6.5)	p>0.05, g = -0.02
	Kinetics (walking) during the second half of stance phase <i>Force plate</i>	Peak anterior ground reaction force (% BW) ^{MA}	Pre: 24.0 (2.0) Post: 24.8 (2.5)	Pre: 22.6 (4.4) Post: 23.8 (4.7)	p>0.05, g = 0.25
		Peak medial ground reaction force (% BW)	Pre: 6.0 (1.6) Post: 5.3 (1.3); p = 0.007*	Pre: 6.2 (0.9) Post: 6.2 (1.3); p>0.01 ^{NS}	p = 0.033* , g = -0.66
		Stance phase duration (ms) ^{MA}	Pre: 610.1 (36.3) Post: 600.4 (34.5)	Pre: 623.6 (36.8) Post: 618.8 (47.1)	p>0.05, g = -0.43
	Spatiotemporal (walking) <i>Motion capture system</i>	Time to reach minimal navicular height (ms) ^{MA}	Pre: 74.1 (5.4) Post: 69.7 (8.4); p = 0.007*	Pre: 74.7 (5.2) Post: 75.0 (5.5); p>0.01 ^{NS}	p = 0.01* , g = -0.71
Suda et al. (2024) ¹⁹¹	Foot joint coordination <i>3D movement analysis with multi-segment foot model</i>	Coordination frequency in the frontal plane, in-phase with proximal dominance	<i>Continuous data – see graphs</i> p significant* (pre>post)	<i>Continuous data – see graphs</i> p NA ^{NS}	p < 0.001*
		Coordination frequency in the frontal plane, anti-phase with distal dominance	<i>Continuous data – see graphs</i> p NA ^{NS}	<i>Continuous data – see graphs</i> p NA ^{NS}	p = 0.049* (IG>CG)
		Coordination frequency in the transverse plane, anti-phase with proximal dominance	<i>Continuous data – see graphs</i> p NA ^{NS}	<i>Continuous data – see graphs</i> p NA ^{NS}	p = 0.004* (IG>CG)
	2° All other foot joint pairs	Coordination frequency for all phases for both proximal or distal dominance	<i>Continuous data – see graphs</i>	<i>Continuous data – see graphs</i>	p>0.05
Taddei et al. (2020) ²²³	Kinematics (running self-selected speed) <i>Motion capture system</i>	Medial longitudinal arch ROM (deg) ^{MA}	Pre: 4.2 (2.4) Post: 3.6 (2.3)	Pre: 4.6 (2.2) Post: 4.6 (1.8)	p = 0.338, g = -0.47
		Medial longitudinal arch stiffness (N·cm ⁻¹ ·kg ^{-2/3}) ^λ	Pre: 165 (NA) Post: 208 (NA)	Pre: 141 (NA) Post: 159 (NA)	p = 0.781, g NA
	Kinetics (running self-selected speed) <i>Motion capture system+force platform</i>	Vertical impulse (BW·s) ^{MA}	Pre: 65.86 (7.91) Post: 67.85 (9.89); p NA	Pre: 74.27 (6.99) Post: 73.49 (6.47); p NA	p = 0.007* , g = -0.66 (pre: IG<CG)
		Horizontal impulse (BW·s) ^{MA}	Pre: 5.89 (1.29) Post: 6.33 (1.75)	Pre: 6.17 (1.26) Post: 6.14 (1.7)	p = 0.315, g = 0.11

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Utsahachant et al. (2023) ¹⁹²	Kinematics during walking <i>Motion capture system</i>	Largest medial longitudinal arch angle during the push-off phase (deg)	IG1: Pre: 158.30 (7.66) Post: 152.41 (9.70); p < 0.017* IG2: Pre: 158.54 (5.40) Post: 152.91 (5.96); p < 0.017*	Pre: 157.26 (5.62) Post: 157.22 (5.49); p>0.017 ^{NS}	p = 0.234, g = -0.59 (IG1 vs CG); p = 0.340, g = -0.73 (IG2 vs CG)
		Smallest medial longitudinal arch angle during the midstance phase (deg)	<i>Median (IQR)</i> IG1: Pre: 152.03 (146.35, 153.30) Post: 147.67 (143.57, 154.39); p < 0.017* IG2: Pre: 152.14 (150.55, 155.34) Post: 149.15 (145.98, 153.37); p < 0.017*	<i>Median (IQR)</i> Pre: 152.06 (148.08, 149.91) Post: 151.13 (150.58, 154.36); p>0.017 ^{NS}	p = 0.059, g = -0.51 (IG1 vs CG); p = 0.152, g = -0.51 (IG2 vs CG)
		Largest medial longitudinal arch angle during the loading phase (deg)	IG1: Pre: 158.78 (5.63) Post: 154.79 (6.52); p < 0.017* IG2: Pre: 157.61 (5.56) Post: 154.17(7.48); p < 0.017*	Pre: 156.87(4.72) Post: 158.39 (4.10); p>0.017 ^{NS}	p = 0.360, g = -0.64 (IG1 vs CG); p = 0.209, g = -0.68 (IG2 vs CG)
		Medial longitudinal arch ROM (deg) ^{MA}	IG1 ^{MA} : Pre: 15.87 (4.88); Post: 13.85 (5.27) IG2: Pre: 16.55 (2.39); Post: 14.79 (2.40)	Pre: 14.99 (4.76) Post: 14.89 (3.60)	p = 0.06, g = -0.22 (IG1 vs CG) and g = -0.03 (IG2 vs CG)
		Midfoot eversion excursion angle (deg)	IG1: Pre: 7.83 (1.64); Post: 7.54 (1.56) IG2: Pre: 7.52 (1.68); Post: 7.86 (1.79)	Pre: 8.42 (2.02) Post: 8.30 (2.08)	p = 0.395, g = -0.40 (IG1 vs CG) and g = -0.22 (IG2 vs CG)
	Kinetics during walking <i>Plantar pressure plate</i>	Center of pressure excursion index (% of foot width; <i>higher means more laterally deviated centre of pressure excursion</i>)	IG1: Pre: 18.20 (3.69) Post: 20.04 (4.20); p>0.017 ^{NS} IG2: Pre: 17.22 (3.66) Post: 20.88 (4.11); p>0.017 ^{NS}	Pre: 15.27 (3.21) Post: 15.23 (3.42); p>0.017 ^{NS}	p = 0.044* , g = 1.22 (IG1 vs CG); p < 0.001* , g = 1.45 (IG2 vs CG)
	Spatio-temporal parameters during walking <i>Motion capture system</i>	Time to lowest medial longitudinal arch (% of stance) ^{MA}	IG1 ^{MA} : Pre: 80.07 (2.99); Post: 78.67 (3.37); p = 0.002* IG2: Pre: 78.53 (2.50); Post: 77.27 (2.19); p = 0.005*	Pre: 79.87 (2.85) Post: 80.00 (2.65); p>0.017 ^{NS}	p = 0.588, g = -0.43 (IG1 vs CG); p = 0.03 ^{NS} , g = -1.09 (IG2 vs CG)

Legend: **Bold*** (significant p-value at p < 0.05 or lower if it results from a post-hoc test), BW (body weight), CG (control group), g (Hedge's g effect size), IG (intervention group), IQR (interquartile range), ^{MA} (data included in the meta-analysis), MRS (minimalist running shoes), MTP (metatarsophalangeal), NA (not available), NS (indicates that p is non-significant, for the cases where the value was not reported in the articles), p (p-value), PF (plantar flexion), PIFM ex. (plantar intrinsic foot muscle exercises), Post (post-intervention), Pre (pre-intervention), ROM (range of motion), SD (standard deviation), SEM (standard error of the mean), [^] (data was estimated from graph).

1.E Meta-analyses of biomechanical outcomes

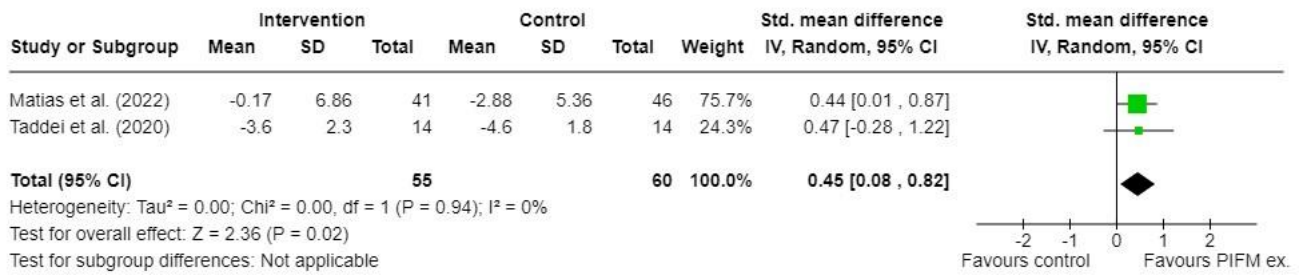


Figure 1.E.1 Medial longitudinal arch mobility during running



Figure 1.E.2 Medial longitudinal arch mobility during walking

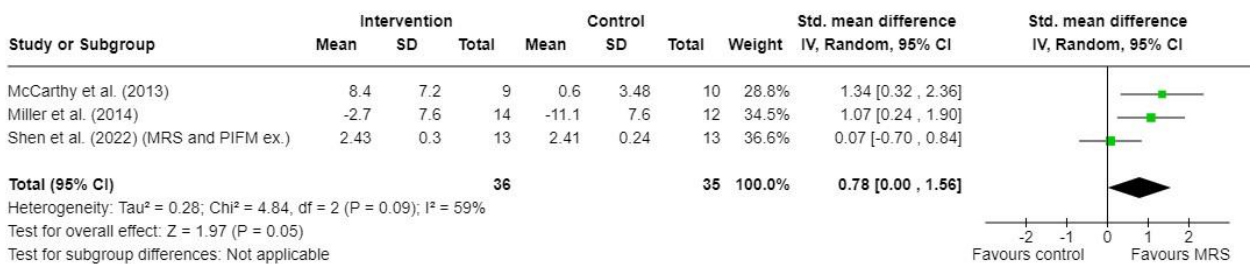


Figure 1.E.3 Foot incidence at foot strike while running

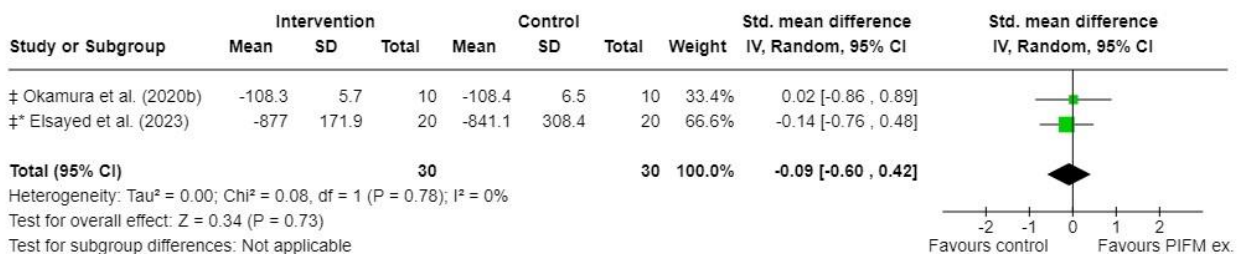


Figure 1.E.4 Vertical ground reaction forces during walking



Figure 1.E.5 Vertical ground reaction forces during running

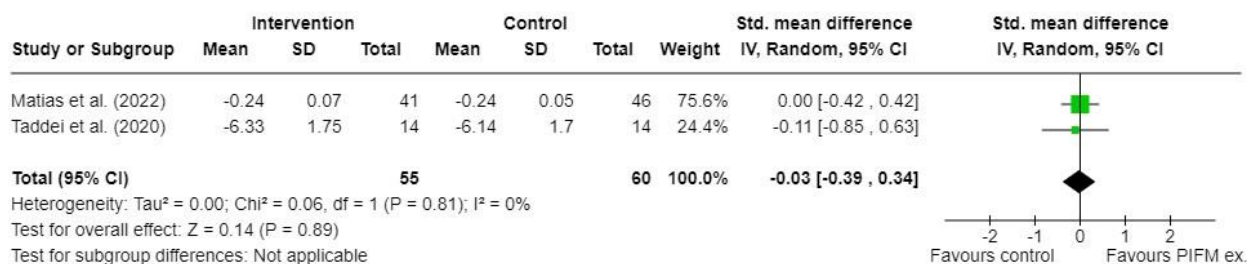


Figure 1.E.6 Anterior ground reaction forces during running

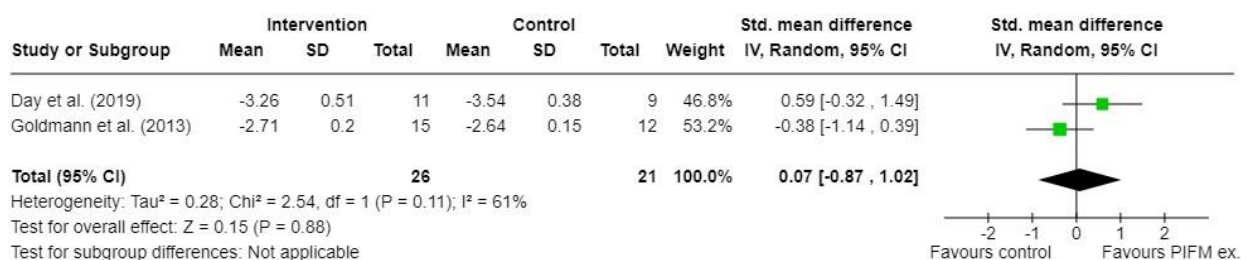


Figure 1.E.7 Peak ankle sagittal plane moment during running

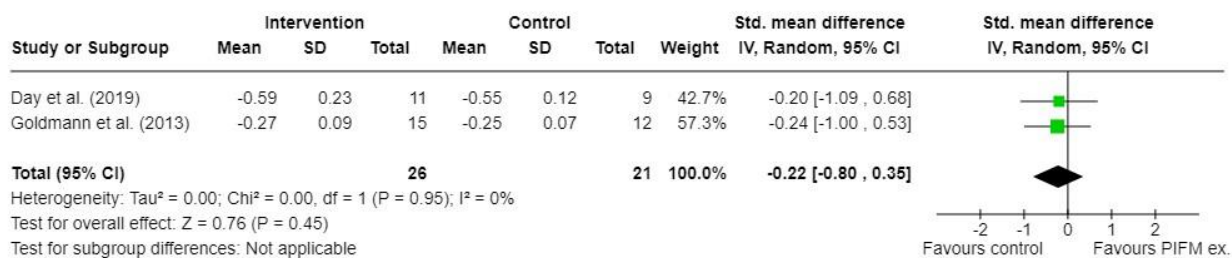


Figure 1.E.8 Peak metatarsophalangeal sagittal plane moment during running

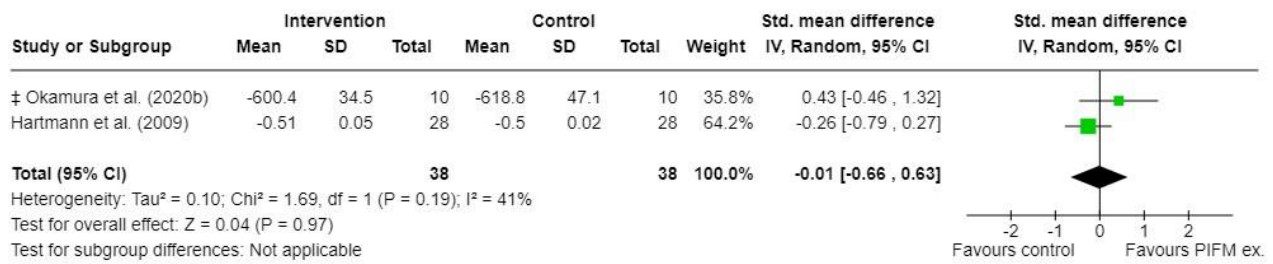


Figure 1.E.9 Stance phase duration during walking

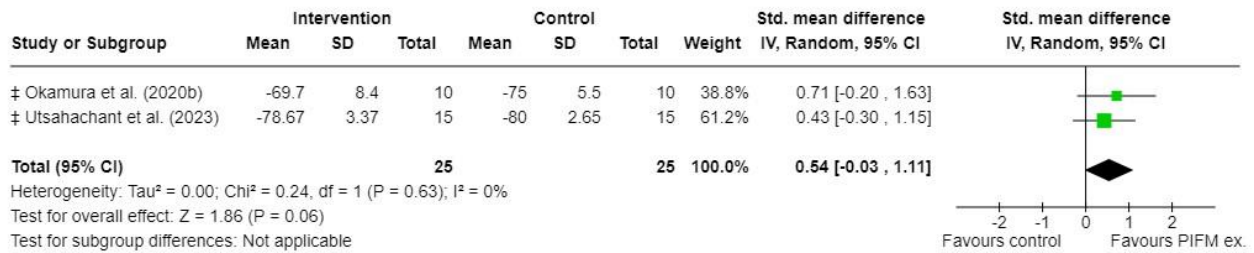


Figure 1.E.10 Time to lowest medial longitudinal arch during walking

Legend: MRS (minimalist running shoes), PIFM ex. (exercises targeting the plantar intrinsic foot muscles), CI (confidence interval), SD (standard deviation), ‡ (participants diagnosed with pes planus), and * (symptomatic participants).

Appendices of Chapter 2: Randomized crossover study

2.A Definitions of running biomechanical variables

Outcomes are defined, along with their units. Further description is available in the user manual provided by h/p/cosmos® (Arsalis). The user manual was retrieved online from https://www.arsalis.com/wp-content/uploads/2023/12/User-manual_hpcosmos_gaitway_3d.pdf, accessed September, 4, 2025.

1) Hypothesis-driven outcomes (alphabetical order)

Active peak force (%BW): The vertical peak force during the foot contact, after the impact peak.

Contact duration (%stride): Duration for which force is applied during a foot strike, which corresponds to the duration from heel contact to toe-off.

Impact Peak Force (%BW): First vertical peak on the vertical force curve shortly after foot impact.

Vertical stiffness (%BW·mm⁻¹): Stiffness of the center of mass computed as the ratio of the maximal vertical force to the vertical displacement of the center of mass. Vertical displacement of the centre of mass is obtained by double integration of the vertical force signal.

2) Other outcomes (alphabetical order)

Antero-posterior mechanical work (J·kg⁻¹·m⁻¹): Antero-posterior positive mechanical work to accelerate the body center of mass in the forward direction. This outcomes is obtained by summing the increments of the kinetic antero-posterior mechanical energy curve during the stride. The value is then normalized per body mass and distance.

Braking impulse (%BW·s): The area under the antero-posterior force time curve during the braking phase of the step.

Braking peak force (%BW): Antero-posterior braking peak force decelerating the body.

Cadence (steps per min): The number of foot strikes per minute.

Duty factor: The ratio of contact duration to stride duration.

Leg stiffness (%BW·mm⁻¹): Stiffness of the leg spring computed as the ratio of the maximal vertical force to the compression of the leg. Leg compression is calculated as the difference in distance between leg length measured at rest (trochanter to medial malleolus) and leg length at maximal compression estimated using McMahon and Cheng's method.⁴²⁹

Propulsive impulse (%BW·s): The area under the antero-posterior time curve during the propulsion phase of the step.

Propulsive peak force (%BW): Antero-posterior propulsive peak force accelerating the body.

Step duration (%stride): The time from initial foot contact to heel contact of the opposite foot.

Stride duration (ms): The time duration from the initial contact of the left foot to the next initial contact of the left foot.

Support width (cm): The average lateral distance between both feet during the contact phases.

Total mechanical work ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$): The total positive mechanical work performed to displace the body centre of mass. Also called external work. The work is computed by summing the increments of the total mechanical energy curve during the stride. The total mechanical energy curve is obtained by summing the instantaneous vertical, antero-posterior and lateral energy curves

Vertical displacement of the center of mass (mm): Vertical displacement of the body center of mass during a step, obtained by double integration of the vertical force signal.

Vertical impulse (%BW·s): The integral, or area under the vertical force time curve during the foot strike.

Vertical loading rate (%BW·ms⁻¹): Slope of the vertical force time curve during the loading phase, taken from one point at 20% of first peak force and one point at 80% of first peak force.

Vertical mechanical work ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$): Vertical positive mechanical work to lift and accelerate the body center of mass in the upward direction. This outcomes is computed by summing the increments of the vertical mechanical energy curve during the stride. The value is then normalized per body mass and distance.

2.B Between-sequence comparison of the demographic characteristics of the 19 participants who completed the study

Outcome	Sequence A-B (n = 9)	Sequence B-A (n = 10)	p-value
Gender	6M/3F	10M/0F	/
Self-reported dominance	1L/8R	5L/5R	/
Age (year)	23.8 (2.8)	26.5 (5.6)	p = 0.38
Weight (kg)	73.0 (14.4)	75.0 (9.0)	p = 0.51
Height (cm)	176.3 (9.7)	179.5 (5.0)	p = 0.41
Body mass index (kg.m ⁻²)	23.3 (3.1)	23.2 (2.2)	p = 0.81
6-item Foot Posture Index	5 [-1.5, 8.5]	4.5 [2.75, 6]	p = 0.81
Neutral (0 to 5)	n = 2	n = 6	/
Pronated (6 to 9)	n = 3	n = 3	/
Extremely pronated (10 to 12)	n = 1	n = 0	/
Supinated (- 4 to -1)	n = 3	n = 0	/
Extremely supinated (-5 to -12)	n = 0	n = 1	/
Main sport			
Running	n = 3	n = 6	/
Strength training	n = 2	n = 2	/
Other [†]	n = 4	n = 2	/
Self-reported physical activity duration per week (hours)	4.4 (2.6)	3.0 (1.0)	p = 0.30

Legend: Continuous variables are presented as mean (standard deviation), and discrete variables are presented as median [interquartile range]. F (female), L (left), M (Male), mSEBT (modified star excursion balance test), n (number of participants), R (right), † (other sports; i.e. cheerleading, cycling, dance, football, ultimate frisbee and rugby). Alpha set at 0.05.

2.C Perceived fatigue, current intensity and strength at four-min intervals

Intervention	Outcome	Fam.	4 min	8 min	12 min	16 min	GLM
Intervention A (concentric)	Perceived fatigue (VAS)	/	3 [3, 4.5]	5 [4.5, 6] ^a	7 [5.5, 7] ^{a,b}	7 [6, 8] ^{a,b,c}	$p < \mathbf{0.001^*}$ $\eta^2_p = 0.81$
	Hallux strength standing (N·kg ⁻¹)	2.0 (0.6)	1.9 (0.6)	2.0 (0.7)	1.9 (0.6)	2.0 (0.7)	$p = 0.44$ $\eta^2_p = 0.05$
Intervention B (isometric + electrical stimulation)	Perceived fatigue (VAS)	/	3 [1.5, 3.5]	4 [3, 5] ^a	6 [4, 6.5] ^{a,b}	6 [4.5, 7] ^{a,b,c}	$p < \mathbf{0.001^*}$ $\eta^2_p = 0.73$
	Current intensity (mA)	14 [11, 14.5]	14 [11.5, 17]	15 [12, 17.5] ^{a,d}	15 [13, 19.5] ^{a,b,d}	16 [13.5, 21] ^{a,b,d}	$p < \mathbf{0.001^*}$ $\eta^2_p = 0.49$

Legend: Continuous quantitative variables are presented as mean (standard deviation), and discrete quantitative variables are presented as median [interquartile range]. Abbreviations: Fam. (Familiarization), GLM (general linear model), VAS (verbal analogue scale), * (significant at $p < 0.05$ for the GLM and Bonferroni correction for the post-hoc tests), ^a (statistically different from 4 minutes), ^b (statistically different from 8 minutes), ^c (statistically different from 12 minutes), ^d (statistically different from Familiarization)

2.D Running biomechanics at 9 and 12 km·h⁻¹

9 km·h⁻¹

Outcomes	Intervention A		Intervention B		Inference	
	Pre	Post	Pre	Post	GLM: p (η^2p)	Post-hoc tests tests (Bonferroni correction): p, g [CI]
Cadence (step·min⁻¹)	163.4 (7.2)	164.6 (6.4)	163.7 (5.9)	165.5 (6.3)	0.12 (0.1)	
Stride duration (ms)	735.8 (32.2)	727.9 (30.2)	734.1 (26.8)	726.1 (28.0)	0.08 (0.12)	
Step duration (% of stride)	50.0 (0.7)	50.0 (0.8)	50.1 (0.5)	50.0 (0.7)	0.70 (0.03)	
Contact duration (% of stride %)	38.5 (2.8)	38.5 (2.8)	38.8 (2.6)	38.6 (2.7)	0.74 (0.02)	
Support width (cm)	5.7 (1.9)	6.0 (2.3)	6.0 (2.7)	6.6 (2.5)	0.03* (0.15)	A: p = 1, g = 0.17 [-0.28; 0.61] B: p = 0.085, g = 0.61 [0.12; 1.09]
Vertical displacement of COM (mm)	81.3 (12.9)	79.7 (11.5)	80.0 (9.6)	78.9 (11.2)	0.46 (0.04)	
Total work (J·kg⁻¹·m⁻¹)	1.35 (0.11)	1.34 (0.11)	1.33 (0.11)	1.33 (0.12)	0.28 (0.07)	
Vertical work (J·kg⁻¹·m⁻¹)	0.87 (0.1)	0.86 (0.1)	0.86 (0.08)	0.86 (0.09)	0.47 (0.04)	
Antero-posterior work (J·kg⁻¹·m⁻¹)	0.5 (0.05)	0.49 (0.04)	0.49 (0.05)	0.49 (0.06)	0.69 (0.03)	
Vertical impulse (% BW·s)	36.6 (1.8)	36.2 (1.7)	36.6 (1.6)	36.0 (1.6)	0.03* (0.15)	A: p = 0.18, g = -0.53 [-1.0; -0.05] B: p = 0.03* , g = -0.72 [-1.21; -0.21]
Vertical loading rate (% BW·ms⁻¹)	4.6 (1.8)	4.5 (1.7)	4.4 (1.4)	4.6 (1.6)	0.68 (0.02)	
Impact peak force (% BW)	143.2 (27.4)	142.8 (29.2)	136.2 (29.1)	141.7 (27.1)	1 (0.12)	
Active peak force (% BW)	229.9 (21.3)	230.5 (21.6)	227.4 (17.9)	229.3 (20.1)	0.47 (0.04)	
Vertical stiffness (% BW·mm⁻¹)	2.86 (0.29)	2.92 (0.28)	2.86 (0.23)	2.93 (0.28)	0.196 (0.082)	
Leg stiffness (% BW·mm⁻¹)	1.91 (0.66)	1.96 (0.65)	1.9 (0.64)	1.97 (0.66)	0.02* (0.65)	A: p = 0.6, g = 0.39 [-0.74; 0.84] B: p = 0.02* , g = 0.75 [0.24; 1.24]
Braking impulse (% BW·s)	1.9 (0.2)	1.9 (0.2)	1.9 (0.2)	1.9 (0.2)	0.49 (0.04)	
Braking peak force (% BW)	24.8 (3.1)	24.5 (3.1)	24.3 (2.5)	24.6 (3.1)	0.61 (0.03)	
Propulsive impulse (% BW·s)	1.8 (0.3)	1.8 (0.3)	1.8 (0.3)	1.7 (0.3)	0.08 (0.13)	
Propulsive peak force (% BW)	22.4 (3.4)	22.3 (3.4)	22.4 (3.5)	21.9 (3.5)	0.23 (0.08)	
Duty factor	0.39 (0.03)	0.39 (0.03)	0.39 (0.03)	0.39 (0.03)	0.74 (0.02)	

Legend: A (Intervention A), B: (Intervention B), BW (body weight), CI (confidence interval), COM (center of mass), GLM (repeated measure general linear model), g (Cohen's g effect size), η^2p (partial eta squared effect size of the GLM), * and bold: significant at p<0.05. Post-hoc p-values were corrected using Bonferroni method.

12 km·h⁻¹

Outcomes	Intervention A		Intervention B		Inference	
	Pre	Post	Pre	Post	GLM: p-value (η^2p)	Post-hoc tests tests (Bonferroni correction): p, g [CI]
Cadence (step·min⁻¹)	173.2 (9.1)	175.2 (9.2)	172.9 (7.2)	174.8 (8.0)	0.04* (0.14)	A: p = 0.13, g = 0.56 [0.08; 1.03] B: p = 0.08, g = 0.61 [0.12; 1.09]
Stride duration (ms)	694.7 (37.2)	686.8 (36.7)	695.6 (30.2)	688.3 (32.9)	0.04* (0.14)	A: p = 0.09, g = -0.6 [-1.08; -0.12] B: p = 0.1, g = -0.60 [-1.07; -0.11]
Step duration (% of stride)	50.2 (0.6)	50.0 (0.6)	50.1 (0.5)	49.9 (0.6)	0.13 (0.1)	
Contact duration (% of stride)	35.0 (2.6)	34.7 (2.8)	34.9 (2.7)	34.8 (2.7)	0.56 (0.04)	
Support width (cm)	4.8 (2.2)	5.1 (2.3)	5.0 (2.4)	5.2 (2.4)	0.49 (0.04)	
Vertical displacement of COM (mm)	78.8 (14.3)	76.8 (13.4)	78.6 (11.5)	76.6 (12.8)	0.34 (0.06)	
Total work (J·kg⁻¹·m⁻¹)	1.27 (0.1)	1.26 (0.1)	1.26 (0.1)	1.25 (0.11)	0.36 (0.55)	
Vertical work (J·kg⁻¹·m⁻¹)	0.66 (0.08)	0.66 (0.08)	0.66 (0.07)	0.66 (0.08)	0.84 (0.01)	
Antero-posterior work (J·kg⁻¹·m⁻¹)	0.62 (0.05)	0.61 (0.05)	0.61 (0.05)	0.61 (0.06)	0.16 (0.09)	
Vertical impulse (% BW·s)	34.7 (2.1)	34.1 (1.9)	34.8 (1.8)	34.2 (1.8)	0.01* (0.21)	A: p = 0.11, g = -0.58 [-1.05; -0.09] B: p = 0.03* , g = -0.73 [-1.22; -0.22]
Vertical loading rate (% BW·ms⁻¹)	6.5 (2.2)	6.4 (2.2)	6.2 (2.1)	6.4 (2.3)	0.33 (0.06)	
Impact peak force (% BW)	160.6 (26.2)	163.9 (28.5)	163.0 (28.6)	163.9 (27.2)	0.37 (0.06)	
Active peak force (% BW)	245.5 (25.4)	247.3 (25.1)	245.0 (21.4)	245.7 (22.5)	0.51 (0.03)	
Vertical stiffness (% BW·mm⁻¹)	3.17 (0.37)	3.27 (0.37)	3.15 (0.3)	3.25 (0.36)	0.055 (0.13)	
Leg stiffness (% BW·mm⁻¹)	2 (0.92)	2.1 (0.98)	2 (0.89)	2.05 (0.89)	< 0.001* (0.29)	A: p < 0.001* , g = 1.03 [0.47; 1.57] B: p = 0.28, g = 0.48 [0.01; 0.94]
Braking impulse (% BW·s)	2.3 (0.3)	2.2 (0.2)	2.3 (0.2)	2.2 (0.2)	0.05 (0.13)	
Braking peak force (% BW)	34.0 (3.3)	33.5 (3.6)	33.7 (3.2)	33.4 (3.4)	0.49 (0.04)	
Propulsive impulse (% BW·s)	2.1 (0.3)	2.1 (0.3)	2.1 (0.2)	2.1 (0.3)	< 0.002* (0.24)	A: p = 0.01* , g = -0.82 [-1.32; -0.30] B: p = 0.06, g = -0.64 [-1.12; -0.15]
Propulsive peak force (% BW)	30.4 (4.0)	30.1 (4.3)	30.5 (4.1)	29.9 (4.4)	0.08 (0.12)	
Duty factor	0.35 (0.03)	0.35 (0.03)	0.35 (0.03)	0.35 (0.03)	0.56 (0.04)	

Legend: A (Intervention A), B: (Intervention B), BW (body weight), CI (confidence interval), COM (center of mass), GLM (repeated measure general linear model), g (Cohen's g effect size), η^2p (partial eta squared effect size of the GLM), * and bold: significant at p<0.05. Post-hoc p-values were corrected using Bonferroni method.

Appendices of Chapter 4: Side-step cutting in CAI, copers and healthy controls

4.A Ankle sprain history, rehabilitation and use of ankle support

Outcome	CAI	Coper	p-value
First ankle sprain			
Duration since it occurred (months) ‡	121.6 (104.3)	120.4 (73.6)	0.523
Immobilization (n)	17 yes / 5 No	11 Yes / 13 No	0.029*
Impact on weight bearing activities (days) ‡	15.2 (17.8)	8.8 (13.1)	0.145
Last ankle sprain			
Duration since it occurred (months) ‡	30.6 (35.9)	99.1 (72.1)	Med. diff. = 68.5 months p < 0.001*
Immobilization (n)	10 Yes / 12 No	9 Yes / 15 No	0.584
Rehabilitation			
Pursued rehabilitation (n)	18 Yes / 4 No	9 Yes / 15 No	0.002*
Duration (weeks) ‡	12.3 (14)	6 (17)	0.294
Duration since rehabilitation started (months) †	56.1 (37.5)	93 (66.2)	0.175
Ankle support used during sport (n)	6 Yes / 16 No	0 Yes / 24 No	0.006*

Legend: Data are presented as proportions or mean (standard deviation). Proportions were compared using Chi-squared tests, while t-test or Kruskal-Wallis tests were used for continuous data. CAI (chronic ankle instability), Med. diff. (median difference), n (number of participants), † (normality not rejected), ‡ (normality rejected), * (statistically significant).

4.B Post-hoc tests

Outcome	ANOVA or Kruskal-Wallis	Post-hoc tests
Weight †	F(69, 2) = 3.42, p = 0.039*	CAI vs coper: MD = 0.38 [-5.73; 6.49], p = 1 CAI vs control: MD = 5.82 [-0.35; 11.99], p = 0.07 Coper vs control: MD = 5.44 [-0.60; 11.48], p = 0.09
CAIT ‡	H(2) = 50.0, p < 0.001*	CAI vs coper: Med. diff. = 13, p < 0.001* CAI vs control: Med. diff. = 13, p < 0.001* Coper vs control: Med. diff. = 0, p = 1
FAAM activities of daily living subscale ‡	H(2) = 43.5, p < 0.001*	CAI vs coper: Med. diff. = 5.4, p < 0.001* CAI vs control: Med. diff. = 5.4, p < 0.001* Coper vs control: Med. diff. = 0, p = 1
FAAM sports subscale ‡	H(2) = 50.8, p < 0.001*	CAI vs coper: Med. diff. = 12.5, p < 0.001* CAI vs control: Med. diff. = 12.5, p < 0.001* Coper vs control: Med. diff. = 0, p = 1
Hallux flexion strength (N/kg) †	F(69,2) = 4.58, p = 0.014*	CAI vs coper: MD = 0.03 [-0.23; 0.27], p = 1 CAI vs control: MD = -0.25 [-0.50; 0], p = 0.050* Coper vs control: MD = -0.28 [-0.52; -0.03], p = 0.023*
Posterior peak GRF force (% of BW) †	F(68,2) = 4.12, p = 0.021*	CAI vs coper: MD = -1.04 [-6.70; 4.62], p = 1 CAI vs control: MD = -6.16 [-11.87; -0.44], p = 0.03* Coper vs control: MD = -5.11 [-10.64; 0.41], p = 0.08

Legend: CAI (chronic ankle instability), CAIT (Cumberland ankle instability tool), F (statistical value of the ANOVA), FAAM (foot and ankle ability measure), GRF (ground reaction force), H (statistical value of the Kruskal-Wallis test), MD (mean difference and 95% confidence interval), Med. diff. (median difference), p (p-value), † (normality not rejected), ‡ (normality rejected), * (statistically significant).

4.C Side-cutting spatio-temporal outcomes

Outcomes	CAI	Coper	Control	p-value
Speed ($\text{m}\cdot\text{s}^{-1}$) †	3.40 (0.34)	3.47 (0.32)	3.49 (0.33)	0.67
Cadence ($\text{steps}\cdot\text{min}^{-1}$) †	167.6 (12.6)	165.4 (14.1)	170.4 (15.7)	0.52
Step time (s) ‡	0.36 (0.03)	0.36 (0.03)	0.36 (0.03)	0.87
Step length (m) †	1.16 (0.13)	1.18 (0.12)	1.17 (0.13)	0.86
Foot strike index (%) ‡	38.9 (16.8)	46.5 (21.1)	39.2 (20.5)	0.17
Foot strike pattern (n)	7 rearfoot / 14 midfoot	5 rearfoot / 19 midfoot	9 rearfoot / 13 midfoot / 1 forefoot	0.38
Foot progression at foot strike, transverse plane (°) ‡	3.7 (7)	5.7 (7.1)	6.8 (9.9)	0.18

Legend: Data are presented as mean and standard deviation, and numbers for categorical data. Foot strike pattern was determined using foot strike index as follows: rearfoot < 34%; midfoot 34-67%; forefoot $\geq$ 68% ²⁹⁷. CAI (chronic ankle instability), † (normality not rejected), ‡ (normality rejected), * (statistically significant).

4.D Ranges of motion and peak moment in the frontal and transverse planes

Outcomes	CAI	Coper	Control	p-value
ANKLE				
Inv/Eve ROM: P 1 (°) ‡	1.5 (1)	0.9 (0.7)	1.4 (1.3)	0.18
Inv/Eve ROM: P 2 (°) ‡	3.6 (1.8)	3.3 (1.9)	3.1 (2)	1
Inv/Eve ROM: P 3 (°) ‡	4.7 (2.4)	4.4 (1.8)	4.4 (1.7)	1
Abd/Add ROM: P 1 (°) ‡	0.8 (0.7)	1 (0.8)	0.9 (1)	1
Abd/Add ROM: P 2 (°) ‡	2.9 (1.6)	2.3 (0.9)	3.8 (2.6)	0.62
Abd/Add ROM: P 3 (°) ‡	4.8 (2.3)	4.3 (1.9)	5.6 (3.1)	0.76
Inv. moment (Nm·kg ⁻¹) ‡	0.23 (0.18)	0.23 (0.16)	0.21 (0.13)	0.91
Eve. moment (Nm·kg ⁻¹) ‡	0.15 (0.13)	0.15 (0.13)	0.17 (0.13)	0.85
Add. moment (Nm·kg ⁻¹) ‡	0.14 (0.05)	0.16 (0.09)	0.14 (0.06)	0.71
Abd. moment (Nm·kg ⁻¹) ‡	0.17 (0.09)	0.15 (0.11)	0.23 (0.22)	0.56
CHOPART				
Inv/Eve ROM: P 1 (°) ‡	2.3 (1.2)	2.1 (1.1)	1.7 (1)	0.58
Inv/Eve ROM: P 2 (°) ‡	3.1 (1.3)	2.7 (1.3)	3.2 (0.9)	1
Inv/Eve ROM: P 3 (°) ‡	7.1 (2.1)	7.1 (1.6)	6.7 (1.9)	1
Abd/Add ROM: P 1 (°) ‡	1.7 (1.1)	1.7 (1.2)	1.5 (1.1)	1
Abd/Add ROM: P 2 (°) ‡	2.6 (0.9)	2.9 (1.7)	3.1 (1.5)	1
Abd/Add ROM: P 3 (°) ‡	7.5 (2.7)	8.6 (2.1)	8.3 (2.6)	0.95
Inv. moment (Nm·kg ⁻¹) ‡	0.05 (0.03)	0.04 (0.03)	0.06 (0.05)	0.76
Eve. moment (Nm·kg ⁻¹) ‡	0.49 (0.2)	0.48 (0.15)	0.5 (0.16)	0.78
Add. moment (Nm·kg ⁻¹) ‡	0.054 (0.027)	0.052 (0.04)	0.052 (0.025)	0.95
Abd. moment (Nm·kg ⁻¹) ‡	0.39 (0.14)	0.37 (0.13)	0.4 (0.2)	0.83
LISFRANC				
Inv/Eve ROM: P 1 (°) ‡	1 (0.7)	0.9 (0.8)	0.9 (0.5)	1
Inv/Eve ROM: P 2 (°) ‡	3.4 (1)	3.1 (0.9)	3.7 (1.9)	0.82
Inv/Eve ROM: P 3 (°) ‡	6.4 (2.3)	6.1 (2.1)	6.9 (2.7)	1
Abd/Add ROM: P 1 (°) ‡	1.2 (0.9)	1 (0.6)	0.9 (0.6)	1
Abd/Add ROM: P 2 (°) ‡	2.9 (1.3)	2.8 (1)	2.9 (1.2)	1
Abd/Add ROM: P 3 (°) ‡	8.4 (2.4)	8.9 (3.2)	8.7 (2.4)	1
Inv. moment (Nm·kg ⁻¹) ‡	0.058 (0.05)	0.051 (0.045)	0.061 (0.063)	0.9
Eve. moment (Nm·kg ⁻¹) ‡	0.15 (0.12)	0.15 (0.13)	0.17 (0.13)	0.75
Add. moment (Nm·kg ⁻¹) ‡	0.04 (0.019)	0.042 (0.024)	0.047 (0.038)	0.99
Abd. moment (Nm·kg ⁻¹) ‡	0.62 (0.17)	0.65 (0.16)	0.63 (0.17)	0.94

Legend: Data are presented as mean and standard deviation. CAI (chronic ankle instability), DF (dorsiflexion), P 1, 2 and 3 (impact peak, active peak and propulsive subphases), PF (Plantarflexion), ROM (range of motion), ‡ (normality not rejected), † (normality rejected), * (statistically significant). P-values related to ROM were corrected to account for the number of subphases, by multiplying the value by 3.

4.E Cross-correlations during side-cutting

Joint couple	CAI	Coper	Control	p
Ankle frontal – Ankle transverse †	0.16 (0.38)	0.13 (0.3)	0.27 (0.42)	0.43
Ankle frontal – Hindfoot-forefoot sagittal †	-0.14 (0.46)	0.06 (0.4)	-0.08 (0.48)	0.33
Shank-rearfoot frontal – Hindfoot-forefoot frontal ‡	0.03 (0.35)	0.08 (0.4)	0.03 (0.41)	0.27
Ankle frontal – Hindfoot-forefoot transverse †	0.28 (0.34)	0.16 (0.31)	0.24 (0.43)	0.52

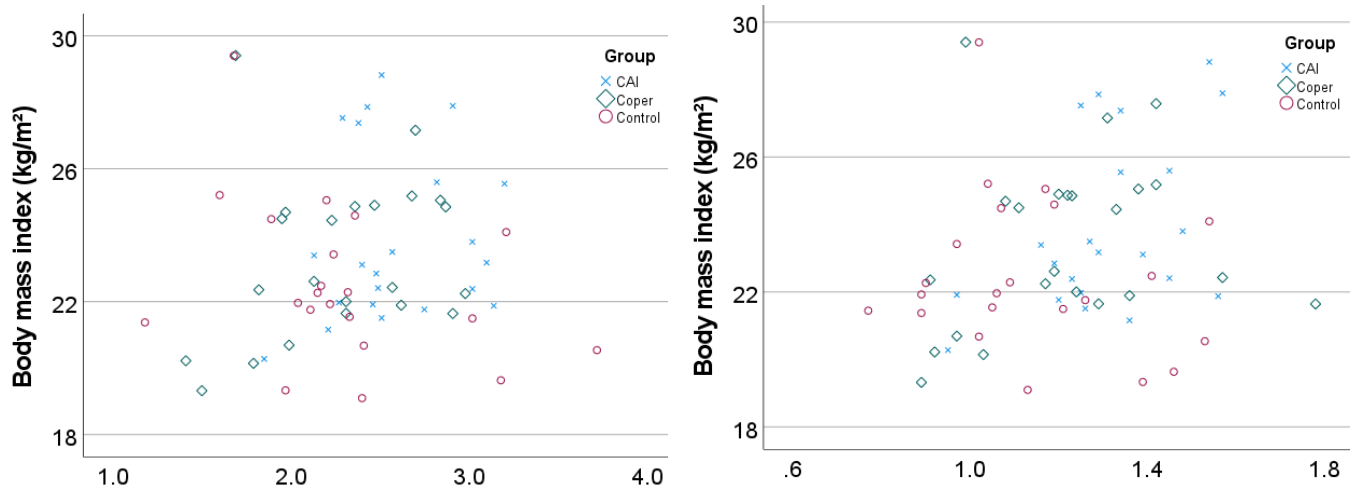
Cross-correlations over the stance phase were calculated between the ankle frontal plane angle and three other foot joint angles.³⁰¹

4.F Foot muscle sizes measured with ultrasound imaging

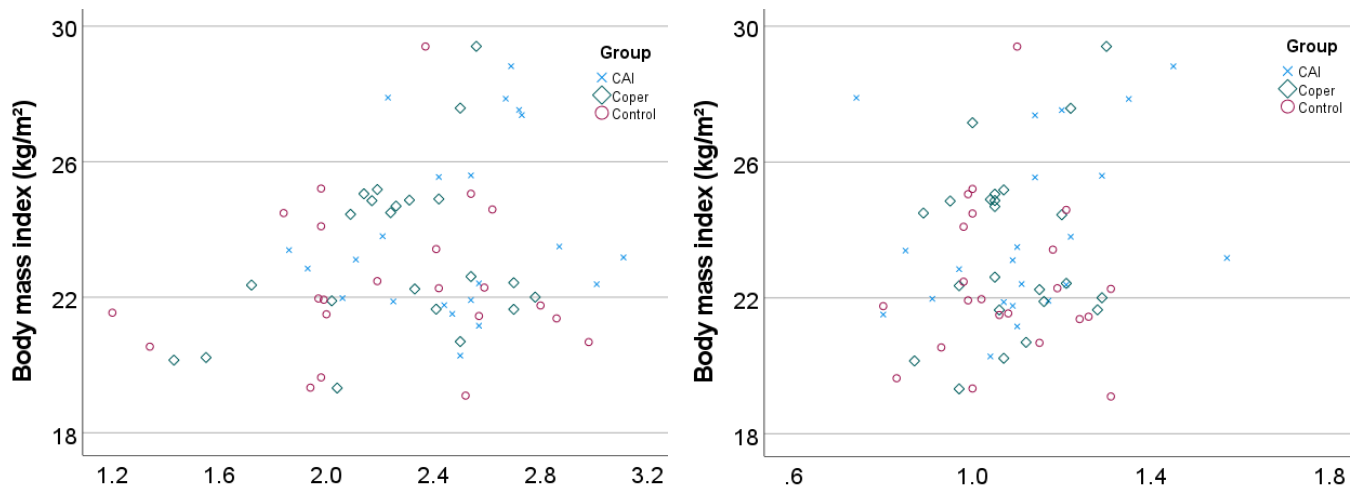
Measure	CAI	Coper	Control
Abductor hallucis cross sectional area (cm ²) †	2.6 (0.4)	2.4 (0.6)	2.3 (0.6)
Abductor hallucis thickness (cm) †	1.3 (0.2)	1.2 (0.2)	1.1 (0.2)
Flexor digitorum cross sectional area (cm ²) †	2.5 (0.3)	2.3 (0.4)	2.2 (0.5)
Flexor digitorum thickness (cm) †	1.1 (0.2)	1.1 (0.1)	1.1 (0.1)

Legend: Data are presented as mean and standard deviation. CAI (chronic ankle instability).

4.G Scatter plots of foot muscle sizes versus body mass index



Scatter plots of abductor hallucis cross-sectional area (cm², left) and thickness (cm, right) versus body mass index (vertical axis).



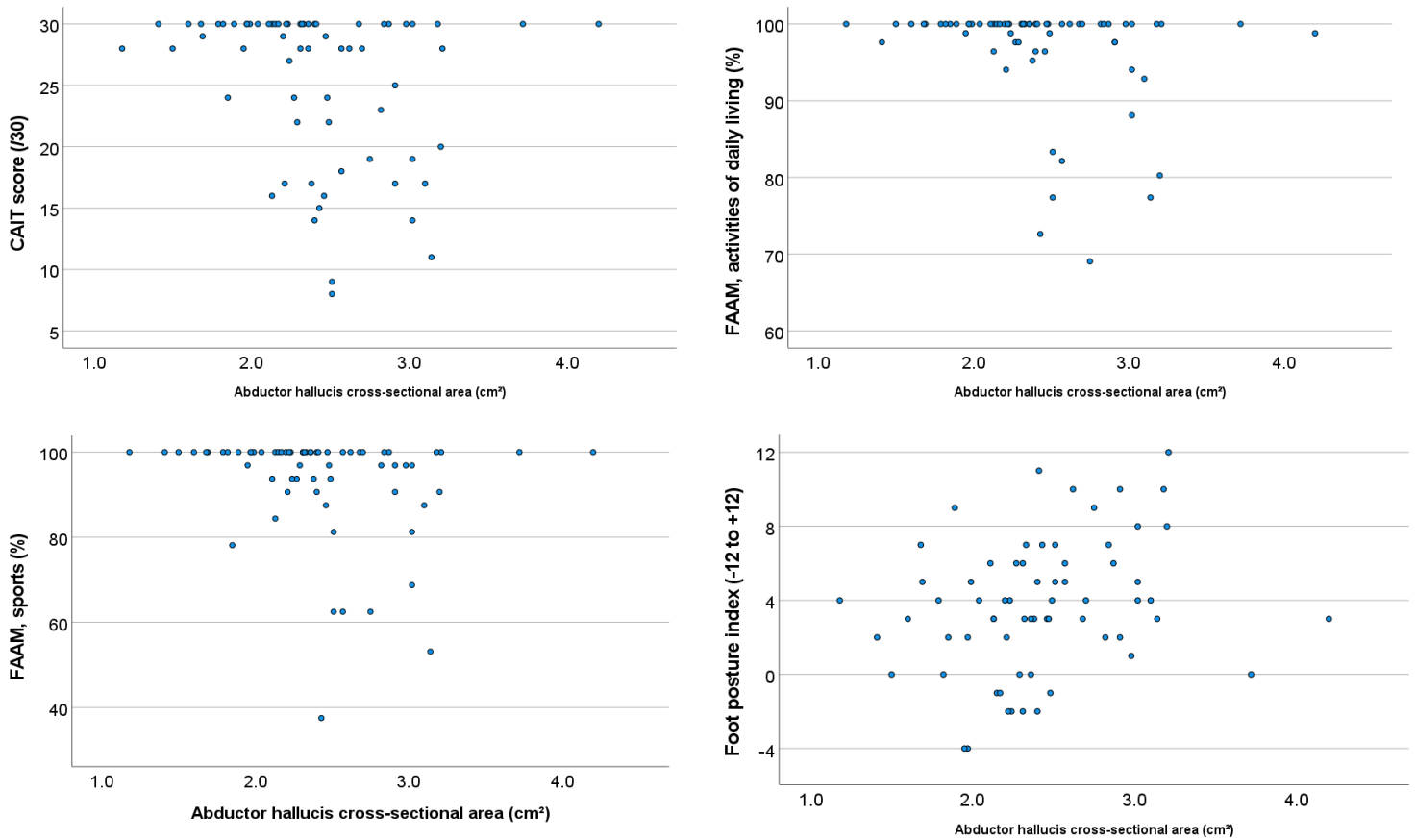
Scatter plots of flexor digitorum brevis cross-sectional area (cm², left) and thickness (cm, right) versus body mass index (vertical axis).

4.H Correlations between intrinsic foot muscle sizes, questionnaires, and the outcomes of the physical evaluation

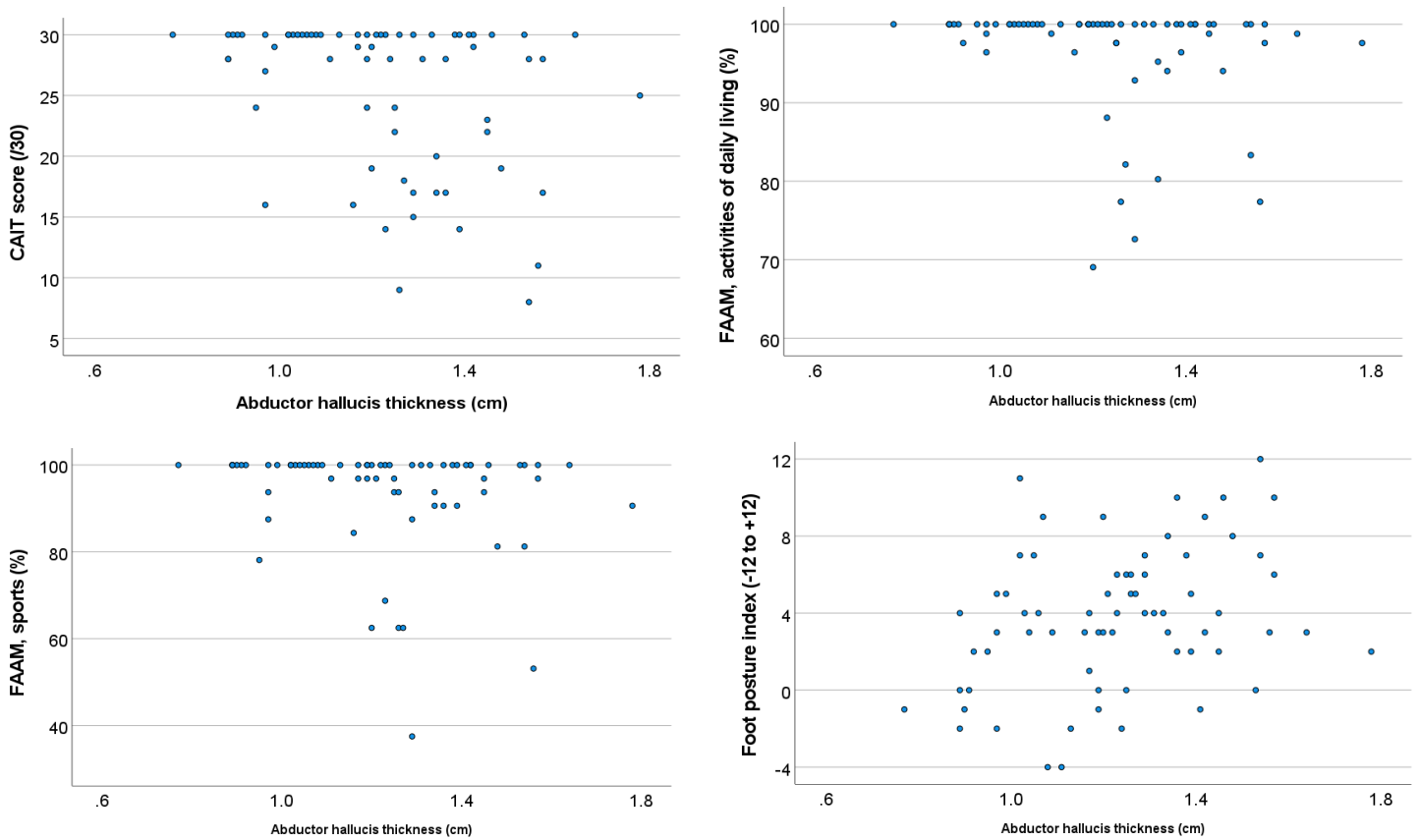
Foot muscle sizes	Stat. values	Questionnaires						Star excursion balance test				Foot posture	Foot mobility	Foot strength	
		CAIT ‡	FAAM ADL ‡	FAAM sports ‡	FABQ PA ‡	FABQ work ‡	FABQ total ‡	Anterior ‡	Postero-medial †	Postero-lateral †	SEBT composite †	FPI-6 score ‡	Navicular drop ‡	Hallux strength †	Toe strength †
Abductor hallucis CSA (n = 66) †	r or p	-0.31	-0.33	-0.33	0.07	0.02	0.07	0.04	0.09	-0.07	0.01	0.32	0.07	-0.01	0.03
	p	0.011*	0.007*	0.006*	0.59	0.86	0.57	0.76	0.48	0.60	0.93	0.009*	0.59	0.96	0.81
Abductor hallucis thickness (n = 68) †	r or p	-0.33	-0.31	-0.26	0.02	0.05	0.03	-0.07	0.06	-0.09	-0.05	0.33	0.09	0.01	-0.07
	p	0.006*	0.010*	0.033*	0.85	0.67	0.82	0.56	0.65	0.49	0.68	0.005*	0.48	0.92	0.55
Flexor digitorum brevis CSA (n = 68) †	r or p	-0.36	-0.27	-0.27	0.04	0.12	0.10	-0.06	-0.03	-0.02	-0.05	0.00	-0.18	0.08	0.17
	p	0.003*	0.024*	0.024*	0.75	0.34	0.44	0.61	0.84	0.89	0.67	0.97	0.14	0.49	0.17
Flexor digotorum brevis thickness (n = 68) †	r or p	-0.16	-0.15	-0.07	0.03	0.21	0.13	-0.17	-0.12	-0.09	-0.18	-0.01	-0.07	-0.21	0.00
	p	0.20	0.21	0.58	0.80	0.09	0.31	0.17	0.33	0.46	0.14	0.94	0.57	0.09	0.97

Legend: Pearson or Spearman correlation coefficients were calculated according to the assumption of normality of the pair of variables. ADL (activities of daily living), CAI (chronic ankle instability), CAIT (Cumberland ankle instability tool), CSA (cross-sectional area), FAAM (foot and ankle ability measure), FABQ (fear avoidance belief questionnaire), FPI-6 (6-items foot posture index), n (number of participants), p (p-value), PA (physical activity), r (Pearson's correlation coefficient), p (Spearman's correlation coefficient), † (normality assumption cannot be rejected), ‡ (normality assumption must be rejected), * (statistically significant).

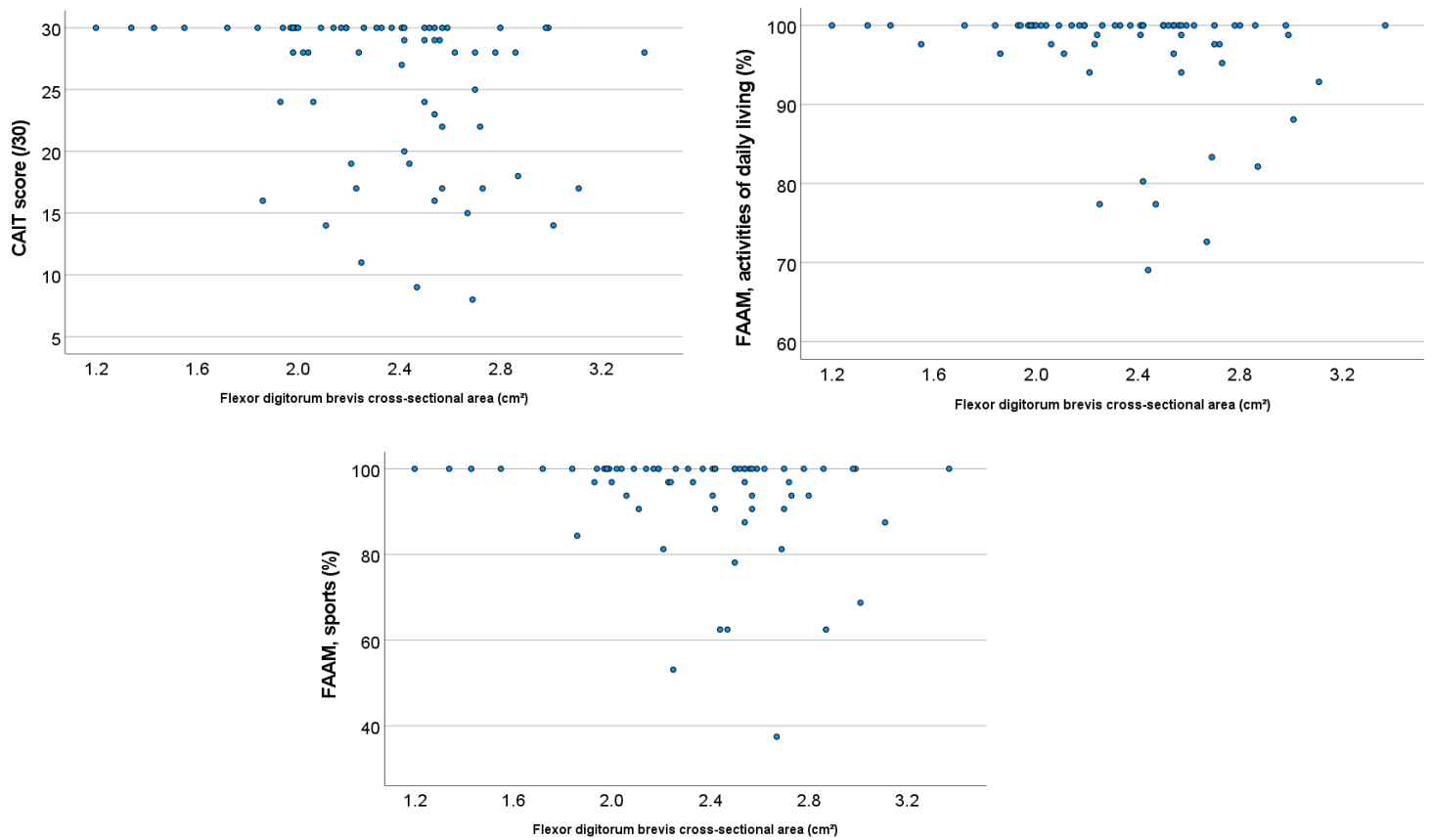
4.1 Scatter plots of foot muscle sizes versus scores of questionnaires and foot posture



Scatter plots of abductor hallucis cross-sectional area (horizontal axis) versus Cumberland ankle instability tool (top left), activities of daily life subscale of the foot and ankle ability measure (top right), sports subscale of the foot and ankle ability measure (inferior left), and Foot-posture index (inferior right).



Scatter plots of abductor hallucis thickness area (horizontal axis) versus Cumberland ankle instability tool (top left), activities of daily life subscale of the foot and ankle ability measure (top right), sports subscale of the foot and ankle ability measure (inferior left), and Foot-posture index (inferior right).



Scatter plots of flexor digitorum cross-sectional area (horizontal axis) versus Cumberland ankle instability tool (top left), activities of daily life subscale of the foot and ankle ability measure (top right), and sports subscale of the foot and ankle ability measure (inferior).

Appendices of Chapter 5: Repeated foot muscle contractions in CAI

5.A Side-step cutting spatiotemporal outcomes

Outcomes	Pre	Post	p-value	Effect size (95% CI)
Speed (m·s ⁻¹) †	3.5 (0.4)	3.3 (0.4)	0.08	0.41 (-0.05 – 0.87)
Cadence (steps·min ⁻¹) †	170.5 (12.0)	165.7 (12.5)	0.04*	0.49 (0.02 – 0.95)
Step time (s) †	0.36 (0.02)	0.37 (0.03)	0.03*	-0.54 (-1.00 – -0.06)
Step length (m) †	1.18 (0.14)	1.13 (0.14)	0.11	0.38 (-0.08 – 0.83)
Foot Strike Index (%) ‡	37.9 [31.5 – 58.4]	41.0 [8.6 – 58.4]	0.68	0.09 (-3.19 – 4.77)
Foot orientation (°)	-18.3 (3.2)	-18.4 (3.3)	0.82	0.05 (-0.38 – 0.48)

Legend: Data are presented as mean (standard deviation) when normality was not rejected (†), and as median [quartile 1 – quartile 3] when normality was rejected (‡). Results of the paired tests are reported as the p-value and the standardized effect size, i.e., Cohen's d or Wilcoxon r, depending on normality. CI (confidence interval), ^a (negative values indicate external rotation), * (significantly different at p < 0.05 level).

5.B Secondary outcomes – including ranges of motion, peaks in moments, power, work, and ground reaction forces (except outcomes reported in Table 5.2)

Variable	Pre	Post	p-value	Effect size (95% CI)
ANKLE				
PF/DF ROM: P1 (°) †	4.5 (1.8)	4.5 (1.8)	1	-0.003 (-0.43 – 0.43)
PF/DF ROM: P2 (°) †	20.2 (2.8)	19.9 (3.0)	1	0.13 (-0.30 – 0.56)
PF/DF ROM: P3 (°) †	39.2 (7.0)	39.2 (6.2)	1	0 (-0.43 – 0.43)
Add./Abd. ROM: P1 (°) †	0.63 (0.46)	0.82 (0.52)	0.43	-0.33 (-0.77 – 0.11)
Add./Abd. ROM: P2 (°) †	3.0 (1.7)	3.1 (1.7)	1	-0.07 (-0.50 – 0.36)
Add./Abd. ROM: P3 (°) †	5.1 (2.5)	4.9 (2.5)	1	0.08 (-0.35 – 0.51)
Peak PF moment (Nm·kg ⁻¹) †	2.5 (0.4)	2.5 (0.3)	0.89	-0.03 (-0.48 – 0.42)
Peak add. moment (Nm·kg ⁻¹) †	0.15 (0.06)	0.12 (0.07)	0.003*	0.77 (0.25 – 1.28)
Peak abd. moment (Nm·kg ⁻¹) †	-0.17 (0.09)	-0.22 (0.12)	0.02*	0.58 (0.08 – 1.06)
Pos. work (J·kg ⁻¹) †	821.7 (209.05)	818.5 (171.7)	0.89	0.03 (-0.42 – 0.48)
Neg. work (J·kg ⁻¹) †	360.9 (146.8)	405.6 (172.4)	0.13	-0.36 (-0.82 – 0.11)
CHOPART				
PF/DF ROM: P1 (°) †	2.9 (2.2)	2.9 (1.8)	1	0.05 (-0.38 – 0.48)
PF/DF ROM: P2 (°) ‡	6.5 [5.7 – 6.7]	6.7 [5.7 – 7.4]	0.67	-0.27 (-1.34 – 0.18)
PF/DF ROM: P3 (°) †	17.5 (3.5)	17.6 (3.4)	1	-0.03 (-0.46 – 0.39)
Inv./Ev. ROM: P1 (°) †	2.4 (1.5)	2.5 (1.3)	1	-0.05 (-0.48 – 0.38)
Inv./Ev. ROM: P2 (°) ‡	3.2 [2.3 – 4.0]	3.0 [2.2 – 4.5]	1	0.04 (-0.89 – 1.00)
Inv./Ev. ROM: P3 (°) ‡	7.4 [5.2 – 8.9]	7.5 [5.6 – 9.6]	1	-0.17 (-0.57 – 0.34)
Add./Abd. ROM: P1 (°) †	1.7 (1.2)	1.7 (1.3)	1	0.06 (-0.37 – 0.49)
Add./Abd. ROM: P2 (°) †	2.6 (0.8)	2.6 (1.1)	1	0.09 (-0.34 – 0.52)
Add./Abd. ROM: P3 (°) †	7.5 (2.9)	7.8 (3.0)	0.89	-0.23 (-0.66 – 0.20)
Peak PF moment (Nm·kg ⁻¹) †	1.8 (0.3)	1.8 (0.3)	0.95	-0.02 (-0.47 – 0.43)
Peak inv. moment (Nm·kg ⁻¹) ‡	0.05 [0.03 – 0.06]	0.04 [0.02 – 0.05]	0.13	0.35 (-0.00 – 0.02)
Peak ev. moment (Nm·kg ⁻¹) †	-0.47 (0.18)	-0.49 (0.17)	0.50	0.16 (-0.30 – 0.61)
Peak add. moment (Nm·kg ⁻¹) †	0.06 (0.03)	0.05 (0.02)	0.17	0.32 (-0.14 – 0.78)
Peak abd. moment (Nm·kg ⁻¹) ‡	-0.38 [-0.46 – -0.27]	-0.36 [-0.53 – -0.31]	0.27	0.26 (-0.03 – 0.08)
Peak gen. power (W·kg ⁻¹) †	4.3 (1.5)	4.1 (1.1)	0.35	0.22 (-0.24 – 0.67)
Peak abs. power (W·kg ⁻¹) †	-2.1 (1.2)	-2.0 (0.8)	0.66	-0.10 (-0.55 – 0.35)
Pos. work (J·kg ⁻¹) †	228.6 (65.5)	223.6 (58.6)	0.69	0.09 (-0.36 – 0.54)
Neg. work (J·kg ⁻¹) †	162.8 (55.7)	161.5 (50.0)	0.89	0.03 (-0.42 – 0.48)

5.B (Continued)

LISFRANC

PF/DF ROM: P1 (°) †	1.26 (0.95)	1.28 (0.81)	1	-0.01 (-0.44 – 0.41)
PF/DF ROM: P2 (°) †	2.6 (1.7)	1.8 (1.0)	0.02*	0.68 (0.20 – 1.15)
PF/DF ROM: P3 (°) †	8.0 (3.3)	8.4 (3.6)	1	-0.21 (-0.64 – 0.23)
Inv./Ev. ROM: P1 (°) †	0.90 (0.56)	1 (0.53)	1	-0.13 (-0.56 – 0.30)
Inv./Ev. ROM: P2 (°) †	3.5 (1.1)	3.4 (1.2)	1	0.12 (-0.31 – 0.54)
Inv./Ev. ROM: P3 (°) †	7.0 (2.5)	7.0 (2.5)	1	0.03 (-0.40 – 0.45)
Add./Abd. ROM: P1 (°) †	1.0 (0.7)	1.3 (0.8)	0.56	-0.30 (-0.73 – 0.14)
Add./Abd. ROM: P2 (°) †	2.8 (1.2)	2.5 (0.8)	0.47	0.32 (-0.12 – 0.76)
Add./Abd. ROM: P3 (°) †	8.5 (2.4)	8.1 (1.9)	0.93	0.23 (-0.21 – 0.66)
Peak PF moment (Nm·kg ⁻¹) †	0.77 (0.21)	0.77 (0.17)	0.73	0.08 (-0.37 – 0.53)
Peak inv. moment (Nm·kg ⁻¹) †	0.05 (0.04)	0.06 (0.06)	0.55	-0.14 (-0.59 – 0.31)
Peak ev. moment (Nm·kg ⁻¹) †	-0.14 (0.11)	-0.14 (0.12)	0.94	0.02 (-0.43 – 0.47)
Peak add. moment (Nm·kg ⁻¹) ‡	0.04 [0.03 – 0.06]	0.03 [0.02 – 0.05]	0.17	0.32 (-0.01 – 0.02)
Peak abd. moment (Nm·kg ⁻¹) †	-0.61 (0.17)	-0.63 (0.16)	0.42	0.19 (-0.27 – 0.64)
Peak gen. power (W·kg ⁻¹) †	2.1 (1.2)	2.1 (1.0)	0.52	0.15 (-0.30 – 0.60)
Peak abs. power (W·kg ⁻¹) ‡	-0.30 [-0.40 – -0.24]	-0.38 [-0.44 – -0.19]	0.59	-0.13 (-0.15 – 0.08)
Pos. work (J·kg ⁻¹) †	133.7 (63.7)	132.9 (56.9)	0.89	0.03 (-0.42 – 0.48)
Neg. work (J·kg ⁻¹) †	23.7 (15.0)	22.6 (11.6)	0.71	0.09 (-0.36 – 0.54)

HALLUX

PF/DF ROM: P1 (°) †	6.4 (5.4)	7.3 (4.2)	0.66	-0.28 (-0.71 – 0.16)
PF/DF ROM: P2 (°) †	6.9 (3.1)	6.6 (3.7)	1	0.09 (-0.34 – 0.52)
PF/DF ROM: P3 (°) †	40.2 (7.3)	42.8 (8.0)	0.02*	-0.68 (-1.14 – -0.19)
Peak PF moment (Nm·kg ⁻¹) †	0.24 (0.08)	0.23 (0.07)	0.46	0.17 (-0.28 – 0.62)
Peak gen. power (W·kg ⁻¹) †	0.32 (0.20)	0.41 (0.29)	0.07	-0.44 (-0.91 – 0.03)
Peak abs. power (W·kg ⁻¹) †	-2.4 (1.2)	-2.3 (1.0)	0.50	-0.16 (-0.61 – 0.30)
Pos. work (J·kg ⁻¹) ‡	13.4 [8.8 – 14.6]	13.5 [8.6 – 17.8]	0.10	-0.38 (-3.98 – 0.91)
Neg. work (J·kg ⁻¹) †	137.5 (64.0)	137.2 (52.8)	0.96	0.01 (-0.44 – 0.46)

PEAKS GRFs

Vertical (% body weight) †	210.6 (21.8)	209.3 (20.4)	0.60	0.12 (-0.31 – 0.54)
Medial (% body weight) ‡	51.6 [46.2 – 55.5]	50.7 [45.6 – 56.1]	0.47	-0.16 (-4.28 – 1.64)
Posterior (% body weight) ‡	30.0 (5.6)	32.6 (5.0)	0.03*	-0.52 (-0.98 – -0.06)
Anterior (% body weight) †	17.5 [14.1 – 18.7]	16.9 [14.3 – 18.9]	0.16	-0.31 (-1.58 – 0.39)

Legend: Data are presented as mean (standard deviation) when normality was not rejected (†) – and as median [quartile 1 – quartile 3] when normality was rejected (‡). Results of the paired tests are reported as the p-value and the standardized effect size – i.e. – Cohen's d or Wilcoxon r – depending on normality. Abd. (abduction) – abs. (absorption) – add. (adduction) – CI (confidence interval) – DF (dorsiflexion) – ev. (eversion) – gen. (generation) – inv. (inversion) – neg. (negative) – P1 (impact peak phase) – P2 (active peak phase) – P3 (propulsive phase) – PF (plantarflexion) – pos. (positive) – ROM (range of motion).