Proof-of-concept of a 16-week foot muscle specific intervention programme on non-contact anterior cruciate ligament and lateral ankle sprain injury risk

A thesis presented in partial fulfilment of the requirements for the degree of

Doctor of Philosophy

in

Sport & Exercise Science

at Massey University, Manawatu

Palmerston North, New Zealand

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2020

ABSTRACT

During high intensity sports where abrupt decelerations and unanticipated changes of direction are prevalent, impaired foot function is thought to play a role in increasing the risk for non-contact anterior cruciate ligament rupture (ACLR) and lateral ankle sprain (LAS) injury. The risk for sustaining an ACLR is linked to increased rearfoot eversion and excessive, dynamic subtalar joint pronation coupled with internal rotation of the shank, leading to a larger knee valgus angle, and increasing anterior cruciate ligament (ACL) strain under high loads. Impaired forefoot stability is linked to larger moment arm lengths around the ankle joint, increasing the risk for lateral ankle sprain (LAS) under high loads. Previous research has shown that dynamic foot function can influence ankle, knee, and hip movement. Increasing forefoot and medial longitudinal arch stiffness modulates the distal-to-proximal transfer of rotational movement in straight-line walking and running as well as low intensity change of direction tasks. Foot stiffness is created by both passive and active structure in the foot. Passive structures include the foot bones and -ligaments and work in conjunction with the plantar fascia to create stiffness and stability. The intrinsic and extrinsic muscles acting on the foot act in synergy to create stiffness actively. However, current prophylactic programmes do not aim to explicitly train the foot muscles. The aim of this project was thus to provide proof of concept for a foot muscle specific training intervention on dynamic foot function, as well as risk factors associated with ACLR and LAS injury. Establishing proof of concept provides valuable information for future large-scale randomised control studies investigating the integration of a foot muscle specific intervention in to ACLR and LAS injury prevention programmes.

The overview of the functional anatomy of the foot in **Chapter 2** described the bones that form the joints of the foot as well as the muscles that move the bones of the foot. The interaction of the segments of the foot is described as well as the way in which the foot is modelled. The chapter also provided the contextual evidence for exercise selection. **Chapter 4** reviewed the literature and highlighted the relationship between dynamic, excessive subtalar joint pronation and rearfoot eversion to internal tibial rotation, the increase in the knee valgus angle and ACL strain. A link between instability of the forefoot and a larger hallux extension range of motion increasing risk for LAS is discussed. The review also discussed the effect of intrinsic and extrinsic foot muscles on movement of the foot segments and general foot function. A hypothesis was formed that a foot muscle specific intervention has the potential to influence dynamic foot function and risk factors associated with ACLR and LAS, supplementing current prophylactic programmes.

Current literature does not definitively describe the coupling relationship between the segments of the lower limb nor the coupling between the segments of the foot during high-intensity unanticipated change of direction tasks. It is thus unknown whether training the foot muscles in an attempt to modulate the transfer of rotational forces from distal-to-proximal segments to decrease injury risk, is justified. A modified vector coding technique was adapted to describe and quantify the coordination patterns between the lower limb and foot segments, respectively. In **Chapter 5** the literature regarding the modifications to the vector coding technique and its development from the continues relative phase method of describing movement relationship between objects was presented.

The coupling relationship between the tri-planar calcaneus and transverse plane rotation of the shank during unanticipated change of direction tasks was described in **Chapter 6.** A distal-to-proximal coupling relationship between frontal- and transverse plane movement of the calcaneus and transverse plane rotation of the shank was established. The distal-proximal coupling between the calcaneus and shank suggested that shank rotation may be modulated by manipulating the frontal and transverse plane movement of the calcaneus. It thus seems worthwhile to investigate whether training the muscles that act on the foot would manipulate shank rotations and, in this way, decrease the strain on the ACL, reducing ACLR risk.

In **Chapter 7** the coupling relationship between the foot segments was described as it influences LAS risk. The forefoot was the only point of contact during unanticipated change of direction tasks. Throughout stance, an anti-phase coupling relationship in the form of metatarsal flexion coupled with calcaneus eversion, and metatarsal extension was coupled with calcaneus inversion. These coupling relationships indicated the importance of forefoot function in calcaneus control and potentially LAS risk. In the loading phase, metatarsal inversion was coupled with calcaneus inversion, potentially increasing LAS risk if metatarsal inversion was not limited. At maximum calcaneus inversion- and adduction velocities, both of which are associated with increased risk for LAS, hallux and metatarsal flexion accelerated. The increased forefoot flexion acceleration indicating the importance of the forefoot stiffness in creating forefoot stability, potentially influencing whole foot stiffness and lateral ankle stability during change of direction tasks. As the forefoot provides stiffness the foot is likely to be more stable and the rearfoot segments are free to move and align with the shank, potentially decreasing LAS risk. It seems thus important to investigate whether training the foot muscles will influence LAS risk.

The 16-week progressive foot muscle specific intervention program was detailed in **Chapter 8**. Exercise components, selection and progression was described. The outcomes of the proof-of-concept study revealed that training the muscles acting on the foot show potential to resist the deformation of the medial longitudinal arch (MLA). Resistance to MLA deformation potentially played a role in the observed decrease of the maximum knee valgus angle, and ACLR risk factor, of the training group (TG). LAS injury risk factors, maximum ankle inversion angle and maximum ankle eversion moment arm length were also smaller for the group undergoing the intervention training. The decrease in the maximums of these LAS risk factors was potentially influenced by the increasing stiffness of the metatarsal anterior transverse arch (MetATA), which is likely to increase foot and ankle stability.

Guidelines for future randomised controlled trials included variables to consider ensuring the groups are well matched before the intervention. A proposal regarding the methodology and implementation of an injury prevention programme as well as sample size recommendations were outlined. To ensure the groups are well matched before the intervention it would be prudent to match athletes for stance times, peak ground reaction forces (GRF) in addition to sport and body mass index (BMI).

During data collection, the approach speed and thus the timing of the unanticipated change of direction task proved difficult to monitor and control, which potentially influenced the kinetic and kinematic outcomes of the pilot study. The TG also had low compliance to training which could have influenced the result of the intervention. Sample size calculations revealed that a sample size of 30 (falling within the recommended size for biomechanical studies) are needed to find significant differences in foot arch and foot segment angle variables. However, a larger number of athletes are needed per group to establish changes to ACLR and LAS risk variables.

Functional anatomy and the coupling relationship between the segments of the lower limb and foot segments justified investigating the effect of foot muscle specific intervention on ACLR and LAS risk factors. The pilot study revealed that training the muscles acting on the foot seem to influence foot function, but the effect on risk factors associated with ACLR and LAS is unclear. Following the guidelines as described in future randomised control trial is necessary to establish the effectiveness of integrating foot muscle specific exercises into current ACLR and LAS prophylactic programmes.

Student Declaration

I hereby declare that this thesis is my own work and does not, to the best of my knowledge, contain material from any other source unless due acknowledgement is made. This thesis was completed under the guidelines set by Massey University's College of Health, for the degree of Doctor of Philosophy and has not been submitted for a degree of diploma at any other academic institution.

Candidate: ______________________________

Date: __________________________________

ACKNOWLEDGEMENTS

I would like to thank all the coaches, athletes and parents who volunteered their time to this project. To the coaches for granting me access to their athletes, the parents for transporting their children to, sometimes very early morning, testing sessions. The athletes who diligently attended the supervised training sessions and committed to performing the exercises unsupervised.

I would also like to extend my gratitude to;

The staff at the large animal teaching unit for the use of the facility and access to the force plate. Research assistants, Anna Coppelmans and Samantha Lockett, for your hard work with data collection and processing.

The staff and fellow Sport and Exercise science postgraduate students at Massey University, Palmerston North campus for your encouragement and sound boarding. I wish to furthermore thank my main supervisor, Dr Philip Fink, and co-supervisors Dr Sarah Shultz and Dr Robert Colborne for their expert advice, guidance and mostly patience during the project. Kim Hébert-Losier for her expert advice and insight on data capture and analysis.

Asics NZ for the generous discount on shoes.

Badminton World Federation who provided financial support for this project.

The numerous people I discussed the project with, from the sceptics to the supportive and excited, you all helped to form a more structured, workable concept.

My biggest gratitude is to my longsuffering husband, Chris. I will (still) follow you to the ends of the earth and back.

Publications & Presentations

Publications

Carla van der Merwe, Sarah P. Shultz, G Robert Colborne, Philip W Fink (2019). Foot Muscle Strengthening and Lower Limb Injury Prevention. (IN PRESS).

Carla van der Merwe, Sarah P. Shultz, G Robert Colborne, Kim Hébert-Losier, Philip W Fink (2019). Describing the shank-calcaneus coupling relationship during high-risk change of direction movement: implications for ACL injury prevention strategies. (IN REVIEW).

Carla van der Merwe, Sarah P. Shultz, G Robert Colborne, Kim Hébert-Losier, Philip W Fink (2019). The coordination patterns of the foot segments in relation to lateral ankle sprain injury mechanism during unanticipated changes of direction. (IN REVIEW).

PRESENTATIONS

Carla van der Merwe, Sarah P. Shultz, G Robert Colborne, Philip W Fink (2018). Tibia/Calcaneus coordination patterns during unanticipated change of direction. 11th Australasian Biomechanics Conference, Auckland, New Zealand, 3 - 5 December

Carla van der Merwe, Sarah P. Shultz, G Robert Colborne, Kim Hébert-Losier, Philip W Fink (2019). The effect of a 16-week foot muscle specific intervention programme on ACL and LAS injury mechanisms. International Society of Biomechanics, Calgary, Canada, 31 July – 4 August

Carla van der Merwe, Sarah P. Shultz, G Robert Colborne, Kim Hébert-Losier, Philip W Fink (2019). The effect of a 16-week foot muscle specific intervention programme on non-contact anterior cruciate ligament (ACL) and lateral ankle sprain (LAS) injury risk. Sport and Exercise Science New Zealand, Palmerston North, New Zealand, 27 – 29 November

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LIST OF ABBREVIATIONS

Δvelocity - Change in velocity

A

ABd - Abduction
ACL - Anterior cruciate ligament
ACLR - Anterior cruciate ligament rupture
ADd - Adduction
ANOVA - Analysis of variance
ATFL - Anterior talofibular ligament

B

BW - Body weight

C

CalcaneusX - Calcaneus rotations in the sagittal plane around the X-axis
CalcaneusY - Calcaneus rotations in the frontal plane around the Y-axis
CalcaneusZ - Calcaneus rotations in the transverse plane around the Z-axis
CalShk - Calcaneus-shank
CFL - Calcaneofibular ligament
CRP – Continuous relative phase

D

D - Distal segment

E

ER - External rotation
EVE - Eversion
EXT - Extension

F

FLX – Flexion

G

GIF - Graphics Interchange Format animation
GRF - Ground reaction force
GRFy - Anterior-posterior ground reaction force
GRFz - Vertical ground reaction force

H

HalMet - Hallux-metatarsal

I

iFS - Initial foot strike
INV - Inversion
IR - Internal rotation

L

LAS - Lateral ankle sprain

M

MetATA - Metatarsal anterior transverse arch
MetMid - Metatarsal-midfoot
MidCal - Midfoot-calcaneus
MLA - Medial longitudinal arch

P

P - Proximal segment
PeakGRFy - Peak propulsion force
PTFL - Posterior talofibular ligament

R

R/L and L/R - Indicating that the left and right foot are executing alternate exercises
R^2 - Coefficient of determination
Rs - Repetitions

S

SD - Standard deviation
SF- Short foot exercise
ShankZ - Shank rotations in the transverse plane around the Z-axis
Ss - Sets

T

TO - Toe off
TstGRFy - Transient braking force

V

VAR – Varus

W

wk – week

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CHAPTER 1
INTRODUCTION

The foot is an intricate structure with complex functions (Gwani, Asari, & Mohd Ismail, 2017; Holowka, O'Neill, Thompson, & Demes, 2017; McKeon, Hertel, Bramble, & Davis, 2014; Williams III, Tierney, & Butler, 2014). The configuration of the foot and ankle joints, as well as the properties of connective tissue surrounding the joints, affects coupling and transfer of rotational movement between the segments (Hicks, 1953; Huson, 1991). Previous research has shown that rearfoot eversion and subtalar joint pronation is linked to increased internal tibial rotation, larger knee valgus angles and increased risk of a non-contact anterior cruciate ligament rupture (ACLR) (Beckett, Massie, Bowers, & Stoll, 1992; Hertel, Dorfman, & Braham, 2004; Loudon, Jenkins, & Loudon, 1996; Woodford-Rogers, Cyphert, & Denegar, 1994). Lateral ankle sprain (LAS) injury risk increases when the calcaneus inverts and adducts in the presence of a large ground reaction force (GRF), increasing the external moment arm around the joint and increasing inversion and/or adduction velocity (Fong, Ha, Mok, Chan, & Chan, 2012; Fong et al., 2009; Gehring, Wissler, Mornieux, & Gollhofer, 2013; Kristianslund, Bahr, & Krosshaug, 2011; Panagiotakis, Mok, Fong, & Bull, 2017; Raina & Nuhmani, 2014). It is thus not surprising that forefoot instability (Fong et al., 2009) and increased hallux extension range of motion (Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005; Willems, Witvrouw, Delbaere, Philippaerts, et al., 2005) are linked to increased risk for LAS injury.

Research in straight-line walking and running and low intensity change of direction tasks have shown that the configuration of the foot and ankle joints guide the transfer of motion to the ankle and knee respectively (Dubbeldam, Nester, Nene, Hermens, & Buurke, 2013; Fischer, Willwacher, Hamill, & Brüggemann, 2017; Nester, Jarvis, Jones, Bowden, & Liu, 2014; Nigg, Baltich, Federolf, Manz, & Nigg, 2017; Pohl, Messenger, & Buckley, 2007). Furthermore, research has shown that activating the muscles of the foot can indeed be used to manipulate foot segments (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014) and that a foot muscle specific training programme can influence foot function (Mickle, Caputi, Potter, & Steele, 2016; Mulligan & Cook, 2013). Previous research has also postulated that strengthening the foot muscles will increase foot and ankle stability, potentially decreasing lower limb injury risk (De Villiers, 2014; Maulder, 2011; Nigg et al., 2017). However, current ACLR and LAS injury prevention programmes do not attempt to influence foot function directly (McCriskin, Cameron, Orr, & Waterman, 2015; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Shultz et al., 2015; Vriend, Gouttebarge, van Mechelen, & Verhagen, 2016).

Mainstream ACLR and LAS prophylactic programmes are successful (McCriskin et al., 2015; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Shultz et al., 2015; Vriend et al., 2016). However, the incidence of ACLR and LAS injury are still prevalent in female court sport athletes (Fong, Hong, Chan, Yung, & Chan, 2007; Joseph et al., 2013; Shultz & Schmitz, 2018). Supplementing current, mainstream ACLR and LAS injury prevention programmes by including foot muscle specific exercises, the bottom-up approach (Nigg et al., 2017), may reduce the prevalence of ACLR and LAS in females competing in court sports. However, a distal to proximal coupling relationship of the lower limb has not been established during high-intensity unanticipated change of direction tasks. It is thus unknown whether changing foot function would have an impact on ACLR and LAS injury risk factors. Investigating the role of the foot in high-intensity unanticipated change of direction tasks would best be assessed by means of biomechanical methods. Biomechanical research typically entails motion capture and force platform(s) to collect kinetic and kinematic data from a multi-segmental model during complex movement. As such, motion capture research projects are expensive and time consuming (Mullineaux & Wheat, 2017). Furthermore, large-scale randomised control intervention studies are equally complex. For this reason, a smaller-scale pilot study needs to be undertaken to determine the feasibility of a foot muscle specific intervention programme on the risk factors associated with ACLR and LAS injury before implementing such a programme (Thabane et al., 2010).

The aim of this project is thus to firstly investigate the functional anatomy of the muscles acting on the foot. **Chapter 2** provides an overview of the location and function of the muscles involved in creating stiffness and moving foot segments. A review of the literature in **Chapter 4** provides a brief overview of the risk factors associated with ACLR and LAS, but mainly focuses on forming a theory regarding the feasibility of a foot muscle specific intervention strategy to supplement current mainstream ACLR and LAS prophylactic programmes. **Chapter 5** outlines the method used in Chapter 6 and 7 to describe the coupling relationship between segments. The chapter chronicles continuous relative phase and how the method and principles was adapted to a vector coding technique. The adapted vector coding technique enables biomechanics researchers to describe and visualise the relationship between segments. **Chapter 6** describes the coupling relationship between the tri-planar movement of the calcaneus and the transverse plane rotations of the shank, as part of a larger discussion concerning the hypothesis of training the foot muscles to modulate the transfer of movement to the knee. A distal-proximal coupling relationship would suggest that manipulation of the calcaneus may be able to modulate rotational movement to the knee (ACLR risk). **Chapter 7** describes the coupling relationship between the sagittal plane movement of the hallux and the frontal and transverse movement of the calcaneus. The coupling between the tri-planar movement of the metatarsal segment and frontal and transverse movement of the calcaneus, respectively, is also described. The coupling of the foot segment investigates the hypothesis that function of the forefoot can influence calcaneal movement (LAS risk) and thus training the muscles acting on the forefoot could influence calcaneal movement.

The data that was used to describe the coordination patterns in Chapters 6 and 7 was captured during the pre-intervention testing session. **Chapter 8** describes how the 16-week progressive foot muscle specific intervention programme was implemented as a pilot study in the same cohort of female court sport athletes to establish proof of concept. The results of the proof-of-concept study and guidelines for future studies of a similar design is detailed here.

Financial support for the study was obtained via a grant awarded by the Badminton World Federation (BWF). The report generated for BWF is attached as

Appendix D and looks at the effect of the intervention on barefoot versus shod unanticipated change of direction tasks**.**

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CHAPTER 2
FUNCTIONAL ANATOMY

2.1. FOOT FUNCTION

The foot is a complex, multisegmental structure consisting of 33 joints which are supported by joint capsules, ligaments, and a neuromuscular system (Agur & Ming, 1991). Movement of one segment can influence movement of not only the adjacent segments, but also segments proximal to the foot (Delgado et al., 2013; Nigg, Baltich, Federolf, Manz, & Nigg, 2017). Stability of the joints of the ankle and foot are dictated by the assimilation of the articulation surfaces under load, static restraints of the ligaments, and dynamic support from the musculoskeletal system (Hertel, 2002). Therefore, it is important to first understand the anatomy of the shank, ankle, rearfoot, midfoot, and forefoot segments prior to understanding foot function and its role in ACL and LAS injury mechanisms.

2.2. Anatomy and kinematics of the joints of the ankle and foot

2.2.1 Talocrural joint

The ankle or talocrural joint (also known as the tibiotalar joint) is a strong mortise and tenon articulation between the proximal dome of the talus and the distal ends of the tibia and fibula (Figure 2.2-1). As a result of the uniquely shaped mortise and tenon, the talocrural is most stable in dorsiflexion. The talocrural joint is especially well equipped to resist ankle frontal plane eversion in the dorsiflexed position (Brockett & Chapman, 2016). The talocrural joint is only supported by groups of ligaments; it receives no additional support

from tendons as the muscles crossing the talocrural joint have no direct attachment to the talus. The joint is medially supported by the medial collateral ligaments and is responsible for stabilising the joint during eversion and limiting valgus stresses. Laterally, the ankle joint is supported by the anterior talofibular ligament, posterior talofibular ligament and calcaneofibular ligament. The anterior talofibular is the most frequently injured of all the lateral ligaments and is the most vulnerable during ankle plantarflexion. Both the anterior talofibular ligament and posterior talofibular ligament are responsible for stability of the joint during inversion, reducing varus stress of the ankle and limiting axial rotations of the tibia.

FIGURE 2.2-1: The Talocrural joint. Images courtesy of visible body (Body, 2019).

2.2.2 SUBTALAR JOINT

The subtalar joint allows movement between the talus and the calcaneus. The concave surface of the talus articulates with the convex surface of the calcaneus. The subtalar joint is primarily supported by the talocalcaneal ligament stretching from the inferior surface of the talus to the superior surface of the calcaneus. In the majority of individuals, the subtalar joint axis runs through the centre of the head of the talus with the axis tilted at 41° from horizontal in the sagittal plane and deviating 23° anterior-medially in the anterior-posterior plane. The oblique axis of the subtalar joint allows for triplanar extension-eversion-abduction (pronation) and flexion-inversion-adduction (supination) when the joint is weight bearing (Jastifer & Gustafson, 2014).

FIGURE 2.2-2: The Subtalar joint. Images courtesy of visible body (Body, 2019).

2.2.3 MIDTARSAL JOINT

The midtarsal joint, also referred to as the transverse tarsal or Chopart's joint, is formed by the two different joint complexes. The joint is formed by the articulations between the talus and the navicular (talonavicular) (Figure 2.2-3) as well as the anterior calcaneus and posterior cuboid (calcaneocuboid) (Figure 2.2-4). The two joints are thought to function as a single unit around an axis, similar to the axis of the subtalar joint. The primary function of the midtarsal joint is to allow pronation and supination motion of the forefoot on the rearfoot (Tweed, Campbell, Thompson, & Curran, 2008). The midtarsal joint is supported by the long plantar ligament, assisting in stabilising the lateral longitudinal arch, which will be discussed later. The midtarsal joint is further supported by the transverse tarsal ligaments interlocking the tarsals and causing movement between tarsals to be interdependent.

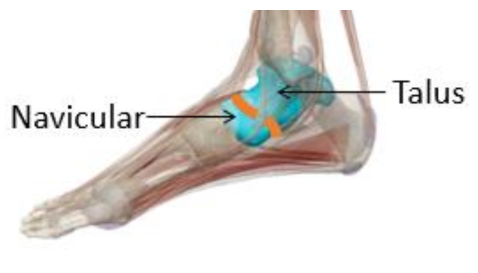

FIGURE 2.2-3: Midtarsal (Talonavicular) joint - Medial view between talus and navicular articulation. Images courtesy of visible body (Body, 2019)

FIGURE 2.2-4: Midtarsal (Calcaneocuboid) joint -Lateral view between calcaneus and cuboid articulation. Images courtesy of visible body (Body, 2019).

2.2.4 TARSOMETATARSAL JOINT COMPLEX

The tarsometatarsal joint complex is situated between the posterior articulation surfaces of the five metatarsals and the anterior aspects of the three cuneiforms as well as the laterally situated cuboid (FIGURE 2.2-5: Tarsometatarsal joint complex). The tarsometatarsal joint can further be divided into a medial column, (first cuneiform and first metatarsal), middle column (second & third cuneiform and second & third metatarsals), and a lateral column (lateral cuboid and fourth and fifth metatarsals). The joint complex is supported by a ligamentous system. In addition, several intrinsic foot muscles cross the joint complex, and some extrinsic foot muscles insert on the tarsals and/or metatarsals.

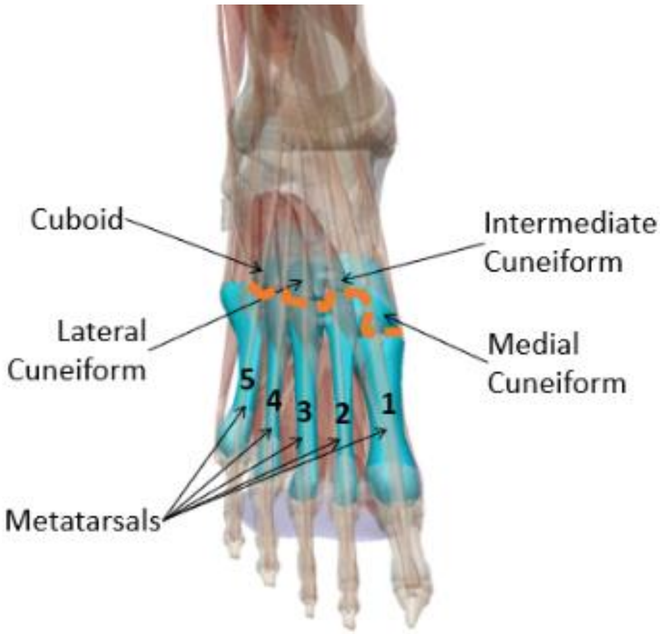

FIGURE 2.2-5: Tarsometatarsal joint complex. Images courtesy of visible body (Body, 2019).

2.2.5 First metatarsophalangeal joint

The first metatarsophalangeal joint is a shallow biaxial ball and socket joint created between the convex head of the first metatarsal and the concave base of the first proximal phalanx (Figure2.2-6). Biaxial ball and socket joints allow multidirectional movement, flexion, extension, abduction, adduction, and circumduction are possible at the metatarsophalangeal joint (Al-Jabri & Charalambides, 2019). The joint is supported by the metatarsophalangeal articular capsule, ligaments and both intrinsic and extrinsic muscles that cross the joint and attach to the first proximal phalanx (Frey, Andersen, & Feder, 1996).

FIGURE 2.2-6: First metatarsophalangeal joint. Images courtesy of visible body (Body, 2019).

2.3. Extrinsic and Intrinsic Muscles

The foot and lower limb are controlled and supported by extrinsic muscles with origin attachment surfaces on the shank and insertion surfaces on the anatomical structures of the foot. Intrinsic muscles originating and inserting within the foot act on the tarsals, metatarsals, and phalanges of the foot. Intrinsic muscles have smaller moment arms and mainly provide immediate sensory input to perturbation and stabilising muscle contractions by means of the stretch-reflex mechanism (McKeon, Hertel, Bramble, & Davis, 2014; Nigg et al., 2017). The extrinsic muscles are slower to respond to perturbations but have larger and more effective moment arms to resist extreme ranges of motion (Jastifer & Gustafson, 2014; Klein, Mattys, & Rooze, 1996). The extrinsic muscles are compartmentalised into anterior, posterior, and lateral sections. The intrinsic muscles are grouped according to their position on either the dorsal or plantar surfaces of the foot.

2.3.1 Extrinsic muscles

2.3.1.1 Anterior compartment

The tibia provides articulation surfaces for the extrinsic muscle tibialis anterior and together with the fibula, an attachment for the extensor digitorum longus. The fibula serves as the single origin for the extensor hallucis longus and fibularis tertius (Figure 2.3-1). The tibialis anterior inserts into the base of the first metatarsal and medial to the cuneiform. The extensor hallucis longus and extensor digitorum longus extend across the dorsum of the foot to respectively insert onto the distal hallux and distal phalanges. The fibularis tertius has its insertion on the base of the fifth metatarsal.

Concentric contraction of either the tibialis anterior and/or extensor hallucis longus, which has their insert on the medial part of the foot, will dorsiflex and invert the foot at the talocrural and subtalar joints. The extensor digitorum longus assists dorsiflexion and extends the toes. The fibularis tertius inserts on the lateral foot and will evert the foot while also assisting dorsiflexion. However, the fibularis tertius is mainly responsible for eversion of the foot as its insertion is on the lateral foot. Eccentric contraction of the anterior compartment slows down plantar flexion of the foot. During weight bearing conditions the eccentric contraction of the tibialis anterior is responsible for providing stability to the subtalar joint by stabilising the shank over the foot (Beynnon, Renstrom, Alosa, Baumhauer, & Vacek, 2001; Jastifer & Gustafson, 2014).

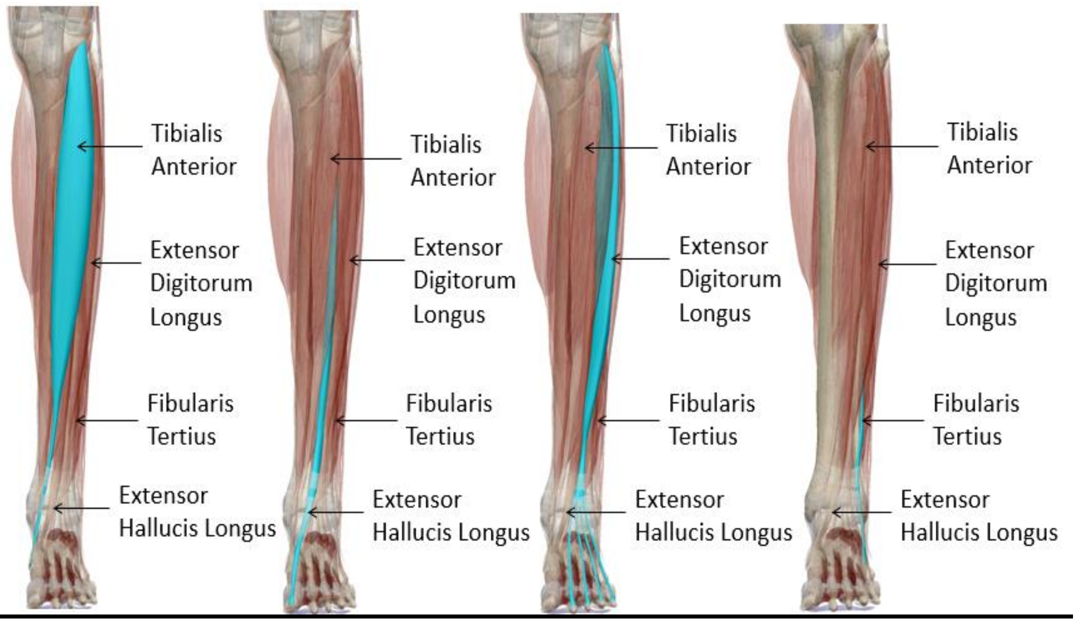

FIGURE 2.3-1: Extrinsic muscles of the anterior compartment. Images courtesy of visible body (Body, 2019).

2.3.1.2 Posterior Compartment

The superficial posterior compartment is formed by the gastrocnemius, soleus and plantaris muscles that combine to form the triceps surae and insert into the posterior calcaneus via the calcaneal tendon (achilles) (Figure 2.3-2). Concentric contraction of the triceps surae group is responsible for plantar flexion of the ankle joint complex and also contributes to knee flexion. Eccentrically the triceps surae assists in dissipating GRF when the forefoot makes initial ground contact (Boden, 2010; Tam, Astephen Wilson, Noakes, & Tucker, 2014).

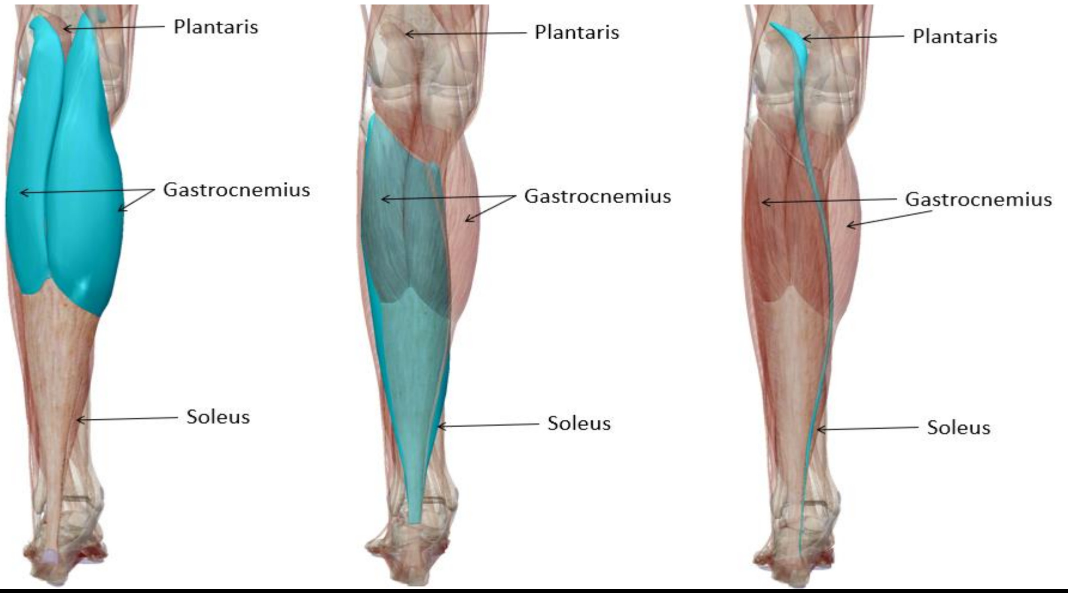

FIGURE 2.3-2: Extrinsic muscles of the superficial posterior compartment. Images courtesy of visible body (Body, 2019).

The deep posterior compartment (Figure 2.3-3) contains the flexor hallucis longus, flexor digitorum longus, and tibialis posterior. The posterior compartment assists in plantar flexion and inversion of the foot. The flexor hallucis longus, flexor digitorum longus inserts on the hallux and phalanges, respectively. These extrinsic muscles run parallel to the plantar aponeurosis and assist the plantar aponeurosis in stabilising the forefoot during the stance phase (Duerinck, Hagman, Jonkers, Van Roy, & Vaes, 2014). The tibialis posterior's unique insertions on the navicular, cuneiform, and cuboid, as well as metatarsals 2-4, enables it to be mainly responsible for maintaining the medial longitudinal arch of the foot (Klein et al., 1996).

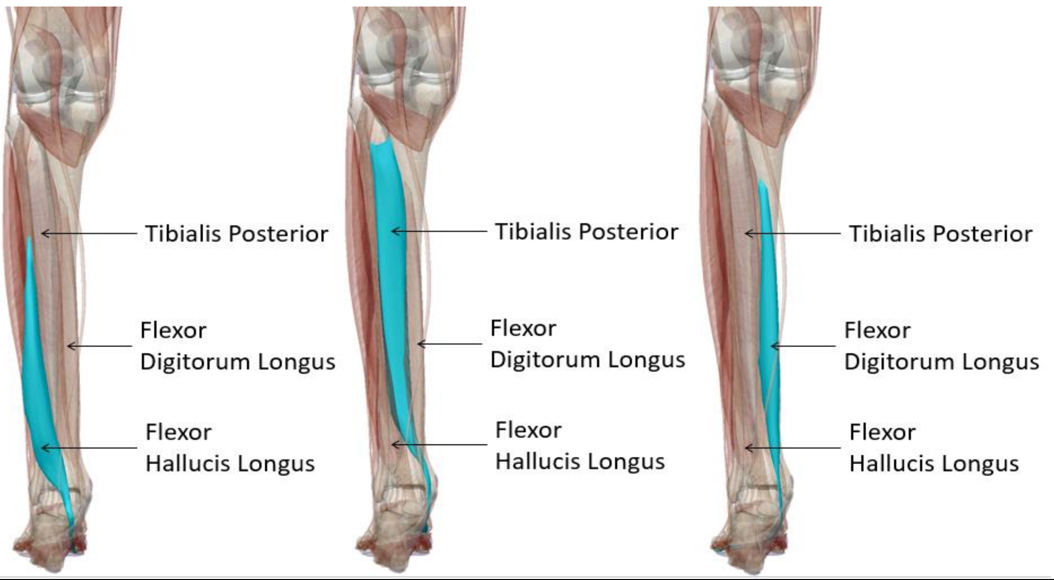

FIGURE 2.3-3: Extrinsic muscles of the deep posterior compartment. Images courtesy of visible body (Body, 2019).

2.3.1.3 Lateral Compartment

The laterally situated fibula provides attachment for the origin of the fibularis longus and brevis, which comprises the lateral compartment of the extrinsic muscles. The muscles of the lateral compartment are able to plantarflex and evert the foot when concentrically contracted. The fibularis longus is able to assist in supporting the medial longitudinal arch. The tendon of the fibularis longus extends posteriorly around the lateral malleolus, runs through the cuboid canal, and crosses diagonally over the plantar foot. The insertion of the fibularis longus is on the medial cuneiform and the base of the first metatarsal, allowing the muscle to pronate the medial forefoot (Figure 2.3-4).

During weight bearing the concentric contraction of the fibularis longus will primarily create a lateral pull on the first ray (first cuneiform and the first metatarsal) (Glasoe, Yack, & Saltzman, 1999). The lateral pull on the first ray causes pronation and an eversion torsion through the forefoot (Johnson & Christensen, 1999). The interdependent nature of movement in the forefoot locks the medial forefoot (Johnson & Christensen, 1999) and the calcaneocuboid joint simultaneously (Bojsen-Møller, 1979), thereby creating a stable lever in the forefoot for propulsion. However, prolonged pronation will lower the first ray and the line of pull of the fibularis longus will lose its ability to limit stresses put on the ligament responsible for first ray integrity, and foot stability (Glasoe et al., 1999). Weight bearing, isometric contraction of the fibularis muscles provides powerful resistance against ankle inversion (Ashton-Miller, Ottaviani, & Hutchinson, 1996). Additionally, eccentric contraction of the fibularis muscles should be able to slow down ankle inversion.

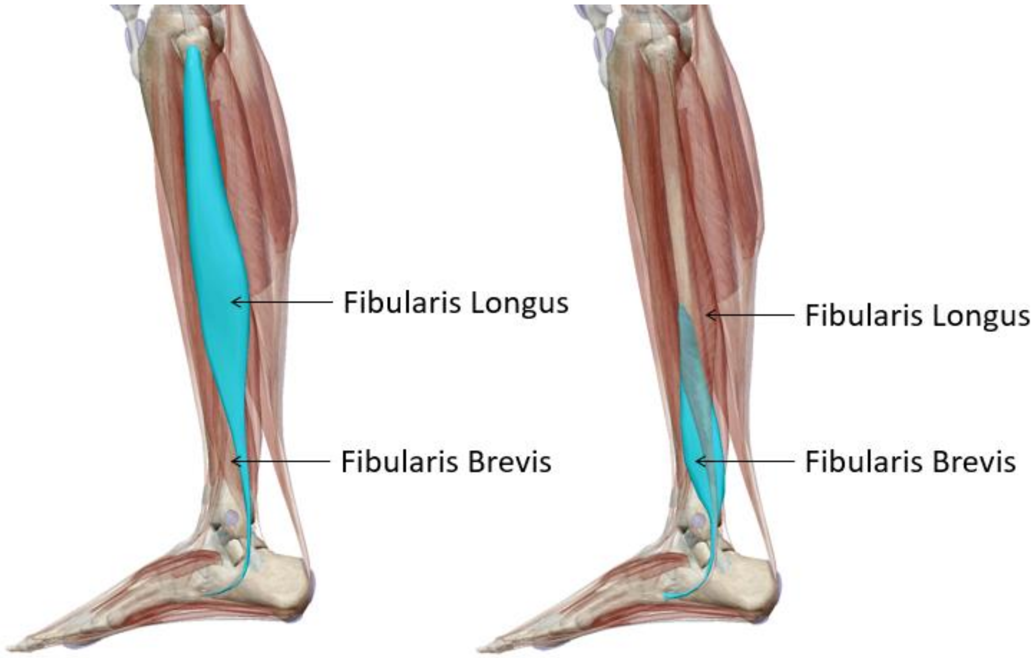

FIGURE 2.3-4: Extrinsic muscles of the lateral compartment. Images courtesy of visible body (Body, 2019).

2.3.2 INTRINSIC MUSCLES OF THE PLANTAR SURFACE

2.3.2.1 FIRST PLANTAR LAYER

Superficial to the sole and directly underneath the plantar fascia is the first plantar layer, consisting of the abductor hallucis, flexor digitorum brevis and abductor digiti minimi (Figure 2.3-5). The muscles of the first plantar layer all originate from the anterior calcaneus and inserts medially on the hallux, middle of the phalanges two to five and lateral on the fifth phalange, respectively. Following the direction of the fibres, it can be expected that the first plantar layer would abduct the first and fifth phalanges and also flex all the toes. During weight bearing (1BW), previous research reported adduction of the metatarsals in relation to the calcaneus when the abductor hallucis and flexor digitorum brevis was electrically stimulated. Furthermore, electrical stimulation of the abductor hallucis and flexor digitorum brevis also inverted and abducted the calcaneus under a weight bearing load of 1BW (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014).

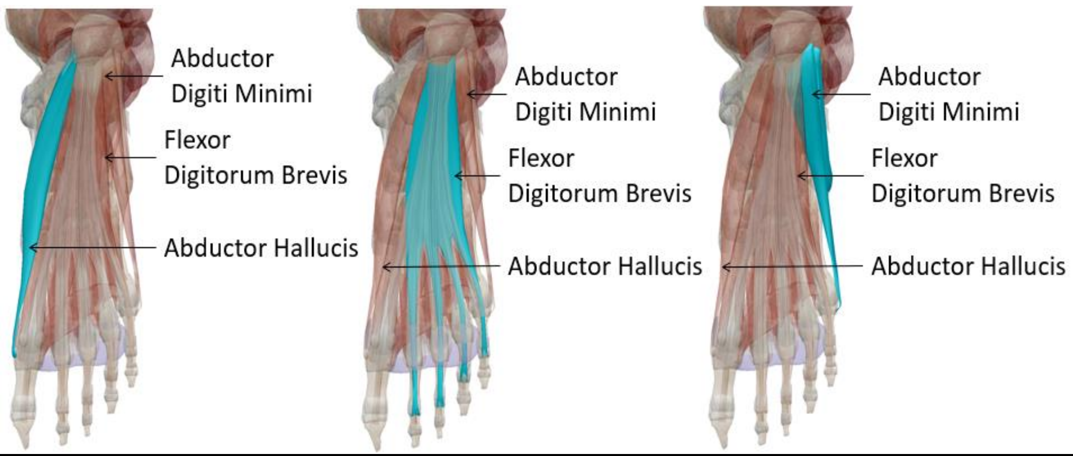

FIGURE 2.3-5: Intrinsic muscles of the first plantar layer. Images courtesy of visible body (Body, 2019).

2.3.2.2 SECOND PLANTAR LAYER

The quadratus plantae and the lumbricals comprise the second plantar layer. The quadratus plantae has a medial and a lateral origin on the calcaneus, the two portions join and insert on the tendon of the flexor digitorum longus. Electrical stimulation of the quadratus plantae abducted the calcaneus in weight bearing (Kelly et al., 2014). The medial surface of the branches of the flexor digitorum longus tendon serves as the origin for the four lumbrical muscles. The lumbricals insert into the dorsal surfaces of phalanges two to four, allowing flexion of the toes at the metatarsophalangeal joints and also extension of the interphalangeal joints (Kelly et al., 2014).

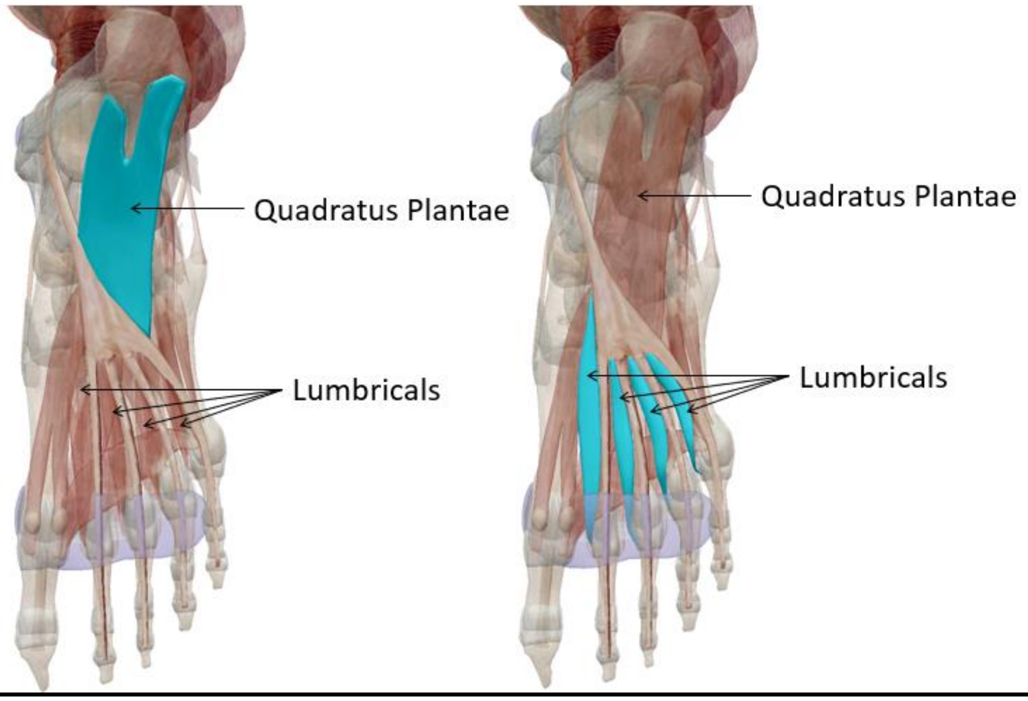

FIGURE 2.3-6: Intrinsic muscles of the second plantar layer. Images courtesy of visible body (Body, 2019).

2.3.2.3 Third plantar layer

The third plantar layer contains the flexor hallucis brevis, adductor hallucis and the flexor digiti minimi. The flexor hallucis brevis originates from the plantar cuboid, as well as the lateral cuneiforms, and splits into two parts to insert on each side of the hallux. The adductor hallucis is composed of a transverse (originating from metatarsophalangeal ligaments three to five) and an oblique head (originating from metatarsals two to four) and inserts to the lateral base of the hallux. The flexor digiti minimi has its origin on the base of the fifth metatarsal and inserts to the lateral side of the base of the fifth phalange, enabling flexion of the fifth metatarsophalangeal joint.

Concentric contraction of the flexor hallucis brevis and adductor hallucis brevis flexes and adducts the hallux, respectively. During the stance phase, it is expected that eccentric contraction of the flexors of the third layer controls extension of the hallux and fifth digit. The hallux plays an important role during push-off and adductor hallucis and flexor hallucis brevis, together with the first plantar layer abductor hallucis, provides stability to the joint (Huson, 1991). It can also be expected that the same muscles will provide stability during the forefoot landing.

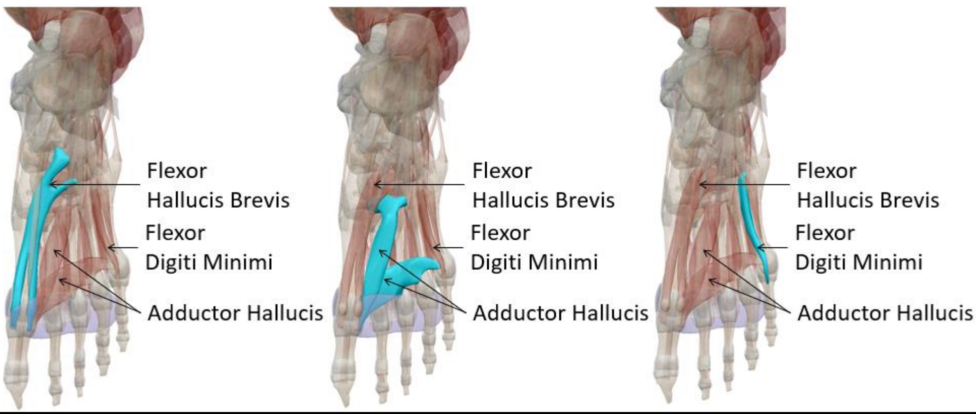

FIGURE 2.3-7: Intrinsic muscles of the third plantar layer. Images courtesy of visible body (Body, 2019).

2.3.2.4 FOURTH PLANTAR LAYER

The fourth and final plantar layer comprises the three plantar interossei, originating from the medial side of each metatarsal three to five and inserts onto the base of these proximal phalanges. The plantar interossei adducts and flexes these digits during concentric contraction. The interossei provides isometric and eccentric control of toe splay, and offers stability to the forefoot (Huson, 1991).

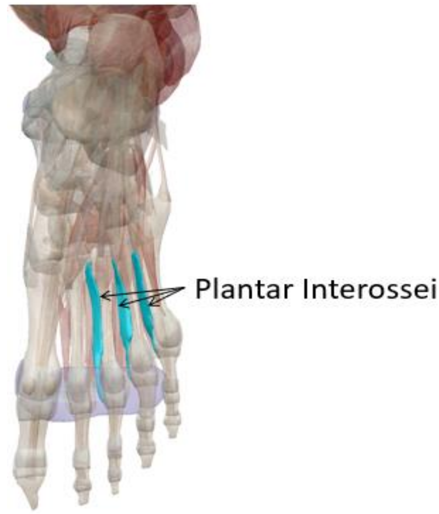

FIGURE 2.3-8: Intrinsic muscles of the fourth plantar layer. Images courtesy of visible body (Body, 2019).

2.4. ARCHES OF THE FOOT

Due to their anatomical configuration and collaborative function foot segments are grouped together to form supportive arches. The main function of these arches is to create stiffness in the foot and to transmit forces from the rearfoot to the forefoot during normal walking gait (Venkadesan et al., 2020). In the normal functioning foot, there are two distinct arches in the anterior-posterior plane and two arches in the transverse plane. In the anterior-posterior plane the medial longitudinal arch (MLA) consists of the calcaneus, talus, navicular, first cuneiform and the first metatarsal (first ray) as well as the middle column of the tarsometatarsal joint (Hicks, 1954). The lateral longitudinal arch (LLA) is formed by the calcaneus, cuboid, and fourth and fifth metatarsals (Nester, 2009). The bases of the metatarsals, cuboids and the three cuneiforms create the transverse arch (McKeon et al., 2014). Anteriorly to the transverse arch, the metatarsal anterior transverse arch (MetATA) is formed by the heads of the metatarsals running from medial to lateral in the forefoot (Bojsen-Møller & Flagstad, 1976; Duerinck et al., 2014; Huson, 1991).

The pronation and supination of the weight bearing subtalar joint, as previously described, is often used to define MLA and foot biomechanics (Hicks, 1954). The height of the navicular relative to the floor is often employed as a measure of MLA function as this bony landmark acts as the keystone in the MLA (Eichelberger, Blasimann, Lutz, Krause, & Baur, 2018). However, as the calcaneus, medial metatarsals and phalanges form part of the MLA, segmental movement of the calcaneus (hindfoot), and metatarsal and phalanges (forefoot) can also be observed and used to describe weightbearing MLA function (Hicks, 1953; Kelly et al., 2014). The weightbearing function of the MLA was also moderate to well correlated with lateral longitudinal arch movement (Gwani, Asari, & Mohd Ismail, 2017). The correlation in movement between the two longitudinal arches, is due to the calcaneus tuberosity forming the posterior mainstay for both the MLA and the lateral longitudinal arches while the bones in the transverse arch (midtarsal joint) form the keystones in both arches. The cuneiforms and cuboid of the midtarsal joint form part of the MLA (cuneiforms) and the lateral longitudinal arch (cuboid). The MLA, lateral longitudinal arch and transverse arch movement are thus correlated (Gwani et al., 2017). Furthermore, a recent study demonstrated that the curvature (height) of the transverse arch increases the stiffness along the longitudinal axis of the foot (Venkadesan et al., 2020).

The correlations between the MetATA and the other arches are not frequently investigated (Duerinck et al., 2014; Gwani et al., 2017). In change of direction tasks, where the heel is typically off the floor (Losito, 2017) creating forefoot stiffness may be of particular importance. When the calcaneus is not in contact with the floor one of the “trusses” of the MLA (Glasoe et al., 1999) is removed, potentially decreasing longitudinal stiffness. Indeed, previous research has shown that when the forefoot is loaded in flat footed primates, the lack of longitudinal stiffness resulted in midfoot extension (midtarsal break) (D'Août, Aerts, De Clercq, De Meester, & Van Elsacker, 2002). In humans, the lack of longitudinal stiffness of the foot during forefoot loading also resulted in midfoot extension (DeSilva et al., 2015), potentially increasing injury risk (Fraser, Feger, & Hertel, 2016; Kagaya, Fujii, & Nishizono, 2015).

DeSilva et al. (2015) hypothesised that increased midfoot mobility was the result of the lack of stiffness in the plantar ligament and the plantar aponeurosis as well as the lack support from the muscles supporting the arches of the foot. It is widely accepted that in the human foot the longitudinal arch is supported by the plantar aponeurosis which stiffens the foot via the windlass mechanism (Hicks, 1954). However, several studies have shown that the windlass mechanism and the aponeurosis are not primarily responsible for MLA stiffness (Farris, Birch, & Kelly, 2020; Fessel et al., 2014; Welte, Kelly, Lichtwark, & Rainbow, 2018). In fact, plantar flexors and intrinsic foot muscles where found to increase and/or resist toe flexion and increase longitudinal foot stiffness (Duerinck et al., 2014; Farris et al., 2020; Sooriakumaran & Sivananthan, 2005). Furthermore, Venkadesan et al. (2020) has shown that the curvature (height) of the arches in the frontal plane is important to creating longitudinal stiffness of the foot.

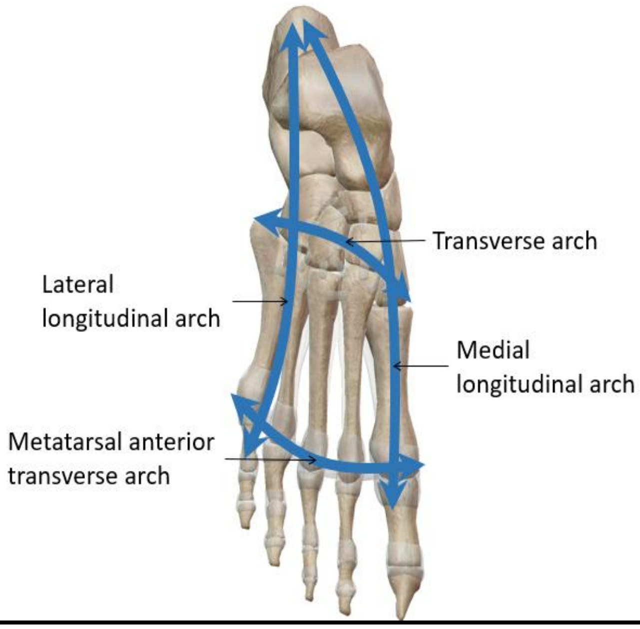

FIGURE 2.4-1: Arches of the foot. Images courtesy of Visible Body (Body, 2019).

2.5. BIOMECHANICAL MODELLING OF THE FOOT AND ANKLE

Biomechanical models are used to study and interpret the motion of the human body. Creating these models requires the accurate tracking of 3-dimensional movement of the segments. The most direct motion tracking method entails capturing the movement of pins inserted directly into the bones of participants (Leardini, Caravaggi, Theologis, & Stebbins, 2019). Bone pin studies are considered the gold standard, but this method is invasive, can cause infection and the pins are prone to bending as a result of soft tissue movement, in this way distorting kinematic results (Leardini et al., 2019; Nester et al., 2007). A recent review of foot modelling by Leardini et al. (2019) noted that less invasive biomechanical methods like inertial sensors and marker-less dynamic 3D scanning exists but tends to be less accurate.

Gait analysis in both clinical and academic settings mostly use less invasive yet adequately reliable optoelectronic stereophotogrammetry. This method involves placing passive or active markers on specific anatomical (bony) landmarks (Leardini & Caravaggi, 2017). Marker placement and skin artifacts, skin distortion and soft tissue displacement are common limitations of this method to which there is currently no workable solution (Leardini, Chiari, Della Croce, & Cappozzo, 2005; Nicholson et al., 2018). However, optoelectronic stereophotogrammetry is considered sufficiently reliable for defining and tracking the movement of the lower limb and foot (Leardini & Caravaggi, 2017). Anatomical landmarks are used to establish the anatomical reference frame of the segment and a minimum of three markers is needed to track the 3-dimensional movement of a single segment (Leardini & Caravaggi, 2017). The motion captured represents the position and orientation of the underlying bone(s) and is used to calculate triplanar rotations of the segment and joints (Leardini et al., 2005). To describe and report on joint motion and kinematics the standardised Joint-Coordinate-System convention is used (Wu et al., 2002).

The kinematic analysis of foot biomechanics has evolved from modelling the foot as a single unit to grouping the foot bones together to create several segments (Leardini et al., 2019). Modelling the foot as a single unit does simplify the interpretation of lower limb biomechanics. However, as can be derived from the previous sections in this chapter, the foot does not function as an inflexible and rigid unit. Indeed, previous research has shown that the foot can and should be studied as several independent segments that acts interdependently (Nester et al., 2007). Several multi-segmental foot models are available and differ in the number of segments, landmarks used, marker sets, definition of the anatomical frames and the joint coordinate system convention used to calculate the joint kinematics (Leardini & Caravaggi, 2017; Leardini et al., 2019). These models group bones together in segments and treat these segments (from 3 to 26 segments) as individual rigid bodies (Leardini & Caravaggi, 2017). The reason for grouping bones together is the assumed function of these bones as a unit. Secondly, the surface of the foot is too small to have several markers in a confined space as the cameras have trouble distinguishing between markers (Leardini & Caravaggi, 2017). Grouping bones together simplifies the kinematic analysis of the foot but may lead to error. The more bones and articulations in a rigid segment, the greater the chance of violating the 'rigid body assumption' (Nester et al., 2010). For example, in a study by Nester et al. (2010) where rigid segment kinematics was compared to kinematic data of the specific bones that make up the rigid segment, the largest error, apart from considering the entire foot as a rigid body, was found in models that combine the navicular, cuneiforms and all five metatarsals. The researchers suggest that kinematic data derived from multi-segmental foot models should be considered as movement between functional segments rather than anatomical joints. This recommendation can also be applied to the ankle joint complex, as previously reviewed in this chapter, the movement in ankle joint is arranged as the combined function of talocrural and the subtalar joints.

This study will use the model as described by Leardini et al. (2007b) and also adopt the modifications made to it (Portinaro, Leardini, Panou, Monzani, & Caravaggi, 2014). The model uses a relatively small number of markers to track several foot segments and was found to be repeatable in adult subjects when markers where attached by an experienced technician (Caravaggi, Benedetti, Berti, & Leardini, 2011; Deschamps et al., 2012). The model consists of a shank, calcaneus, midfoot, metatarsal and hallux segments. A detailed definition of the segments' composition, anatomical landmarks used to track and define the segment reference frames as well as the definition of segment rotations can be found online on the motion analysis software Visual 3D™ (v5.0, C-Motion, USA) web page.

2.5. CONCLUSION

To study the functional anatomy of the foot and lower limb, individual structures (bones, ligaments, and muscles) form segments which act as a collective to perform a specific function. Individual segments create arches (supported by passive and active structures) which can act independently or in unison allowing complex movement and ensuring optimal foot function. Passive structures like the plantar aponeurosis provide global support to the whole foot while the intrinsic and extrinsic muscles control individual bones and/or segments. The subtalar and talocrural joints act as conduits transferring rotational movement from the distal to proximal (foot to the shank) and from proximal to distal (shank to the foot). The complex and intricate movement of the foot thus plays a role in the movement observed in joints proximal to the foot, like the ankle and knee. It is therefore important to understand how manipulating foot function may influence the transfer of forces to the ankle and knee.

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CHAPTER 3
AIMS AND STRUCTURE

3.1. BACKGROUND

Suboptimal foot mechanics are linked to both non-contact anterior cruciate ligament (ACL) rupture and lateral ankle sprain (LAS) injury risk factors. For example, calcaneal eversion and subtalar joint pronation are coupled with internal rotation of the shank which increases ACL strain. In LAS injuries, forefoot instability is linked with an excessive ankle inversion/supination angle and/or velocity. LAS injury risk increases when the resultant ground reaction force (GRF) acts too far medially or laterally to the ankle joint centre and the forefoot is unable to maintain ground contact.

Previous research in the field of foot biomechanics has shown that movement of the subtalar joint and the calcaneus are coupled with rotations of the shank. Specifically, subtalar joint pronation is coupled with internal shank rotation and supination of the subtalar joint is coupled with external rotation of the shank. Manipulation of the calcaneus and subtalar joint have been shown to influence rotational movement of the shank in both static and dynamic conditions. It is thought that the distal-proximal coupling between the foot, ankle, and shank can thus modulate the transfer of rotation to the knee, possibly reducing ACL strain.

LAS risk is mitigated by resisting ankle inversion and supination. Extrinsic muscles of the foot are responsible for controlling ankle inversion and supination. However, the extrinsic muscles are only effective in resisting possible injurious inversion and supination motion if the muscle has a sufficient moment arm length.

Sufficient moment arm length is only possible when the subtalar joint is able to resist deformation under load (pronation). Subtalar joint movement (pronation-supination) is also manifested in the movement of the medial longitudinal arch (MLA). Recent research has revealed that the intrinsic and extrinsic muscles acting on the foot are able to create the MLA and resist its deformation. Creating the MLA will also increase the moment arm length of extrinsic muscles responsible for controlling ankle inversion and supination, in this way decreasing LAS injury risk. Current mainstream prophylactic programmes do not include specific strategies focused on foot musculature. A crucial part of lower limb ligament injury prevention may thus be overlooked.

3.2. Aims

Previously, the effect of foot muscle specific intervention programmes was mostly assessed during low intensity, straight line walking and/or running tasks. The coupling between the foot segments and the foot and shank changes as task and intensity of the task changes. It is thus unclear whether strengthening the muscles acting on the foot would translate to dynamic, unanticipated change of direction tasks. It is also unclear whether training the muscles acting on the foot would influence the risk factors associated with ACLR and LAS, during high intensity, unanticipated changes of direction. The aim of this thesis was to:

a) Describe and quantify the coupling relationship between the tri-planar movement of the calcaneus and the transverse plane rotations of the shank during high intensity unanticipated changes of direction.
b) Describe and quantify the coupling relationship between the movements of the segments of the foot during high intensity unanticipated changes of direction.
c) Undertake an intervention pilot study to establish proof of concept of strengthening the muscles of the foot on ACL and LAS injury mechanisms.

3.3. THESIS STRUCTURE

The thesis is structured to investigate the biomechanical theory that training the muscles acting on the foot would influence foot function and the risk factors associated with ACLR and LAS risk as it occurs in court sports. **Chapter four** (literature review) and **chapters six and seven** describing inter-segmental coordination patterns are written and presented as manuscripts suitable for stand-alone publications in refereed academic journals. (These manuscripts, attached in the Appendices, are under in press and review, respectively, and thus edited to suit the specific journal style).

Chapter eight describes the 6-week progressive intervention programme. A randomised controlled pilot study was undertaken to provide proof of concept. The workflow of the pilot study is presented, and the results of the proof-of-concept study are discussed in in **chapter eight.** In **chapter nine** the key findings of the study and future directions for a randomised controlled injury prevention study is discussed.

The **Appendix A** contain material from chapter five and seven which was presented as a conference poster and presentation, respectively. **Appendix B** contain the manuscripts put forward for publication and pertaining the content in chapters four, six and seven, respectively. The literature review (chapter 4) and chapter seven is in press. The manuscripts representing chapters six and are currently under review. **Appendix C** contain the individual coordination patterns of each athlete. These figures are applicable to chapters six and seven. The report submitted to the Badminton World Federation which provided financial support for this study is attached as **Appendix D**. A sample of coach and athlete information packs, consent forms and injury history questionnaires are provided in **Appendix E**. The ethical approval granted from Massey University is attached in **Appendix F**. Each exercise of the intervention programme is described in **Appendix G.**

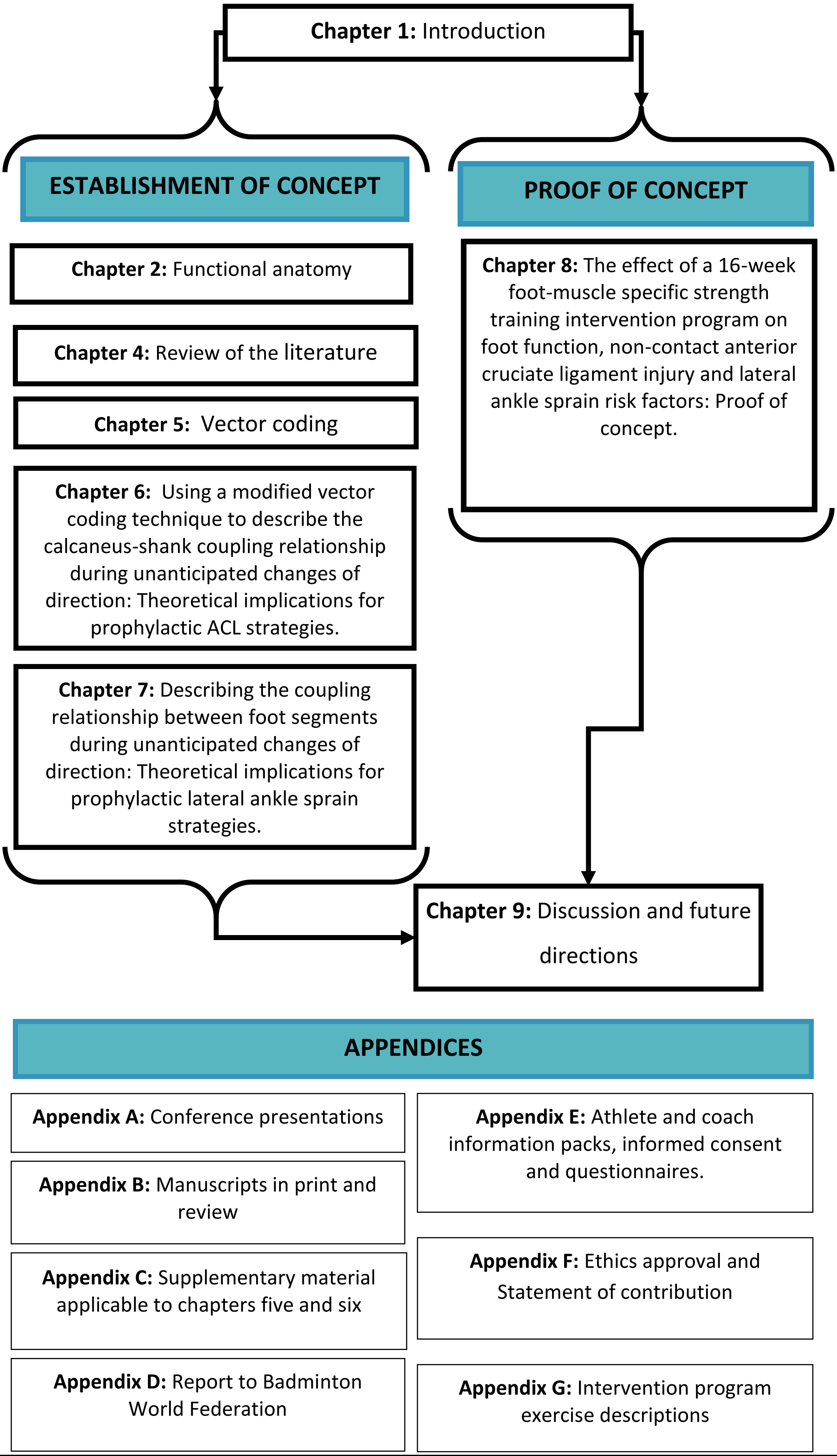

FIGURE 3.3-1: Schematic of thesis outline.

CHAPTER 4
REVIEW OF THE LITERATURE

PUBLICATION:

Carla van der Merwe, Sarah P. Shultz, G. Robert Colborne, Philip W. Fink. *Research Quarterly for Exercise and Sport*, (In Press).

4.1. ABSTRACT

BACKGROUND AND OBJECTIVES

The active and passive structures of the foot act in unison to not only be compliant enough to assist in ground reaction force attenuation but also resist deformation to provide a stable base of support. A foot that is unable to adjust to the imposed demands during high-intensity sporting activities may alter the moments and forces acting on the joints, increasing the risk of non-contact anterior cruciate ligament ruptures (ACLR) and lateral ankle sprains (LAS). Prophylactic strengthening programmes are often used to reduce the risk of these injuries, but at present, very few prophylactic programmes include foot specific strengthening strategies. The aim of this theoretical review is to ascertain the prophylactic role strengthening muscles acting on the foot may have on ACLR and LAS injury risk.

METHODS

Literature relating to risk factors associated with ACLR and LAS injury and the anatomy and biomechanics of the lower limb was searched. A theoretical, narrative approach was followed to synthesize the information gathered from the articles.

RESULTS

The foot segments are governed by the congruity of the articulations and the activity of the foot muscles. The coupling effect between the foot segments, and the shank and the foot play a role in both ACLR and LAS injury mechanisms.

CONCLUSIONS

Strengthening the muscles acting on the foot may have a significant impact on ACLR and LAS injury mechanisms.

KEY WORDS

Athletic injury, anterior cruciate ligament, ankle sprain, biomechanics

4.2. INTRODUCTION

The anatomy and function of the foot are remarkably complex with both passive and active structures functioning as a single component. The foot segments act as an interactive unit to not only become a stiff configuration to support and propel the body but also become compliant enough to adapt to the contact surface and dissipate ground reaction forces (GRF) (Gwani, Asari, & Mohd Ismail, 2017; Holowka, O'Neill, Thompson, & Demes, 2017; McKeon, Hertel, Bramble, & Davis, 2014; Williams III, Tierney, & Butler, 2014). Yet very little is known about the foot's role in injury mechanics and subsequently, the possible prevention of the injuries that athletes are exposed to during high impact sports. In sports where abrupt, unanticipated changes of direction are common, the athlete is at an increased risk for sustaining non-contact injuries, such as anterior cruciate ligament ruptures (ACLR) and lateral ankle sprains (LAS) (Cassell, 2012; Doherty et al., 2014; Fong, Chan, Mok, Yung, & Chan, 2009; McKay, Goldie, Payne, & Oakes, 2001). ACLR and LAS injuries primarily occur during unbalanced single leg decelerations (Boden, Torg, Knowles, & Hewett, 2009; Fong, Ha, Mok, Chan, & Chan, 2012; Kristianslund, Bahr, & Krosshaug, 2011; McCriskin, Cameron, Orr, & Waterman, 2015; Olsen, Myklebust, Engebretsen, & Bahr, 2004; Stuelcken, Mellifont, Gorman, & Sayers, 2016). During movements where lower body stability is compromised (Powers, 2010; Stacoff, Steger, StüSsi, & Reinschmidt, 1996) and GRFs are large (Boden, Sheenan, Torg, & Hewett, 2010; Boden et al., 2009; Hewett et al., 2005; Serpell, Scarvell, Ball, & Smith, 2012), external moments around the knee or ankle can exceed the injury threshold of the structures stabilising the joint, causing injury to the soft tissue of joints (Fong, Chan, et al., 2009; Fong et al., 2012; Kristianslund et al., 2011; Powers, 2010; Raina & Nuhmani, 2014).

Prophylactic programmes have been successful in minimising the incidence of ACLR and LAS injury; they commonly include physical (i.e. balance, plyometric, strength, coordination) training, education about injury mechanisms, and/or technique training and feedback (Kaminski et al., 2013; Kerkhoffs et al., 2012; McCriskin et al., 2015; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Paszkewicz, Webb, Waters, McCarty, & Van Lunen, 2012; Schiftan, Ross, & Hahne, 2015; Shultz et al., 2015; Vriend, Gouttebarge, van Mechelen, & Verhagen, 2016; Yoo et al., 2010). However, these programmes often fail to report on the mechanisms involved in the reduction of injury incidences (Eils, Schroter, Schroder, Gerss, & Rosenbaum, 2010; Hewett & Johnson, 2010; Kaminski et al., 2013; Kynsburg, Panics, & Halasi, 2010; McKay et al., 2001; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Paszkewicz et al., 2012; Vriend et al., 2016; Yoo et al., 2010). Jump landings, stopping, and changing direction are all closed kinetic chain manoeuvres, with at least one foot in contact with the ground. The stability and orientation of the foot is likely to play a significant role in the position of the centre of pressure under the foot, influencing the moments and forces acting on the lower limb (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014; McKeon et al., 2014; Nigg, Baltich, Federolf, Manz, & Nigg, 2017; Vriend et al., 2016; Williams III et al., 2014). It is thus reasonable to expect the intrinsic and extrinsic muscles of the foot to play a role in influencing risk factors associated with lower limb injuries. Yet remarkably very little is known about the prophylactic effect training the muscles acting on the foot may have on the risk factors associated with ACLR and LAS injury. The aim of this review is to investigate the theoretical probability that training the intrinsic and extrinsic foot musculature may have on the foot's function and the risk factors associated with ACLR and LAS injuries.

4.3. Risk factors

4.3.1 Non-contact anterior cruciate ligament injury mechanisms

ACL injuries are caused by a multitude of factors. Intrinsic and non-modifiable risk such as anatomy, genetics, gender, maturation, hormonal status, joint geometry, and previous injuries are factors unique to the athlete. In contrast, extrinsic risks such as nutrition, neuromechanics, and level of play are generalisable to all athletes and therefore modifiable (Barber-Westin & Noyes, 2018; Shultz et al., 2015; Smith et al., 2012). The focus of most ACLR injury prevention programmes is therefore to change extrinsic risk factors related to high-risk body positions associated with increased ligament strain (Barber-Westin & Noyes, 2018). At risk athletes are thought to display rotation and lateral trunk flexion, adduction and internal rotation of the hip, large knee valgus/abduction angle and moments, and tibial rotation (Gamada, 2014; Shultz et al., 2015; Stuelcken et al., 2016).

Proximal to distal transfer of movement along the thigh towards the knee is modulated at the hip (Griffin et al., 2000; Powers, 2010). The muscles of the core, as well as the hip extensors and abductors, stabilise the trunk over the hips, preventing the lateral shift of the GRF vector (Powers, 2010). The muscles acting at the hip also prevent hip adduction and internal rotation of the thigh. When these muscles are not providing sufficient control of the thigh it can result in valgus collapse of the knee, straining the ACL (Hollman et al., 2009; Powers, 2010). It is also thought that limited eccentric strength of the hip extensors alters hip flexion angles. A decrease in hip flexion, as often seen in female athletes, decreases hamstring activity and increases quadriceps activation resulting in a large intrinsic knee extensor moment, possibly increasing ACL strain (Pollard, Sigward, & Powers, 2010; Withrow, Huston, Wojtys, & Ashton-Miller, 2008).

Rotational movement is also transferred to the knee from the distal shank which is modulated by rotations at the foot-ankle-shank complex (Nigg et al., 2017). Previous research has revealed a coupling relationship exists between subtalar joint pronation and shank internal rotation, and subtalar joint supination and shank external rotation (Hicks, 1953; Huson, 1991). During rearfoot eversion and subtalar joint pronation the talus moves into adduction. As a result of the mortise-tenon shape of the talocrural joint and the tensile properties of the surrounding ligaments, the shank follows the adducting movement of the talus into internal rotation (Huson, 1991). In this way, rearfoot eversion and subtalar joint pronation may have a link to high-risk movement associated with increased ACL injury risk. Indeed, previous research has linked static measures of navicular drop, indicative of subtalar joint pronation, to larger knee valgus angles and increased risk of ACL injury (Beckett, Massie, Bowers, & Stoll, 1992; Hertel, Dorfman, & Braham, 2004; Loudon, Jenkins, & Loudon, 1996; Woodford-Rogers, Cyphert, & Denegar, 1994). Furthermore, excessive dynamic rearfoot pronation and rearfoot eversion angles during change of direction and jumping tasks have been linked to increased knee valgus angles which in turn may increase the risk for ACL injuries (Holland, 2016; Kagaya, Fujii, & Nishizono, 2015; McLean, Lipfert, & Van Den Bogert, 2004).

4.3.2 LATERAL ANKLE SPRAIN RISK FACTORS

Lateral ankle sprain refers to injury to any of the three main ligaments located on the lateral side of the foot-shank complex (Ferran & Maffulli, 2006; Gribble et al., 2016). The lateral ligaments are tasked to assist in the stabilisation of both the talocrural (tibia-talus) and subtalar (talus-calcaneus) joints (Hertel, 2002; Stephens & Sammarco, 1992). The anterior talofibular ligament (ATFL) is commonly injured during LAS and 50 – 75 % of LAS injuries include the calcaneofibular ligament (CFL). Posterior talofibular ligament (PTFL) injuries are infrequent and are mostly injured when the ATFL and CFL have failed (Ferran & Maffulli, 2006; Rasmussen, Kromann-Andersen, & Boe, 1983; Stephens & Sammarco, 1992). In court sports, an injurious external moment is typically the result of an athlete landing on another athlete's foot (Gribble et al., 2016; Panagiotakis, Mok, Fong, & Bull, 2017) or during unanticipated changes of direction or side shuffling movements (Fong et al., 2012; Fong, Hong, et al., 2009; Kristianslund et al., 2011). After initial foot contact, the athlete's centre of mass is likely to continue in the same direction as at initial foot strike. Thus, in jump landings, the centre of mass will descend while during a change of direction or shuffling manoeuvre the centre of mass will continue to shift horizontally after the athlete has planted the supporting foot. Both the increasing resultant GRF and laterally moving centre of mass will increase the external inversion-adduction moment around the subtalar joint axis (Kirby, 2001).

Analysis of LAS incidences reported substantial deviations in sagittal plane ankle angles, ranging from a large plantar flexion angle (up to 50°) to a large dorsiflexed (up to 22°) position (Fong, Hong, et al., 2009; Gehring, Wissler, Mornieux, & Gollhofer, 2013; Kristianslund et al., 2011). Although sagittal plane ankle angles varied, a common trend amongst LAS incidences was an inverted and adducted foot position at initial foot strike. Both the ATFL and CFL are strained when the subtalar joint is inverted and internally rotated in conjunction with the externally rotating shank (Hollis, Blasier, & Flahiff, 1995; Panagiotakis et al., 2017; Renstrom et al., 1988; Stormont, Morrey, An, & Cass, 1985). The more frequently injured ATFL is more likely to be injured in isolation on a more plantarflexed foot, particularly at initial foot strike when landing from a jump (Rasmussen et al., 1983; Renstrom et al., 1988; Stormont et al., 1985). The dome of the talus is wider anteriorly than posteriorly and has a small contact area between the tibia and talus in plantarflexion, decreasing ankle stability. As there are no muscular attachments to the talus, the ATFL is the main structure to prevent anterior-lateral displacement and excessive inversion and adduction of the talus when the calcaneus inverts (Hertel, 2002; Huson, 1991; Raina & Nuhmani, 2014). During LAS incidences, where a large GRF acts medial to the subtalar joint and the centre of pressure is positioned in the forefoot, the talocrural joint is forced into further inversion and adduction (Fong et al., 2012; Fong, Hong, et al., 2009; Gehring et al., 2013; Kristianslund et al., 2011; Panagiotakis et al., 2017; Raina & Nuhmani, 2014). As the forefoot is fixed, the calcaneus inverts while the rearfoot swings laterally to the anterior-medial located centre of pressure creating a pivoting, internal rotation movement of the foot (Fong, Hong, et al., 2009; Gehring et al., 2013). The inversion movement of the calcaneus is coupled with the extension and abduction of the talus (Huson, 1991; Raina & Nuhmani, 2014). The extension-abduction movement of the talus is in turn coupled with external rotation of the shank as a result of the mortise shaped talocrural joint and the tensile properties of the ATFL (Huson, 1991; Ito et al., 2017; Siegler, Chen, & Schneck, 1988).

The CFL is more likely to be injured when initial foot contact occurs more posteriorly towards the midfoot or heel, as may occur during side shuffling or on awkward landing from a jump. In midfoot to heel landings, the calcaneus can be expected to be more extended and internally rotated creating a supinated foot (plantarflexed-inverted-adducted) (Brockett & Chapman, 2016; Hertel, 2002). The talocrural joint is very stable in the supinated foot position as a result of its mortise-tenon shape but also due to the shape of articulating surfaces of the tibial cochlea and talar trochlea. The congruity of the tibial-talar surfaces allows minimal frontal and transverse plane movement at the talocrural joint in this position (Hollis et al., 1995). Frontal- and transverse plane movement of the rear-foot is thus mostly rendered from the subtalar joint (Hollis et al., 1995). Furthermore, in weight-bearing activities plantar ligaments and intrinsic foot muscles create tension, compacting the joints of the foot and forming a stable unit (Fessel et al., 2014; Ito et al., 2017; Kirby, 2001). However, under a large external supination moment, an extreme supination range of motion in the 'locked' supinated foot will be reached, demanding frontal plane movement at the subtalar joint, and potentially straining the CFL (Hertel, 2002; Panagiotakis et al., 2017; Siegler et al., 1988).

4.4. Preventing injuries – mechanics and muscle strength

The biggest risk to the ACL seems to arise from the shear forces created by tibial rotation, resulting in excessive strain to the ligament (Mokhtarzadeh et al., 2015; Oh, Lipps, Ashton-Miller, & Wojtys, 2012). As previously mentioned, there appears to be a link between rearfoot eversion, subtalar joint pronation and rotation of the shank that could increase ACLR injury risk. In this way controlling rearfoot eversion and resisting subtalar joint pronation may result in decreasing the magnitude of rotational forces transferred to the knee, via the shank. Previous research has found that stimulation of the intrinsic foot muscles, specifically the abductor hallucis, flexor digitorum brevis and quadratus plantae, inverted and abducted the calcaneus and also countered subtalar joint pronation by resisting medial longitudinal arch deformation under load (Kelly et al., 2014; Kelly, Farris, Lichtwark, & Cresswell, 2018). In addition to the possible support rendered to the medial longitudinal arch by the intrinsic muscles, the extrinsic muscles, tibialis posterior, peroneus longus, flexor digitorum longus and to a lesser extent, the tibialis anterior and the flexor hallucis longus, are also involved in supporting the medial longitudinal arch (Fraser, Feger, & Hertel, 2016b; Kaye & Jahss, 1991; Klein, Mattys, & Rooze, 1996).

Intervention studies focusing on improving intrinsic foot muscle function are starting to emerge. At least two studies found that following an intervention programme that focused on training the intrinsic muscles of the foot improved dynamic stability during single limb reaching tasks (Mickle, Caputi, Potter, & Steele, 2016; Mulligan & Cook, 2013). They concluded that the increasing toe flexor strength and executing the ‘short foot exercises’ prescribed respectively improved the ability of the medial longitudinal arch to provide dynamical stability. It is unclear whether improving arch function during low intensity, relatively static tasks will transfer to more dynamic, high intensity activities such as change of direction where forces are high and segment angles changes rapidly. However, smaller, intrinsic muscles are adept in detecting inversion/eversion and/or abduction/adduction movement of the foot and it is thought that strong, smaller muscles may act to minimise lower limb injuries (Nigg, 2009; Nigg et al., 2017). It thus may be possible that by increasing the activity and strength of the intrinsic and extrinsic foot muscles, arch function can be optimised, conceivably decreasing the transfer of rotation to the knee, possibly decreasing ACL injury risk during dynamic activities.

LAS injury risk increases when an increasing external inversion-adduction moment is not counteracted by an internal everting moment and/or the inversion velocity is not dampened (Ashton-Miller, Ottaviani, Hutchinson, & Wojtys, 1996; Chu et al., 2010; Fong, Chan, et al., 2009; Kristianslund et al., 2011). Creating an eversion moment and controlling the inversion velocity requires the medial metatarsal segment to plantarflex and evert, enabling the forefoot to maintain contact with the floor. As initial foot contact is made, the eccentric contraction of both the extrinsic evertors and invertors of the foot, fibularis longus, tibialis posterior, and tibialis anterior act to slow down the inversion velocity of the foot (Brockett & Chapman, 2016; Hertel, 2002; Munn, Beard, Refshauge, & Lee, 2003). The internal eversion moment is initiated in the forefoot and can be created by means of concentric contraction of the extrinsic fibularis longus muscles. The fibularis longus creates a mainly lateral pull on the first metatarsal and medial cuneiform (i.e., first ray) (Glasoe, Yack, & Saltzman, 1999; Hertel, 2000). The diagonal pull on the first ray causes pronation and an eversion torsion through the forefoot, increasing the internal eversion moment (Ashton-Miller et al., 1996; Johnson & Christensen, 1999). The interdependent movement of the metatarsals in relation to the calcaneus locks the medial forefoot (Johnson & Christensen, 1999) and the calcaneocuboid joint simultaneously (Bojsen-Møller, 1979), thereby creating a stable metatarsal segment. When the metatarsal segment is stable, the calcaneus is free to align with the shank (Graf & Stefanyshyn, 2012). When the shank and calcaneus segments are aligned, shear forces between the segments are limited and relative segmental velocities are minimized, possibly minimising the LAS risk.

The stable metatarsal segment also implies a medially shifted centre of pressure (Fong et al., 2012; Fong, Hong, et al., 2009; Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005). Preventing a lateral shift in the centre of pressure under the foot is important in minimising the length of the extrinsic inversion-adduction moment. In change of direction tasks, the forefoot typically makes initial contact (Losito, 2017). Maintaining forefoot contact and stability is thus important while the calcaneus is off the floor. The fibularis longus and flexor hallucis longus flexes and everts the hallux, respectively and assist the plantar aponeurosis in stabilising the forefoot (Duerinck, Hagman, Jonkers, Van Roy, & Vaes, 2014; Fraser, Feger, & Hertel, 2016a). The importance of hallux stability in decreasing LAS risk is emphasised by previous research that reported that athletes with a larger hallux extension range of motion (interpreted here to imply weaker flexor hallucis longus) were more prone to sustain LAS than athletes with a smaller hallux extension range of motion

(Willems, Witvrouw, Delbaere, De Cock, et al., 2005; Willems, Witvrouw, Delbaere, Philippaerts, et al., 2005). Furthermore, the abductor hallucis is particularly equipped to withstand fatigue, allowing the muscle to maintain the prolonged integrity of the foot (Kelly, Racinais, & Cresswell, 2013). Thus, increasing the eccentric and concentric strength of the muscles acting on the hallux and metatarsals may assist in mitigating both LAS and ACLR injury risk factors.

4.5. CONCLUSION

ACLR and LAS injuries are multifactorial and injury mechanisms are dependent on the task performed and the interaction between foot and shank segments. A common observation in both ACLR and LAS injury mechanisms is the coupling between longitudinal axis rotations of the shank and the subtalar joint. In dynamic, weight-bearing tasks, the triplanar movement of the foot-shank complex is governed by both the congruity of the articulating surfaces as well as the neuromuscular activity of the muscles linked to it. In this way, the stability and movements of the foot segments also effect the subtalar and talocrural joints. For this reason, strengthening the muscles acting on the foot segments may have a significant impact on ACLR and LAS injury risk factors, highlighting the importance of research into foot muscle-specific strength training as a means of reducing common lower limb ligament injuries.

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CHAPTER 5

QUANTIFYING SEGMENT COORDINATION

5.1. INTRODUCTION

The fundamental aim of this project was to use sports biomechanical principles to quantify and describe the adaptation to foot and shank movement after a foot muscle specific intervention programme in relation to ACL and LAS injury risk. Human movement is complex and the interaction between segments are intricate. Biomechanics research have adopted methods from nonlinear systems theory to describe and quantify movement relationships. Dynamical systems provide the theoretical framework in which complex systems can be described in terms of the interactions between segments (Hamill, Haddad, & Van Emmerik, 2005). The coordination between segments (or joints) is most often described using either continuous relative phase or vector coding techniques (Miller, Chang, Baird, Van Emmerik, & Hamill, 2010). Both these methods describe the coordination between segments by quantifying the change in the dynamical system over time (Miller et al., 2010). Hamill et al. (2005) suggested that one method is not necessarily superior to the other and that the choice of which measure to use depend on the type of data and the question the research aim to answer. This chapter aimed to guide the reader through the decision-making process in determining the best way to answer the specific question for this project. Data form this project was used to illustrate the decision-making process.

5.2. DYNAMICAL SYSTEMS

In physics, dynamical systems deal with the description of complex systems where the behaviour of the system as a whole is defined rather than the individual components of the system. The science of self-organization to human coordination borrows from physics and the principles of dynamical systems. The original dynamical systems research in human movement studied rhythmic bimanual finger coordination, where the index fingers of the right and left hands were moved in time with an auditory metronome (Kelso, 1984). Participants were asked to perform the task in one of two patterns: in-phase, where the two fingers moved in the same direction at the same time (e.g., both fingers flexed at the same time); and anti-phase, where the two fingers moved in opposite directions at any point in time (e.g., one finger flexed while the other extended). The subjects were then asked to increase the frequency of the oscillations. The researchers observed that the pattern spontaneously shifted from anti-phase to in-phase at a specific critical frequency- no transitions were observed in the opposite direction.

Spontaneous transitions allow for the identification of a collective variable, a quantity which provides a low-dimensional quantification of a higher dimensional system. In the case of the Kelso system, relative phase between the fingers was identified as the collective variable, providing a one-dimensional characterization of the coordination between two fingers, each of which has two degrees of freedom. The plots of the phase positions of the left and right fingers in a time series is illustrated in Figure 5.2-1, in-phase (the fingers oscillating in the same direction) (Figure 5.2-1 A and anti-phase (the fingers oscillating in the opposite direction)

(Figure 5.2-1 B). Kelso calculated the continuous relative phase (CRP) by plotting each finger's velocity, (ω) , versus its position (θ) in the phase-plane. The oscillations of each finger were normalized to the unit circle and the phase angle of each finger was calculated as the arctangent of velocity and position (ω/θ) when θ is the normalised position (Equation 5.2-1). The CRP angle is then obtained from subtracting the phase angle of the two segments at every sample $(\varphi_L - \varphi_R)$ (Figure 5.2-2) and (Equation 5.2-2) (Kelso, 1995).

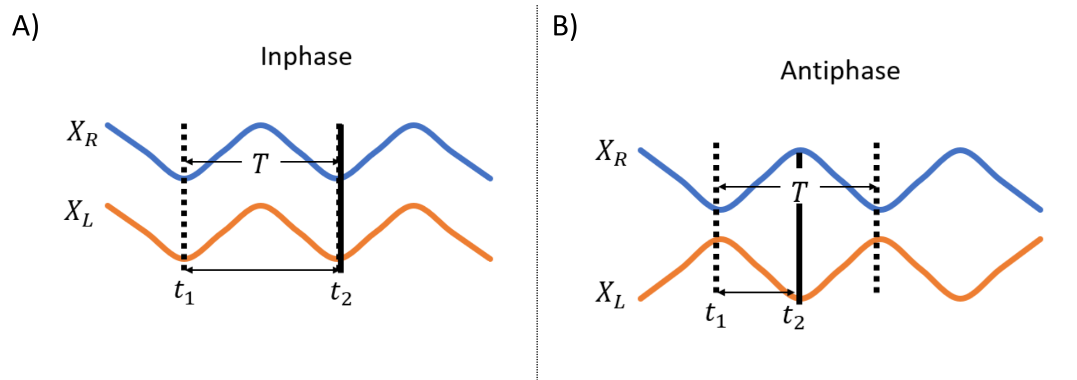


FIGURE 5. 2-1: The phase relationship between the fingers is the difference between the time of two adjacent peaks of ($t_1 - t_2$) of finger X_L divided by the period or cycle (T) of finger X_R. (A) display in-phase where the right (X_R) and left (X_L) fingers oscillating in the same direction at the same time and B) antiphase - the fingers oscillating in the opposite direction. Copied from (Kelso, 1995).

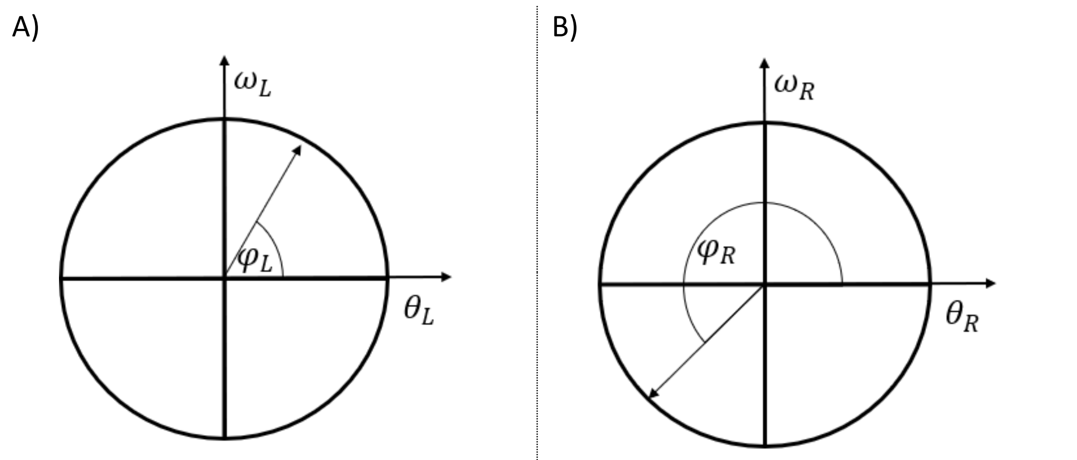


FIGURE 5. 2-2: Calculating the continuous relative phase from the phase plane trajectory. Copied from (Kelso, 1995).

$$\varphi = arctan\frac{\omega(t)}{\theta(t)}$$

EQUATION 5. 2-1: Calculating the phase angle. Where $\boldsymbol{\omega}$ and $\boldsymbol{\theta}$ is the segment velocity and angle of a segment at a specific time, respectively (Hamill, van Emmerik, Heiderscheit, & Li, 1999).

$$CRP_t = (\varphi_{L\,t} - \varphi_{R_t})$$

EQUATION 5. 2-2: Calculating the continuous relative phase angle. (Hamill et al., 1999)

To model the bimanual coordinating behaviour, Kelso turned to the physicist Hermann Haken, who had previously done pioneering work on lasers (Haken, 1983). Noting qualitative similarities between the behaviour of lasers, where the individual photons are phase locked in an in-phase pattern, and the finger movements, Haken and colleagues applied the theoretical tools developed to model lasers to the Kelso finger experiment. The result was the HKB, or Haken-Kelso-Bunz, model (Haken, Kelso, & Bunz, 1985). The two fingers were modelled as oscillators which were coupled through non-linear interactions in such a way that the transition behaviour of the fingers was reproduced in the model. While reproducing the behaviour of the system in the model was important, what was more important was that the model introduced the concept of self-organization as a way of explaining human coordination (Kelso, 1995). Briefly, self-organization posits that complex behaviour and pattern formation can arise from the interaction between components without the need to invoke a controller: control could be possible without a controller. Rather than focusing on the individual components, scientists studying self-organization focus on the nature of the coupling, or interaction, between the components. Using this approach, biological behaviour ranging from bimanual finger motion, to flocking behaviour of birds or fish, to firefly flashing, to insect nest building has been studied (Camazine et al., 2003). By placing the results of the finger experiment (Kelso, 1984) within the concept of self-organized system, Kelso (1995) was able to suggest that coordination behaviour arises spontaneously out of the interaction between the neurons in the brain, the effectors used in a task, and the environment in which the behaviour took place. This was a marked contrast to the previously dominant idea of generalized motor programs, where coordination was thought to be governed by a centrally controlled set of plans (e.g. (Schmidt, 1975)). This change of approach suggested that the key to understanding coordination behaviour was not the movement of the individual components (e.g., the fingers), but rather how the components were coupled to each other, and the resulting spatial- temporal coupling that emerged between the components.

5.2.1 Coordination dynamics in biomechanics

The concept of dynamical systems and self-organization was established in the disciplines of motor control, motor learning, and motor development by 1990, but the first application of Kelso's theory to biomechanics was by Hamill and colleagues (Hamill et al., 1999). Hamill used continuous relative phase to describe the coordination of the lower extremity in order to evaluate the gait of runners presenting with patellofemoral pain. They used the changes in the CRP to identify stability and variability of the coordination patterns and compared the variability in the coordination between runners with and without pain. A larger coordination variability was associated with a reduced risk of patellofemoral pain in runners (Hamill et al., 1999).

An example of the phase angle calculation as applied in biomechanical research of non-oscillating segments, like the fore- and rearfoot movement data from this study, is discussed and illustrated in Figure 5.2-3. When calculating the CRP for two segments that are not oscillating at the same frequency, various normalisation methods are suggested to counter for inconsistency of amplitude and frequency (Lamb & Stöckl, 2014; Peters, Haddad, Heiderscheit, Van Emmerik, & Hamill, 2003; Van Emmerik, Rosenstein, McDermott, & Hamill, 2004). In the example the segment angle (θ) and segment angular velocity (ω) was time scaled to 100 points and normalised to fit [-1,1] (Hamill et al., 1999). The segment's phase plot had the segment angle on the x-axis and segment velocity on the y-axis. The intercept of the two axes represented the mid-range of the normalised angular position and velocity. The phase angle was calculated as in Equation 5.2-1. The phase angle for each instance during the stance of the calcaneus (Figure 5.2-3 B) and metatarsal segments (Figure 5.2-3 (D)) was plotted to a four-quadrant arctangent function. The CRP was calculated by subtracting the phase angle of the distal segment (metatarsals) from the phase angle of the proximal segment (calcaneus) at each instance during stance (Figure 5.2-3 (E)). As previously mentioned, in sinusoidal movement where oscillations occur at relative similar frequency, a CRP of 0° (or 360°) represent a perfect in-phase movement and 180° perfect anti-phase movement (Hamill, Haddad, & McDermott, 2000). The negative and positive signs of the CRP indicate which segment has the greater phase angle and leading the movement. A positive CRP indicate the proximal segment having the greater phase angle and leading the distal segment and a negative CRP angle would have the opposite interpretation (Hamill et al., 1999; Lamb & Stöckl, 2014). In the example data (Figure 5.2-3) the metatarsal segment is dominant in the first 40% of the stance phase with the two segments close to perfect in-phase between ≈5 % to 20 % of stance. The metatarsal segment dominance is counterintuitive as with a visual inspection of the phase plane and phase angle plots it looks like the calcaneus segment (Figure 5.2-3 B) has a larger phase angle and larger segmental movement than the metatarsal segment (Figure 5.2-3 (D)) during the first 40 % stance.

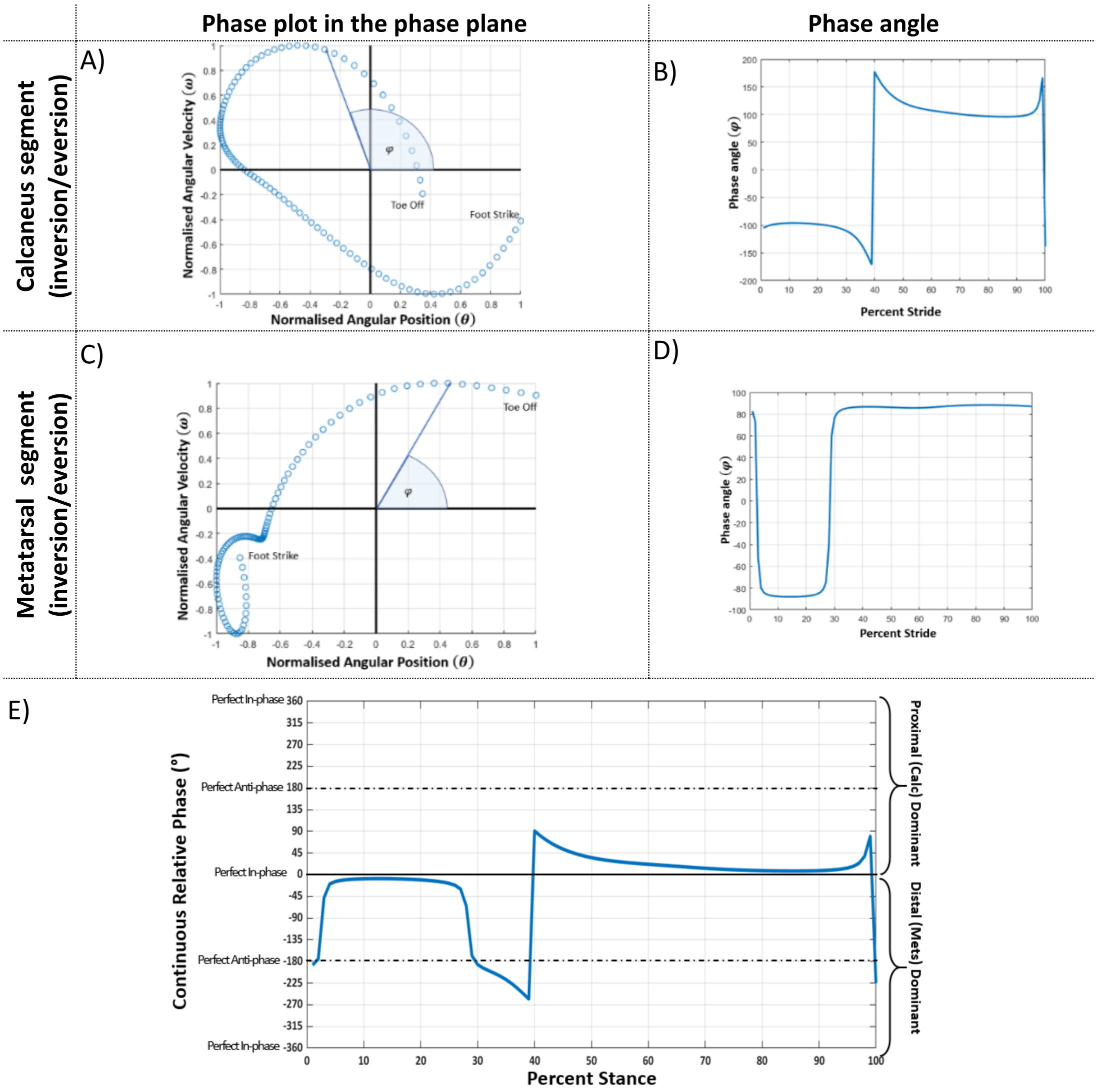


FIGURE 5. 2-3: Example of the continuous method based on the angle-angle plot of frontal plane movement of A) the calcaneus segment and C) the metatarsal with an illustration of the orientation of the coupling angle between two points. The coupling angle over the whole stance phase of B) the calcaneus and D) the metatarsal segment; E) the continuous relative phase over the complete stance phase.

Caution should be taken when interpreting CRP results using non-sinusoidal signals. The normalisation calculations can introduce artifacts and relative phase techniques are not relevant to multi-frequency couplings (Van Emmerik et al., 2004). Furthermore, attempting to interpret CRP in terms of the discrete relative phase angles can be problematic. Discrete relative phase and CRP provide different information and one should not attempt to use phase angles to explain the CRP angle (Peters et al., 2003). The purpose of CRP is not to interpret and describe the movement of the segments, but rather the nature of the coordination of the segments as in-phase or anti-phase (Lamb & Stöckl, 2014). Indeed, the Hamill et al. (1999) study that used CRP to compare the coordination variability in runners presenting with and without patellofemoral pain was repeated later using a vector coding technique (Heiderscheit, Hamill, & van Emmerik, 2002). The researchers citing limitations with using CRP with a non-sinusoidal time series producing inaccurate results.

5.3. VECTOR CODING

CRP provide qualitative information regarding the relationship between segments as in-phase or anti-phase as well as some information regarding which segment is leading the relationship (Hamill et al., 1999; Lamb & Stöckl, 2014). Inspired by the continuous relative phase methods of studying coordination, a vector coding technique was developed to provide a better description of the coordination between segments. It should be noted that while vector coding was motivated by the study of relative phase, and incorporates some of the same terminology, there is not a one-to-one mapping between vector coding and relative phase. For example, a perfect in-phase coordination pattern (i.e., with no time lags) would have a CRP angle of zero, while the angle from the vector coding techniques could range from zero to ninety degrees depending on the amplitude of movement of the two segments. Caution should thus be taken to compare coordination studies that used different methods to describe the coupling relationship between segments (Miller et al., 2010).

To determine the intricate interaction between the segments for the current study, a method was needed that would provide more detailed information. Vector coding provides additional data compared to CRP method in that it preserves the magnitude and direction of the changes between segments (Tepavac & Field-Fote, 2001). In contrast to the CRP, it allows to interpret more than just the phase relationship but the complex interaction of the individual segments from the calculated coupling angle. With vector coding techniques angle-angle plots are used to calculate and quantify the relative change in position between two segments over the entire movement cycle (Sparrow, Donovan, Van Emmerik, & Barry, 1987). Previously an encoded chain technique was used to digitise the angle-angle curve which provided a single value for each frame in the movement (Figure 5.3-1) (Sparrow et al., 1987; Tepavac & Field-Fote, 2001). The encoded chain technique projected the image onto a grid and used an 8-point scale to describe the position of a single point relative to the previous point. The encoded chain method required each point in the grid to be equally spaced (height and lengthwise) which risks the loss of temporal information. Furthermore, the method converts data that has potential directional meaning into arbitrary numbers (0 to 7) which limits interpretation of coordination and statistical analysis (Tepavac & Field-Fote, 2001).

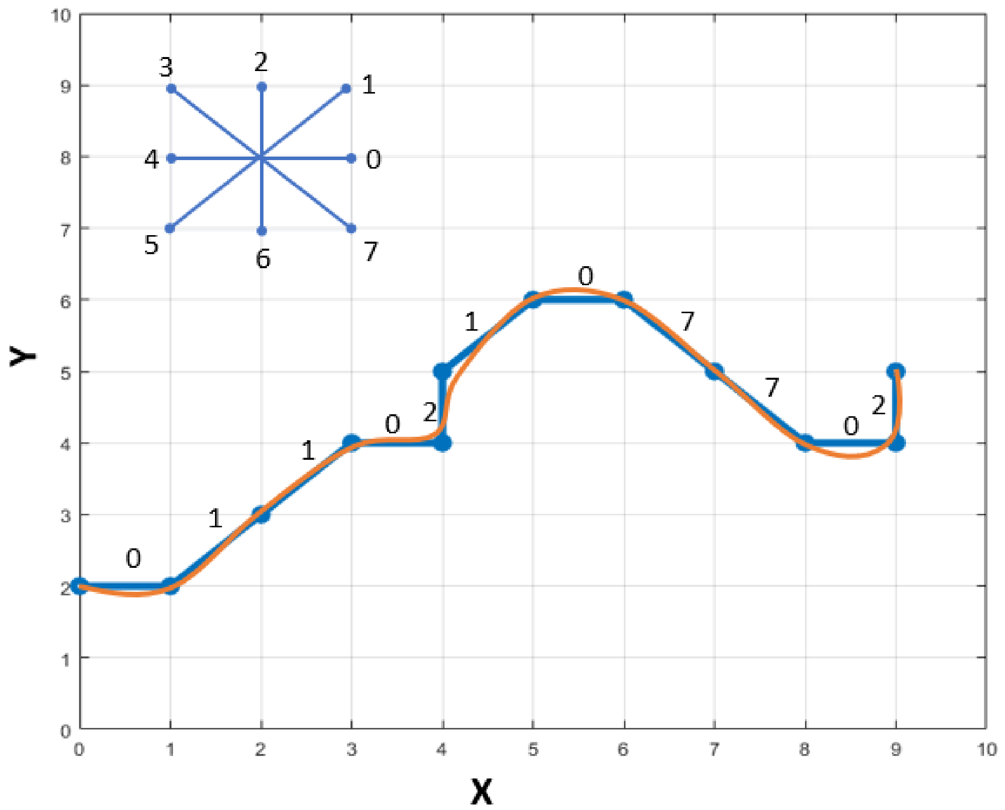


Figure 5. 3-1: Encoded chain method. The image (orange line) is projected onto a grid and the shape is encoded using an 8-point scale. Each frame is assigned a value for 0 to 7, reflecting its position relative to the position of the preceding point (Tepavac & Field-Fote,2001).

To compensate for the loss of orientation data, Sparrow et al. (1987) calculated the absolute value of right-angle between two consecutive points (Figure 5.3-2 and Equation 5.3-1). This calculated relative angle is called the coupling angle and represent the orientation of the vector between successive points. The coupling angle ranges between 0° and 360°. (Hamill et al., 2000). Directional data are classified as a circular variable, circular statistics must be used to determine the mean and standard deviation over multiple trials (Batschelet, 1981). The cosine and sine of each direction of the relative motion angle (γ) is calculated over a number of trails (n) to determine the mean direction (Equation 5.3-2) (Hamill et al., 2000).

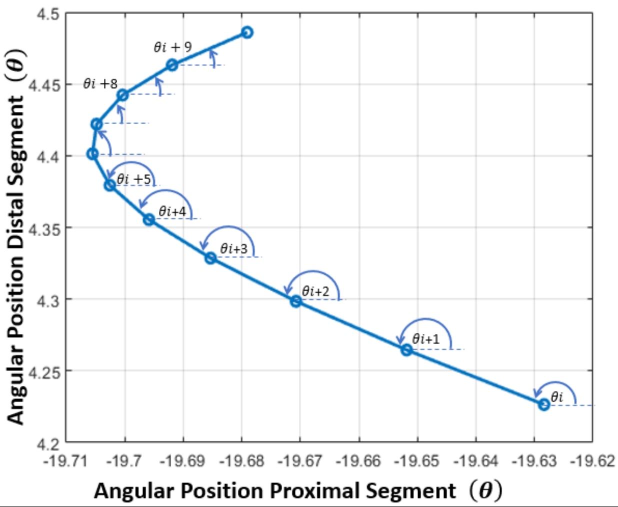


FIGURE 5. 3-2: Coupling angle calculated as the right horizontal angle between two adjoining points

$$\varphi_{i=}\ \mathrm{atan}\left(\frac{y_{i+1}-y_i}{x_{i+1}-x_i}\right)$$ Where i = 1,2,..., n-1

EQUATION 5. 3-1: Calculating the absolute value of the right-angle between two segments.

$$\overline{x_i} = \frac{1}{n}\sum_{i=1}^{n}\cos\gamma_i \text{ and } \overline{y_i} = \frac{1}{n}\sum_{i=1}^{n}\sin\gamma_i$$

Mean direction (γ): $\overline{\gamma_i} = \mathrm{atan}\left(\frac{\overline{y_i}}{\overline{x_i}}\right)$ if $\overline{x_i} > 0$

and $\overline{\gamma_i} = 180 + \mathrm{atan}\left(\frac{\overline{y_i}}{\overline{x_i}}\right)$ if $\overline{x_i} < 0$

EQUATION 5. 3-2: Calculating the mean coupling angle.

Figure 5.3-3 A illustrate the angle-angle diagram from the metatarsal and calcaneus segment using data from this thesis. In contrast to CRP calculations, where the results only indicated in-phase or anti-phase coordination, the coupling angle from vector coding provide additional information regarding the segment relationship. Similar to CRP, the coupling angles 0° (or 360°) and 180° indicate that the distal segment being static and the proximal moving, the opposite relationship is represented by 90° and 270°. In addition, the coupling angle derived from vector coding indicate that equal movement between the segments occur when the vector orientation is 45°, 135°, 225°, and 315°. At 45° and 225° the segments have similar movement and direction and at 135° and 315° the segments have equal movement, but opposite directions (Hamill et al., 2000) (Figure 5.3-4).

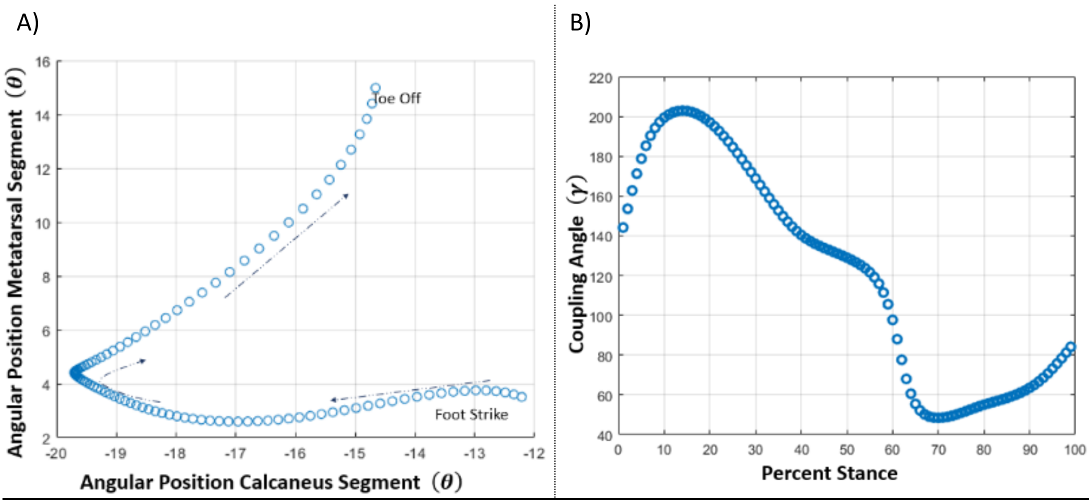
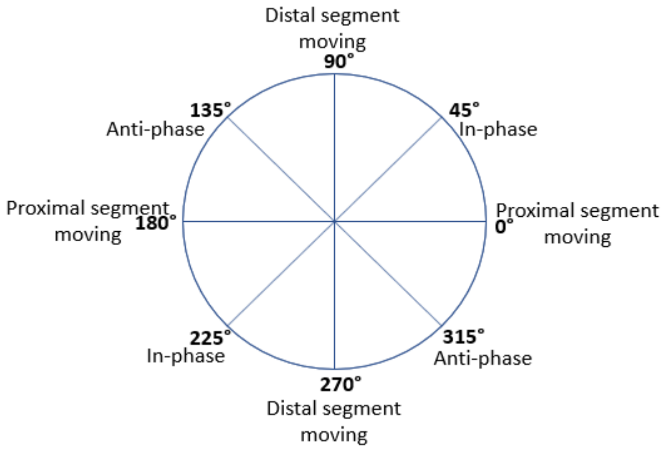


FIGURE 5. 3-3: An illustration of the continuous relative phase calculation derived from the angle-angle plot: A) is the angle-angle diagram represented by the orientation of each point of the stance phase of gait between the two segments; B) the calculated coupling angle over the whole stance phase. (Adapted from (Hamill et al., 2000).

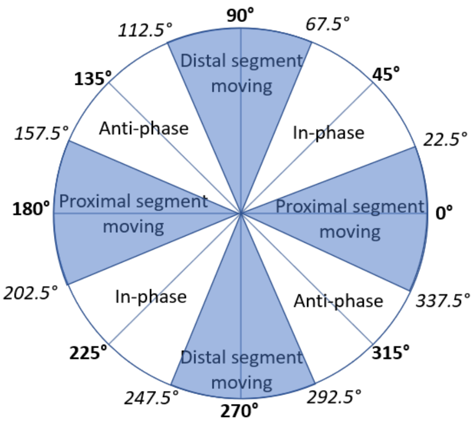


Coordination pattern	Coupling angle definitions
Anti-phase (opposite direction)	135° , 315°
In-phase (same direction)	45°, 225°
Proximal segment phase	0°, 180°
Distal segment phase	90°, 270°

FIGURE 5. 3-4: Illustration showing the coupling angle definitions and coordination pattern categories as proposed by Chang, Van Emmerik, and Hamill (2008).

5.4. Adaptations to presentation of coordination patterns

Quantification and interpretation of inter-joint and/-segment was advanced by introducing a set of operational terms and enhanced visualisation of the coupling angle (Figure 5.4-1) (Chang et al., 2008). This illustration shows the range of coupling angle in which the specific coordination phase occurs. Frequency tables were also displayed to illustrate the number of times a certain coordination pattern occurred (see Needham, Naemi, and Chockalingam (2014) for detail). The clinical value of this data is unclear as the purpose of coordination patterns is to evaluate the temporal interaction between segments over the stance phase. Needham et al. (2014) identified a lack of clarity in the mathematical equations used to calculate the coupling angle and provided a comprehensive, step by step approach to calculate the coupling angle. To quantify the coordination between the segments a modified vector coding technique was used to classify the calculated coupling angle (Equation 5.4-1 to Equation 5.4-4). Kinematic angles of two segments in the global reference frame were normalised, and time scaled to represent 100% of the gait cycle of a single stride. Angle-angle plots were created similar to (Hamill et al., 2000) . The mean coupling angle and coupling variability was calculated for each participant over five trials and across all their participants using circular statistics(Needham et al., 2014). This thesis did not investigate coupling angle variability and is therefore not discussed further. Needham et al. (2014) used circular statistics for averaging the coupling angle. Corrections as presented in Equation 5.4- 5, was used to present the coupling angle between 0° and 360°.



Coordination pattern	Coupling angle definitions
Anti-phase (opposite direction)	112.5° ≤ γ < 157.5°, 292.5° ≤ γ < 337.5°
In-phase (same direction)	22.5° ≤ γ < 67.5°, 202.5° ≤ γ < 247.5°
Proximal segment phase	0° ≤ γ < 22.5°, 157.5° ≤ γ < 202.5°, 337.5° ≤ γ ≤ 360°
Distal segment phase	67.5° ≤ γ < 112.5°, 247.5° ≤ γ < 292.5°

FIGURE 5. 4-1: The coupling angle definitions as proposed by Chang et al. (2008) to simplify the interpretation of the coordination patterns.

A new method to illustrate the coupling angle was also proposed in Needham et al. (2014). Presenting the coupling angle and segment angles in the same figure serves to simplify the interpretation of the coordination between segments, especially when readers are not familiar with dynamic systems theory. The combination of the angular movement of the two segments in the same graph as the coordination pattern makes interpretation of the coordination more intuitive. In this illustration it is easy to identify which segment has the greater range of motion in a certain time frame/coordination pattern. The segment with the greater range of motion is referred to as being dominant in the coordination. The calculations of Needham et al. (2014) were applied to the data generated in this study and presented according to their proposed illustration (Figure 5.4-2).

$$\gamma_{ij} = Atan\left(\frac{\theta_{D(j,i+1)} - \theta_{Dj.i}}{\theta_{P(j,i+1)} - \theta_{Pj.i}}\right).\frac{180}{\pi} \qquad \theta_{P(j,i+1)} - \theta_{Pj.i} > 0$$

EQUATION 5. 4-1: Coupling angle when segment angle at a certain point in time is larger than previous instance

$$\gamma_{ij} = Atan\left(\frac{\theta_{D(j,i+1)} - \theta_{Dj.i}}{\theta_{P(j,i+1)} - \theta_{Pj.i}}\right).\frac{180}{\pi} + 180 \qquad \theta_{P(j,i+1)} - \theta_{Pj.i} < 0$$

EQUATION 5. 4-2: Coupling angle when segment angle at a certain point in time is smaller than previous instance

$$\gamma_{j,i} = \begin{cases} \gamma_i = 90 & \theta_{P(i+1)} - \theta_{Pi} = 0 \; and \; \theta_{D(i+1)} - \theta_{Di} > 0 \\ \gamma_i = -90 & \theta_{P(i+1)} - \theta_{Pi} = 0 \; and \; \theta_{D(i+1)} - \theta_{Di} < 0 \\ \gamma_i = -180 & \theta_{P(i+1)} - \theta_{Pi} < 0 \; and \; \theta_{D(i+1)} - \theta_{Di} = 0 \\ \gamma_i = undefined & \theta_{P(i+1)} - \theta_{Pi} < 0 \; and \; \theta_{D(i+1)} - \theta_{Di} = 0 \end{cases}$$

EQUATION 5. 4-3: Conditions applied to the coupling angle

$$\gamma_{j,i} = \begin{cases} \gamma_{i,j} + 360 & \gamma_{i,j} < 0 \\ \gamma_{i,j} & \gamma_{i,j} \geq 0 \end{cases}$$

EQUATION 5. 4-4: Corrected coupling angle to represent a value between 0° and 360°.

$$\bar{\gamma_i} = \begin{cases} \operatorname{atan}\left(\frac{\bar{y}_i}{\bar{x}_i}\right).\frac{180}{\pi} & x_i > 0, y_i > 0 \\ \operatorname{atan}\left(\frac{\bar{y}_i}{\bar{x}_i}\right).\frac{180}{\pi} + 180 & x_i < 0 \\ \operatorname{atan}\left(\frac{\bar{y}_i}{\bar{x}_i}\right).\frac{180}{\pi} + 360 & x_i > 0, y_i < 0 \\ 90 & x_i = 0, y_i > 0 \\ -90 & x_i = 0, y_i < 0 \\ undefined & x_i = 0, y_i = 0 \end{cases}$$

EQUATION 5. 4-5: Corrections applied to average coupling angle to present the angle between 0° and 360°.

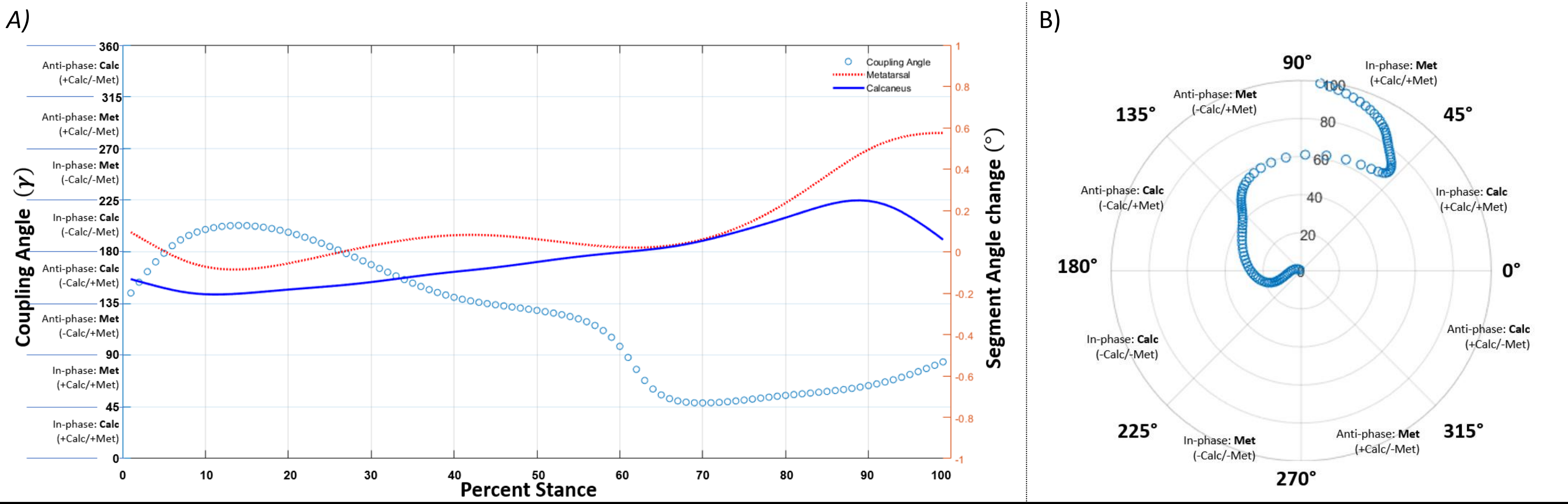


FIGURE 5. 4-2: A) Illustration using the proposed guidelines of Needham 2014. The change in the individual segment angles (red - metatarsals and blue - calcaneus solid lines) and coupling angle (dotted line) is displayed on the same graph. This method aids in visualising of the individual segment movement in the coordination pattern. B) The coupling angle on a polar plot as in Chang et al. (2008).

Further advancements to the classification of the coordination pattern were done in Needham, Naemi, and Chockalingam (2015). Here they expanded on the concept of segment range of motion and segment dominancy to improve interpretation of segment interaction in the coordination pattern. The purpose of the adaptation was to illustrate the overlap in phase dominancy (in-phase or anti-phase) and segmental dominancy (distal (D) or proximal (P)). Thus, expansion on dominancy in in-phase or anti-phase coordination was made. The presentation of polar plot presented in Chang et al. (2008) was advanced by adding segment dominancy phases to coordination phases and illustrating the direction of rotation of the segments (Figure 5.4-3). The direction of rotation of the segment is determined by the polar position of the coupling angle and the rectangular coordination of the axis. Positive and negative rotations follow the right-hand rule of axis rotation convention. Where (+) rotation indicate extension, adduction, and inversion and (–) rotation flexion, abduction, and eversion. Proximal and distal segment dominancy was presented as a percentage. A unit circle consists of 400 gradians each quadrant thus presents 100 gradians conveniently matching 100 percentage points. Segment contribution or dominancy are interpreted as having equal (D50 – P50) dominance at a coupling angle of 45° and rotating in the same direction. The segment is considered dominant when the percentage contribution to the coordination patterns exceeds 50. Complete dominance is achieved at 0°–360°/180° (D0–P100) or 90°/270° (D100–P0).

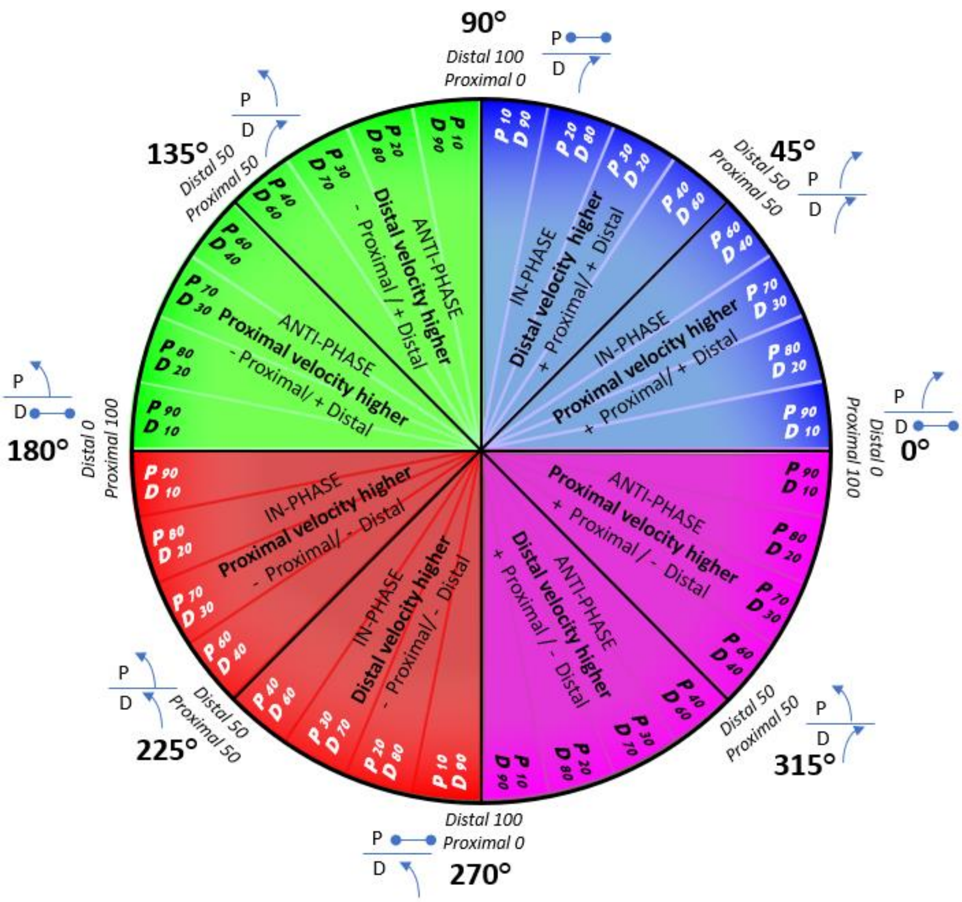


FIGURE 5. 4-3: The coordination pattern classification proposed in Needham et al. (2015). Segment dominancy and percentage contribution to the coordination patters is shown around the circumference of the polar plot. The arrows show the direction of the movement of the proximal and distal segments in the coordination pattern.

In 2020 Needham presented a novel data visualisation technique that made use of colour and data bars to illustrate amongst other variables the coordination patterns, segment dominancy and range of motion on a single graph (Needham, Naemi, Hamill, & Chockalingam, 2020). The term “coupling angle mapping” is introduced and each colour represents a specific coordination pattern (Figure 5.4-4 A & B). The coupling angle diagram is shown in a bar graph with each time point represented by the colour based on the polar position of the coordination classification. The height of the bars in the graph indicated segment dominancy. Superimposed in the bar graph was also the quantified range of motion of the dominant segment as “inter-data point range of motion” (solid line). Displaying the data as bar graphs instead of frequency plots aids to visualise and interpret the change in coordination and segment interaction over time. Colour mapping improves data visualisation and interpretation (Silva, Santos, & Madeira, 2011). However, segment dominance within the phase relationship was difficult to visualise as segment dominance was indicated with different shades of the same colour (Needham et al., 2020). Shading may also create the illusion that one phase is more desirable than the other or somehow scaled (darker colours present higher value in the field of data visualisation) (Silva et al., 2011).

In this thesis the coordination pattern was displayed using bar graphs and coordination pattern classifications similar to Needham et al. (2020). However, segment dominance is displayed on the vertical axis with the proximal segment on the top of the horizontal axis and the distal segment on the bottom. The mid-point of the vertical axis represents equal contribution (D50-P50) with proximal contribution increasing on the top (D00-P50) and the distal segment on the bottom (D50-P00) of the horizontal axis. This representation simplifying the interpretation of the coordination pattern and changes between segment dominancy. Furthermore, no shading is needed to indicate segment dominance as the dominant segment is displayed separately rendering interpretation more intuitive (**Error! Reference source not found.**Figure 5.4-4 A & B).We displayed individual segment movement as separate figures as to not clutter the bar graph and complicate interpretation.

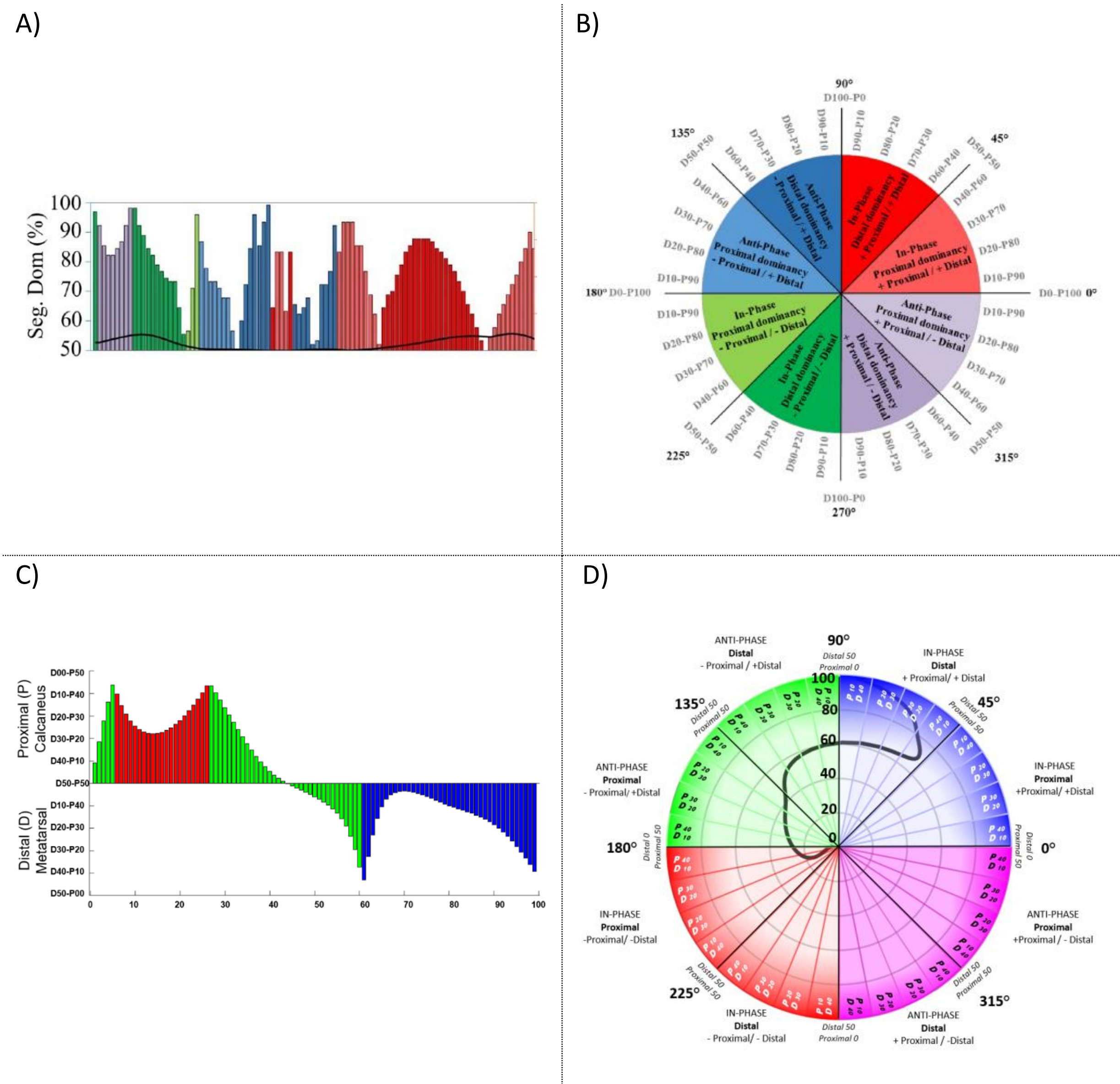


FIGURE 5. 4-4: A) bar graph displaying coupling angle mapping, segmental dominancy, and change in the range of motion (solid line) profiling of the dominant segment. B) The colour scale legend for each classification. Segment dominancy for each quadrant is displayed around the circumference. (Copied from (Needham et al., 2020). C) The illustrations used in this thesis derived from (Needham et al., 2020). The proximal segment is displayed on the top and the distal segment on the bottom. When the bar graph is displayed on the top, the proximal segment is dominant on the bottom when the distal segment is dominant. D) The colour legend used for each classification. The segment dominancy is displayed in the wedges around the circumference. The coupling angle (solid line) is added in the polar plot to aid interpretation of the coordination pattern.

5.5. ADAPTATIONS TO CALCULATING THE COUPLING ANGLE

Vector coding calculations are simple as it needs no complicated normalization controls as in the continuous relative phase method (Hamill et al., 2000). Furthermore, the results are intuitive and easy to interpret (Needham et al., 2020). However, the vector orientation is calculated from angle-angle plots, only providing directional information. As such valuable information may be lost as no inferences can be made about the temporal information that exist in the coupling relationship (Hamill et al., 2000).

Excessive segment rotational velocity increases the risk both ACLR (Dowling, Favre, & Andriacchi, 2012; Nagano, Ida, Akai, & Fukubayashi, 2009) and LAS (Chu et al., 2010). Understanding segment velocity and the influence that segment angular velocity may play in coordination may therefore be vital in injury prevention research. Indeed, earlier research found a strong correlation between increased cortical brain activity and segment velocity (Kelso et al., 1998). The researchers studied cortical brain activity during flexion-extension tasks in-sync or out of sync with a metronome. They found that cortical brain activity increased as the velocity of the flexion-extension movement increased while the finger movement synchronised with the metronome (see Kelso et al. (1998) for more detail on the experiment). Furthermore, they found that the peak velocity of the movement preceded the peak cortical activity, indicating the significance of afferent activity from the muscle to the brain. The Kelso et al. (1998) experiment showed the brain is more sensitive to the velocity of movement, interpreting and reacting to the movement than it is to the position of the segment, reiterating the importance of segment velocity in injury prevention research.

Using segment velocity in calculating the coupling angle instead of segment angles also served an additional purpose. In biomechanical research segment angles are obtained from the segment's X-Y-Z Cardan sequence coordination in relation to the laboratory's X-Y-Z coordination system. This convention works well when the segment's reference frame is perfectly aligned with the laboratory's global reference frame in tasks like vertical jumps and straight-line walking or running tasks. However, for the purpose of this thesis we made use of change of direction tasks. The segment's local reference frame was therefore no longer aligned with the global reference frame, rendering segment angles unreliable. Using segment velocities allows the analysis of the change of motion in the segment's local reference frame. As a result of using segment velocities to calculate the coupling angle, adaptations were made calculating the coupling angle (Figure 5.5-1) (Equation 5.5-1, Equation 5.5-2, Equation 5.5-3). The modifications made to calculating the coupling angle meant that coupling angle represented the velocity-acceleration rather than position-velocity relationship between segments. The coordination pattern classifications are similar to (Needham et al., 2020) (Table 5.5-1). Interpretation of velocity-acceleration coordination patterns are also easy and intuitive.

As in positional data a change in movement (velocity) was either positive or negative. The change in velocity obeyed the right-hand rule where a positive change in velocity indicates the change towards the positive direction (anti-clockwise) and a negative velocity a change towards a clockwise direction. In our example, a positive change in velocity for the calcaneus indicates either an acceleration in a positive direction around the axis (increasing inversion velocity) or deceleration of the negative rotational direction around the axis (decreasing eversion velocity). To make the interpretation to coordination easier we also included the velocity signals in the illustrations (Figure 5.5-2).

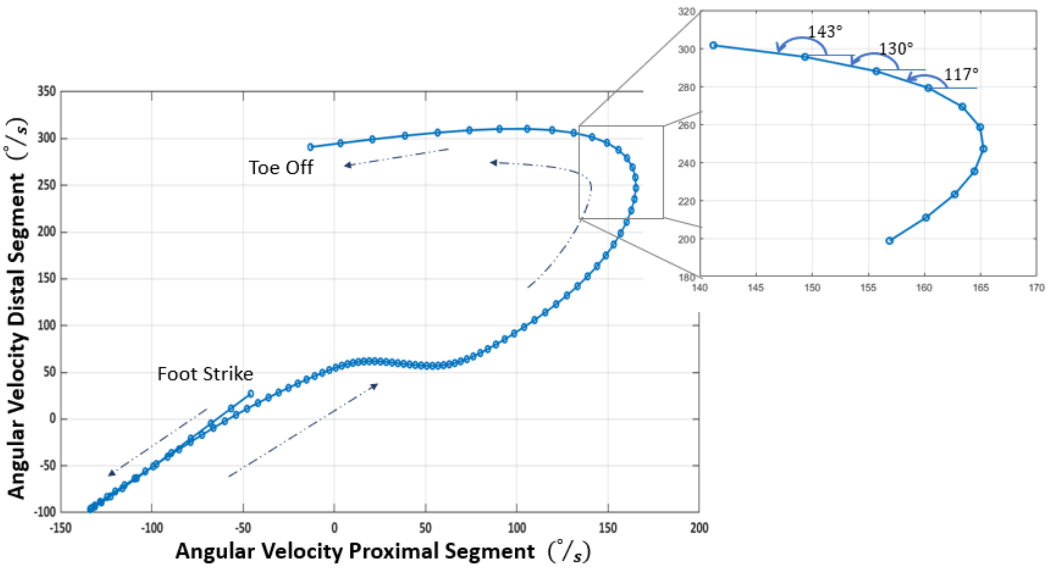


FIGURE 5. 5-1: An example of a relationship plot for the metatarsal/calcaneus velocities. The arrows indicate the direction of movement. The inset explains the coupling angle between three successive points .

$$\gamma_{ij} = Atan\left(\frac{\omega_{D(j,i+1)} - \omega_{Dj.i}}{\omega_{P(j,i+1)} - \omega_{Pj.i}}\right) . \frac{180}{\pi} \qquad \omega_{P(j,i+1)} - \omega_{Pj.i} > 0$$

EQUATION 5. 5-1: Coupling angle when segment velocity at a certain point in time is larger than previous instance

$$\gamma_{ij} = Atan\left(\frac{\omega_{D(j,i+1)} - \omega_{Dj.i}}{\omega_{P(j,i+1)} - \omega_{Pj.i}}\right) . \frac{180}{\pi} + 180 \qquad \omega_{P(j,i+1)} - \omega_{Pj.i} < 0$$

EQUATION 5. 5-2: Coupling angle when segment velocity at a certain point in time is smaller than previous instance.

$$\gamma_{j,i} = \begin{cases} \gamma_{i,j} + 360 & \gamma_{i,j} < 0 \\ \gamma_{i,j} & \gamma_{i,j} \geq 0 \end{cases}$$

EQUATION 5. 5-3: Corrected coupling angle.

TABLE 5. 5-1: Coordination pattern classifications used in this thesis. Adapted (Needham et al., 2015).

Coordination pattern	Coupling angle	Coupling direction
In-phase (*Proximal Δvelocity larger*)	0° to 45° (blue – top) 180° to 225° (red – top)	Both segments positive Δvelocity (0° to 45°), Both segments negative Δvelocity (180° to 225°)
In-phase (*Distal Δvelocity larger*)	45° to 90° (blue – bottom) 225° to 270° (red – bottom)	Both segments positive Δvelocity (45° to 90°), Both segments negative Δvelocity (225° to 270°)
Anti-phase (*Distal Δvelocity larger*)	90° to 135° (green – bottom) 270° to 315° (magenta – bottom)	Distal positive Δvelocity and proximal negative Δvelocity (90° to 135°) Distal negative Δvelocity and proximal positive Δvelocity (270° to 315°)
Anti-phase (*Proximal Δvelocity larger*)	135° to 180° (green – top) 315 to 360° (magenta – top)	Proximal positive Δvelocity and distal negative Δvelocity (90° to 135°) Proximal negative Δvelocity and distal positive Δvelocity (270° to 315°)
In-phase (*Equal Δvelocity*)	45° and 225°	Distal and proximal positive Δvelocity (45°) Distal and proximal negative Δvelocity (225°)
Anti-phase (*Equal Δvelocity*)	135° and 315°	Proximal negative Δvelocity and distal positive Δvelocity (135°) Proximal positive Δvelocity and distal negative Δvelocity (315°)
Proximal Δvelocity only	0°, 360°, and 180°	Proximal positive Δvelocity only (0°/360°) Proximal negative Δvelocity only (180°) (distal segment does not move)
Distal Δvelocity only	90° and 270°	Distal positive Δvelocity only (90°) Distal negative Δvelocity only (270°) (the proximal segment does not move)

Note: Positive Δvelocity for a segment indicates the change of directional rotation towards the positive direction (anti-clockwise) around the axis (e.g., accelerating shank internal rotation or decelerating shank external rotation). Negative Δvelocity indicates a change of directional rotation towards the negative direction (clockwise) around the axis (e.g., accelerating shank external rotation or decelerating shank internal rotation).

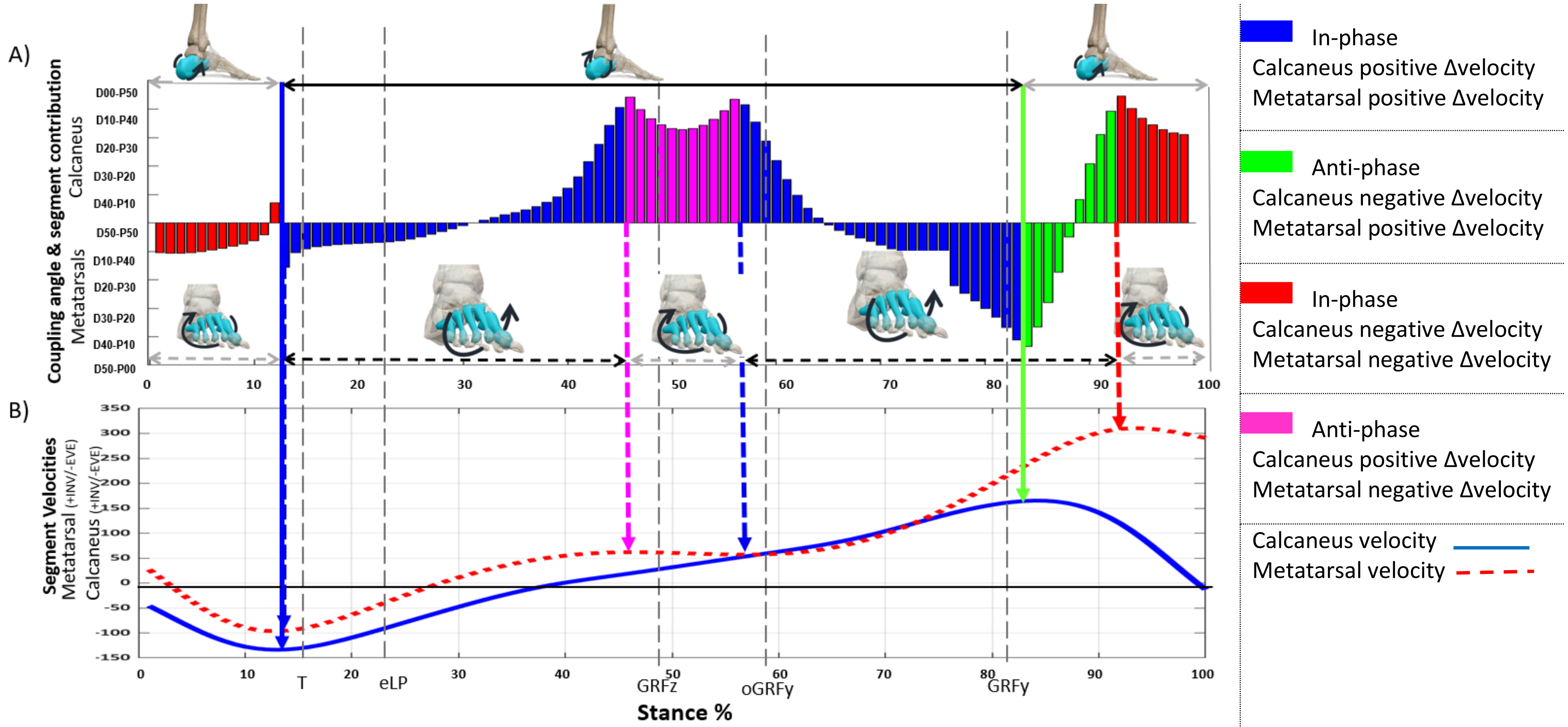


FIGURE 5. 5-2: The illustrations used in this thesis. The example is of the A) CalcaneusY (+INV/-EVE)/MetatarsalY (+INV/-EVE), coordination pattern and B) the CalcaneusY (+INV/-EVE)/MetatarsalY (+INV/-EVE), segmental velocities.

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CHAPTER 6

DESCRIBING CALCANEUS-SHANK COUPLING DURING UNANTICIPATED CHANGES OF DIRECTION: THEORETICAL IMPLICATIONS FOR PROPHYLACTIC ACL STRATEGIES

PUBLICATION

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6.1. ABSTRACT

BACKGROUND AND OBJECTIVES

Shank rotation is associated with increased risk of lower limb injuries. Straight-line running injury preventions propose a 'bottom-up' approach due to the distal-proximal coupling relationship between rearfoot- and shank rotations observed in previous research. The coupling relationship between the calcaneus and shank, in movements associated with anterior cruciate ligament (ACL) injury risk, is unknown and thus the advisability of a similar 'bottom-up' approach to ACLR injury prevention. Our aim was to test the hypothesis that a distal-proximal coupling relationship exists between the calcaneus and shank during unanticipated changes of direction.

METHOD

We implemented a modified vector coding technique using segmental velocities in a local, anatomical reference frame to calculate the coordination relationship (coupling angle) between the individual tri-planar calcaneus movements and transverse plane shank movements.

Results

During the loading phase, a distal-proximal coupling relationship between calcaneus eversion deceleration, abduction acceleration, and shank internal rotation deceleration was observed. In contrast, a proximal-distal coupling relationship between the sagittal plane calcaneus and transverse plane movement of the shank was identified in the same period.

Conclusions

The proximal-distal coupling found between the calcaneus and shank justifies exploring the manipulation of the calcaneus to modulate the transfer of rotation to the knee during unanticipated change of direction tasks associated with ACL injury risk.

KEY WORDS

unanticipated change of direction, coupling angle, coordination pattern

6.2. INTRODUCTION

A non-contact anterior cruciate ligament rupture (ACLR) is a disruptive injury incurring substantial costs and its rehabilitation can be a lengthy and complicated process (Cassell, 2012; Forsdyke, Smith, Jones, & Gledhill, 2016). In sports, 'high risk' movements associated with ACL injury include a trunk that is laterally flexed and rotated away from the stance leg, an adducted and internally rotated hip, and an internally rotated shank (Gamada, 2014; Stuelcken, Mellifont, Gorman, & Sayers, 2016). This configuration of the trunk and lower limb can result in a substantial knee valgus angle, which can stress the ACL in the presence of a considerable resultant ground reaction force and create a potentially injurious valgus moment (Powers, 2010). Consequently, ACL injury prevention programmes integrate multimodal strategies to decrease the development of 'high risk' movement patterns that result in large rotational and translational forces at the knee (Noyes & Barber-Westin, 2015). Current ACL prophylactic training programmes mainly aim to modulate the transfer of proximal-to-distal movement by strengthening the core, hip, and posterior chain muscles with varying degrees of success (Noyes & Barber-Westin, 2014; Paszkewicz, Webb, Waters, McCarty, & Van Lunen, 2012; Shultz et al., 2015; Yoo et al., 2010). Comparatively, little emphasis has been placed on modulating the transfer of distal-to-proximal rotational motion to the knee through the foot and shank. There are reasons, however, to believe that distal-to-proximal transfer could play a role in reducing the risk of ACL injuries.

Previous research has shown a coupling relationship between subtalar joint pronation/supination and shank internal/external rotation, respectively (Hicks, 1953; Huson, 1991). A more recent in vitro study found that frontal and, to a greater extent, transverse plane rotations of the calcaneus were transferred to transverse plane rotation of the talus. The congruity of the talo-crural joint relays the transverse plane rotation of the calcaneus to the internal/external rotation of the shank, suggesting a mechanism for overuse knee injury (Fischer, Willwacher, Arndt, & Brüggeman, 2018). In vivo studies also suggest a transfer of rotation from the foot to the shank during dynamic movements, possibly increasing rotational forces at the knee and the risk of injury. Prominent researchers in the field of lower limb injury prevention have indeed suggested a distal-to-proximal approach to injury prevention, which entails training the muscles acting on the foot and ankle (Nigg, Baltich, Federolf, Manz, & Nigg, 2017). However, the 'bottom-up' approach to injury prevention was based on straight-line running. It may be tempting to assume the same distal-proximal coupling relationship and risk to knee injury exists in activities other than straight-line running. Indeed, previous research found an association between dynamic rearfoot eversion angles during high intensity change of direction, drop landing and static squat tasks, and increased knee valgus angles, which could potentially increase ACL injury risk (Kagaya, Fuji, & Nishizono, 2013; McLean, Lipfert, & Van Den Bogert, 2004; Peel, Thorsen, Schneider, & Weinhandl, 2020). Although a link was made between rearfoot movement and increased knee valgus, it is unclear whether it was due to the distal-proximal coupling relationship. Function of the foot muscles and coupling relationship between segments differ as intensity, foot strike angle, and direction of progression change (Pohl, Messenger, & Buckley, 2007; Weir, Jewell, Emmerik, & Hamill, 2017). It is thus unclear what the coupling relationship between the foot and longitudinal shank rotation is during dynamic, unanticipated cutting movements associated with ACL injury risk. Quantifying the nature of the coupling relationship between the foot and shank may provide an understanding of how controlling foot function may impact force transmission to the knee during dynamic movements.

Coupling properties and coordination patterns between segments in a time series provide insight into the interaction between segments. Describing joints in isolation using traditional biomechanical analysis (e.g., joint angles, displacements, and velocities) provides limited information about the singular joint's attributes at a specific point in time. Functional data analysis methods are often used in studying the joint function across a movement of interest as these approaches consider the time-dependent structure of biomechanical data (Hébert-Losier et al., 2015). However, these statistical methods do not provide information about the multifaceted interactions between segments. The dynamic coordination between segments can be studied by means of non-linear, continuous relative phase or vector coding techniques (Glazier, 2006; Hamill, van Emmerik, Heiderscheit, & Li, 1999). More recent approaches, such as the modified vector coding technique (Chang, Van Emmerik, & Hamill, 2008; Needham, Naemi, & Chockalingam, 2014), enable the quantification and analysis of movement relationships, direction of movement of individual segments, and magnitude of movement of segments in relation to others. A further advantage of the modified vector coding technique is that information regarding the relative contribution of each segment is determined (Rodrigues, Chang, TenBroek, van Emmerik, & Hamill, 2015). The segment with the larger contribution may be interpreted as leading or driving the movement at a specific time (Needham et al., 2014).

The purpose of this study was to use the advances in describing and quantifying the temporal coupling relationship between segments to investigate the coordination patterns between the shank and calcaneus. We hypothesised that a distal-to-proximal coupling relationship would emerge between the tri-planar movement of the calcaneus and the longitudinal rotation of the shank during dynamic, unanticipated change of direction tasks. A modified vector coding technique was used to quantify the nature of the coupling relationship between the movements of the calcaneus and shank, as stated. Establishing the coupling direction between the calcaneus and the shank during unanticipated movements provided the premise to further investigate whether controlling calcaneus movement can modulate the distal-proximal transfer of rotation to the knee, potentially supplementing prophylactic ACL programmes.

6.3. METHODS

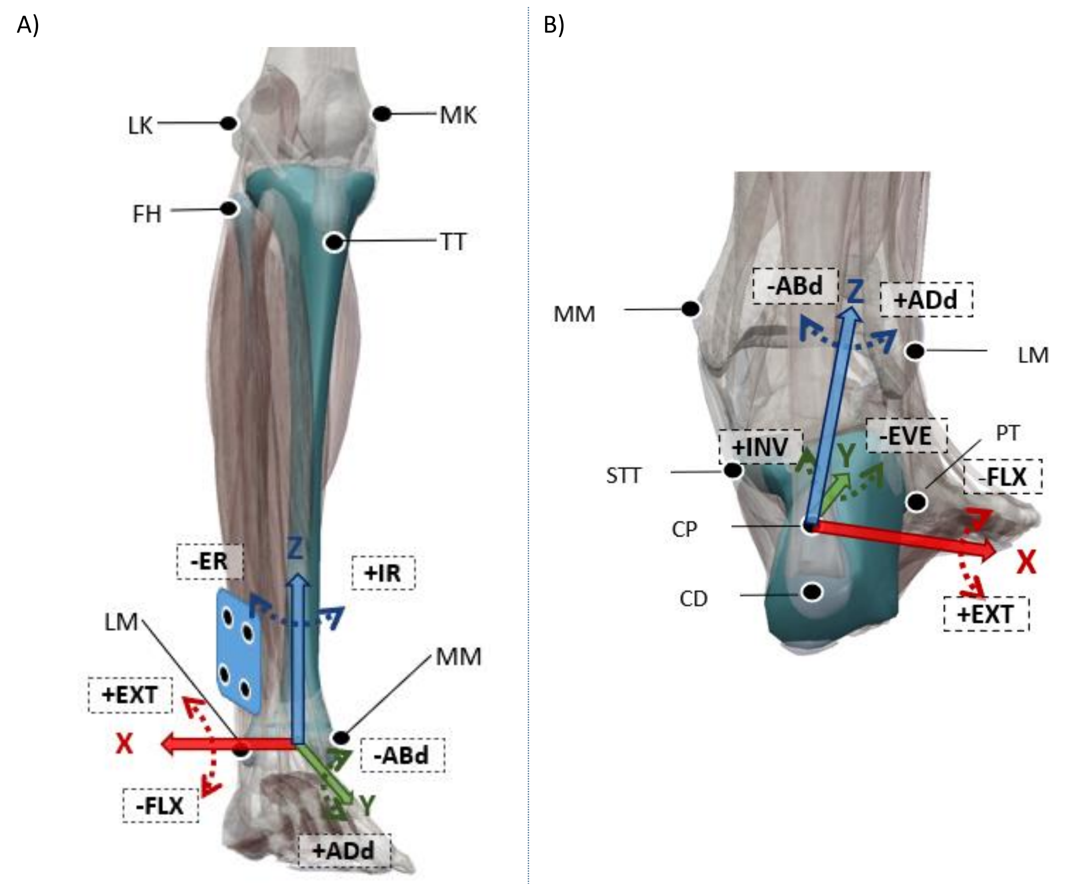
6.3.1 PARTICIPANTS

Eighteen female athletes (age: 17.4 ± 1.7 years; height: 1.68 ± 0.06 m; mass: 65.53 ± 8.96 kg) gave written informed consent to take part in this study. All athletes were skilled court sports athletes (netball, volleyball, tennis, badminton, and squash), and had sufficient skill to perform 45° changes of direction at running speed (4.5 ± 0.47 $m.s^{-1}$). All athletes were injury- and pain-free for the six months preceding study participation and had no musculoskeletal or neurological conditions affecting movement or foot function at the time of testing. Ethical approval was sought and received from the University Human Ethics Committee of our institution (SOA 16/77).

6.3.2 DATA COLLECTION

6.3.2.1 MARKER PLACEMENT

The Leardini-multisegmental foot model was used to track the motion of the lower extremity segments (Leardini et al., 2007). The Leardini-multisegmental foot model has good inter-trial and between-day coefficient of multiple correlation scores and results from this model are therefore repeatable and reliable (Caravaggi, Benedetti, Berti, & Leardini, 2011). Small retro-reflective markers (8 mm) were used for the foot. In addition, larger markers (10 mm) were used in a rigid set of 4 markers placed bilaterally on the lateral side of the thighs and shanks (Figure 6.3-1). Single markers were also placed bilaterally on the anterior superior iliac spines, posterior superior iliac spines, greater trochanters, head of the fibula, tibial tuberosities, and medial and lateral femoral epicondyles. In the current study, only the coordination between the shank (tibia and fibula) and calcaneus segments was of interest.



Landmarks	Local reference frame	Rotation	Landmarks	Local reference frame	Rotation
TT: tibial tuberosity FH: fibular head MM: medial malleolus LM: lateral malleolus	**Z-axis** (vertical/longitudinal along tibia): The projection of the line originating at the midpoint between LM and MM and joining the origin and TT on the frontal plane. The frontal plane was described as the plane joining MM, LM and TT. -external rotation (-ER)	+internal rotation (+IR) -external rotation (-ER)	STT : sustentaculum tali PT: peroneal tubercle CD: distal calcaneus CP: proximal calcaneus	**Y-axis**: The line between CP and the midpoint between STT and PT, and originating at CP -eversion (-EVE) /	+Inversion (+INV)
	Y-axis (direction of progression): Orthogonal to Z-axis and in the frontal plane	+adduction (+ADd) -abduction (-ABd)		**X-axis**: The line normal to the Y-axis and parallel to the plane defined by CP, PT, and STT, originating at CP	-flexion (-FLX) / +extension (+EXT)
	X-axis (medio-lateral): Orthogonal to the Z- and Y-axis	+extension (+EXT) -flexion (-FLX)		**Z-vertical axis:** Orthogonal to the X and Y axes, originating at CP	-abduction (-ABd) +adduction (+ADd)

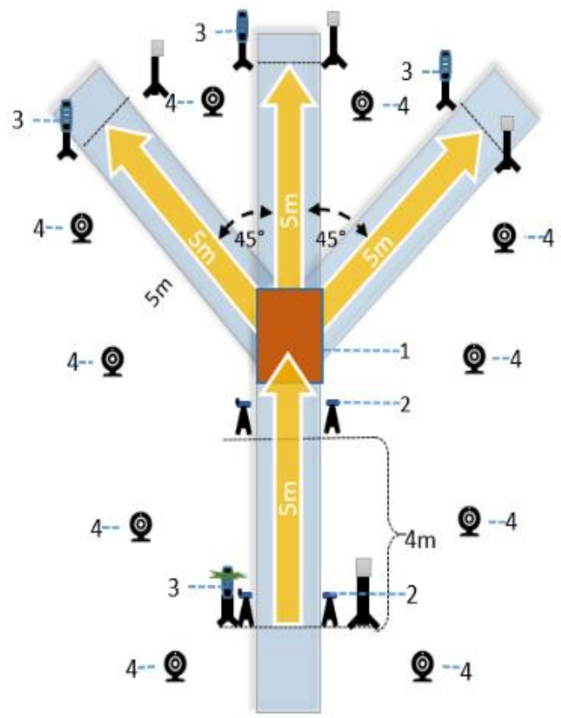
FIGURE 6. 3-1: Bony landmarks used to define (A) Right shank and local reference frame orientation, (B) Right calcaneus in the local reference frame. Adapted from previous models (Cappozzo, Catani, Della Croce, & Leardini, 1995; Leardini et al., 2007). Images courtesy of visible body (Body, 2019).

6.3.2.2 EXPERIMENTAL PROTOCOL

Athletes were required to perform unanticipated 45° change of direction tasks on their dominant leg while running barefoot. Leg dominance for each participant was determined as the leg freely chosen to step on to a 45 cm high bench (Hoffman, Schrader, Applegate, & Koceja, 1998). Only data from the dominant leg were captured given that the dominant limb is at a higher risk of injury (Gamada, 2014; Hoffman et al., 1998).

Kinematic and kinetic data were concurrently collected using a ten camera (six Oqus 3-series and four Oqus 7-series) Qualisys system (QTM software v2.17, Qualisys AB, Gothenburg, Sweden) in conjunction with a force platform (Kistler, 569x DAQ, Winterhur, Switzerland). The sampling rates of the kinematic and kinetic data were both 200 Hz. The force platform was used to identify initial foot strike and toe-off events.

Athletes had a 5 m run-up to the force platform (measured from the starting timing gate to the middle of the force platform). Approach speed was monitored using a photocell timing system (Brower Timing System, USA) to ensure locomotive speeds were within 10% of the selected 4.5 m.s^{-1}, similar to speeds used in other research involving change of direction tasks (Alenezi, Herrington, Jones, & Jones, 2016; Spiteri, Cochrane, Hart, Haff, & Nimphius, 2013). Three runways indicated the direction and angle of the running and cutting tasks, respectively (Figure.6.3-2). Three sets of electronic timing gates (Smart Speed, Fusion Sport, Queensland, Australia) were placed at the end of each runway. The Smart Speed timing gate system was pre-programmed to indicate a random direction (left, right, or straight ahead) by means of a flashing light, 0.7 s after the approach run was initiated. Athletes were required to perform five successful trials for each randomly indicated direction to maintain the unanticipated demand of the task. A trial was deemed successful if the approach run was within the desired range of the approach speed, force platform was struck within the boundaries of the four transducers with the dominant foot, and the cut was performed towards the indicated direction. Athletes rested for at least 60 seconds between trials to minimize the potential for fatigue.



1. Force platform – identify initial foot strike and toe-off events
2. Brower timing gates – monitor approach speed
3. SmartSpeed timing gates – random flashing light to indicate directional change
4. 10 Qualisys cameras – motion capture

FIGURE 6. 3-2: Experiment set-up.

6.3.3 DATA PROCESSING

The stance phase was recorded from the instant the vertical ground reaction force (GRFz) exceeded 20 N (initial foot strike, iFS) to when it dropped below 20 N (toe-off, TO) (Pfile et al., 2013). The GRFz impact transient was identified as the rapidly developing spike in the GRFz curve, occurring within the first 50 ms of stance (Lieberman et al., 2010; Whittle, 1999). The loading phase was defined as the time between initial foot strike and the average occurrence of the first reduction in the GRFz data. The anterior-posterior braking phase was identified as the period where the anterior-posterior ground reaction force (GRFy) curve was negative and the propulsion phase where GRFy was positive.

Marker trajectories were interpolated for missing signals with the Qualisys Track Manager software, allowing a maximum of 10 frames (50 ms) of gap filling. The interpolated signals were exported to Visual 3D™ (v5.0, C-Motion, USA) and smoothed using a fourth-order low-pass Butterworth filter with a 10 Hz cut-off frequency. Signals were time-scaled and normalized to 100 % of the stance phase. Visual 3D™ was used to filter the analogue signals (fourth-order low-pass Butterworth filter with a cut-off frequency of 25 Hz) used to calculate the ground reaction forces. Visual 3D™ was used to calculate the segmental velocities relative to the lab and expressed in the segment's coordinate system. The segment velocity signals were exported to MATLAB (R2018a, Mathworks, MA USA) and used to calculate the coupling angles. MATLAB was also used to create diagrams and plots, and to identify GRF events throughout the stance phase.

6.3.4 Calculating the coupling angle

To simplify the interpretation of the coupling relationship between the calcaneus and shank only the trials where the cut was executed away from the stance leg was analysed. A modified vector coding system was adapted to analyse the motion patterns between the shank and calcaneus segments (Chang et al., 2008). The adaptation of segmental rotational velocities, rather than segmental angles, in a local reference frame, allows for the description of segmental rotations around their own anatomical axis system and in relation to each other. The original vector coding (Chang et al., 2008; Hamill et al., 1999) was applied to straight line movement and was therefore able to meaningfully analyse segment motions in a global coordinate system. To study motions with a change in direction, where the global coordinate system was no longer aligned with the body, we made two modifications. First, we used a local coordinate system, examining segment motions within the local coordinate system of the segments. Second, because segment position within its own coordinate system is, by definition, fixed, we analysed the segment motion using segment angular velocities, which are not constrained to be fixed. The latter modification means the data were analysed as velocity-acceleration rather than position-velocity.

The segmental rotations were interpreted using the X-Y-Z Cardan sequence corresponding to the segment's coordination system. Rotations of the shank in the transverse plane around the Z-axis (ShankZ) are interpreted as internal rotation/external rotation. The calcaneus rotations in the sagittal plane around the X-axis (CalcaneusX) are interpreted as extension/flexion, frontal plane around the Y-axis (CalcaneusY) as inversion/eversion, and transverse plane around the Z-axis

(CalcaneusZ) as adduction/abduction. Rotational velocity relationship plots were generated for all trials to calculate coupling angles for the following paired rotations:

1) Shank Z-axis/calcaneus X-axis (ShankZ/CalcaneusX)
2) Shank Z-axis/calcaneus Y-axis (ShankZ/CalcaneusY)
3) Shank Z-axis/calcaneus Z-axis (ShankZ/CalcaneusZ)

The coupling angle (γ) was calculated for each instant (i) during the stance phase of each trial (j) (Figure 6.3-3). The coupling angle was the vector linking two successive time points relative to the right horizontal between the proximal ($\omega_{P(i)}, \omega_{P(i+1)}$) and distal ($\omega_{D(i)}, \omega_{D(i+1)}$) segmental velocities (Equation. 6.3-1 & 2) (Chang et al., 2008; Needham et al., 2014).

$$\gamma_{ij} = Atan\left(\frac{\omega_{D(j,i+1)} - \omega_{Dj.i}}{\omega_{P(j,i+1)} - \omega_{Pj.i}}\right).\frac{180}{\pi} \qquad \boldsymbol{\omega_{P(j,i+1)} - \omega_{Pj.i} > 0}$$

EQUATION 6. 3-1: Coupling angle when segment velocity at a certain point in time is larger than previous instance.

$$\gamma_{ij} = Atan\left(\frac{\omega_{D(j,i+1)} - \omega_{Dj.i}}{\omega_{P(j,i+1)} - \omega_{Pj.i}}\right).\frac{180}{\pi} + 180 \qquad \boldsymbol{\omega_{P(j,i+1)} - \omega_{Pj.i} < 0}$$

EQUATION 6. 3-2: Coupling angle when segment velocity at a certain point in time is smaller than previous instance.

The coupling angles were corrected to represent a value between 0 and 360° (Eq. 6.3-3) (Chang et al., 2008; Needham et al., 2014).

$$\gamma_{j,i} = \begin{cases} \gamma_{i,j} + 360 & \gamma_{i,j} < 0 \\ \gamma_{i,j} & \gamma_{i,j} \geq 0 \end{cases}$$

EQUATION 6. 3-3: Corrected coupling angle.

Mean $\overline{\gamma_i}$ were determined by calculating the average horizontal and vertical components at each instant for each trial, and then for all athletes using circular statistics (Batschelet, 1981).

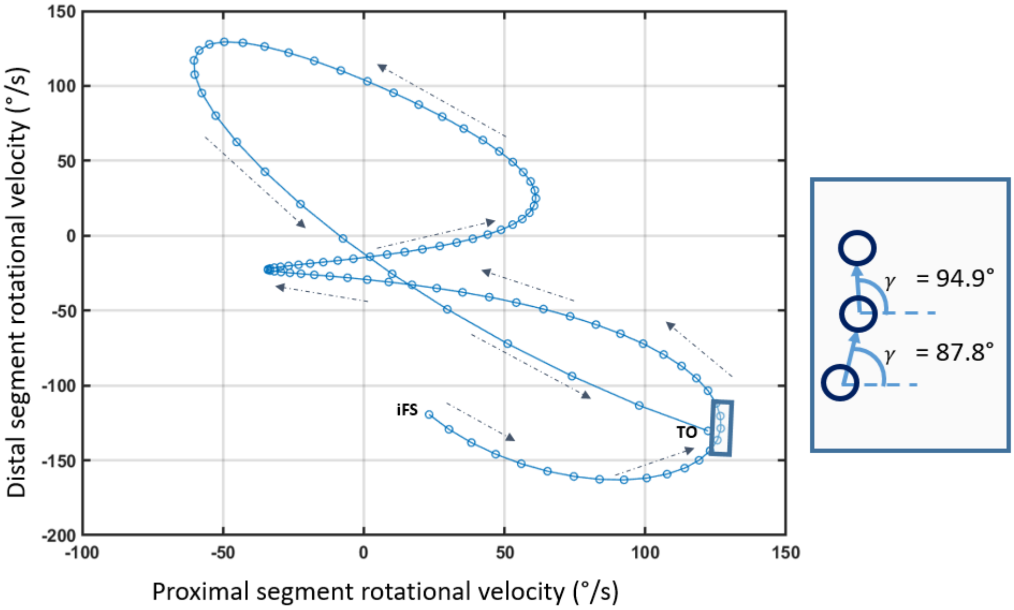
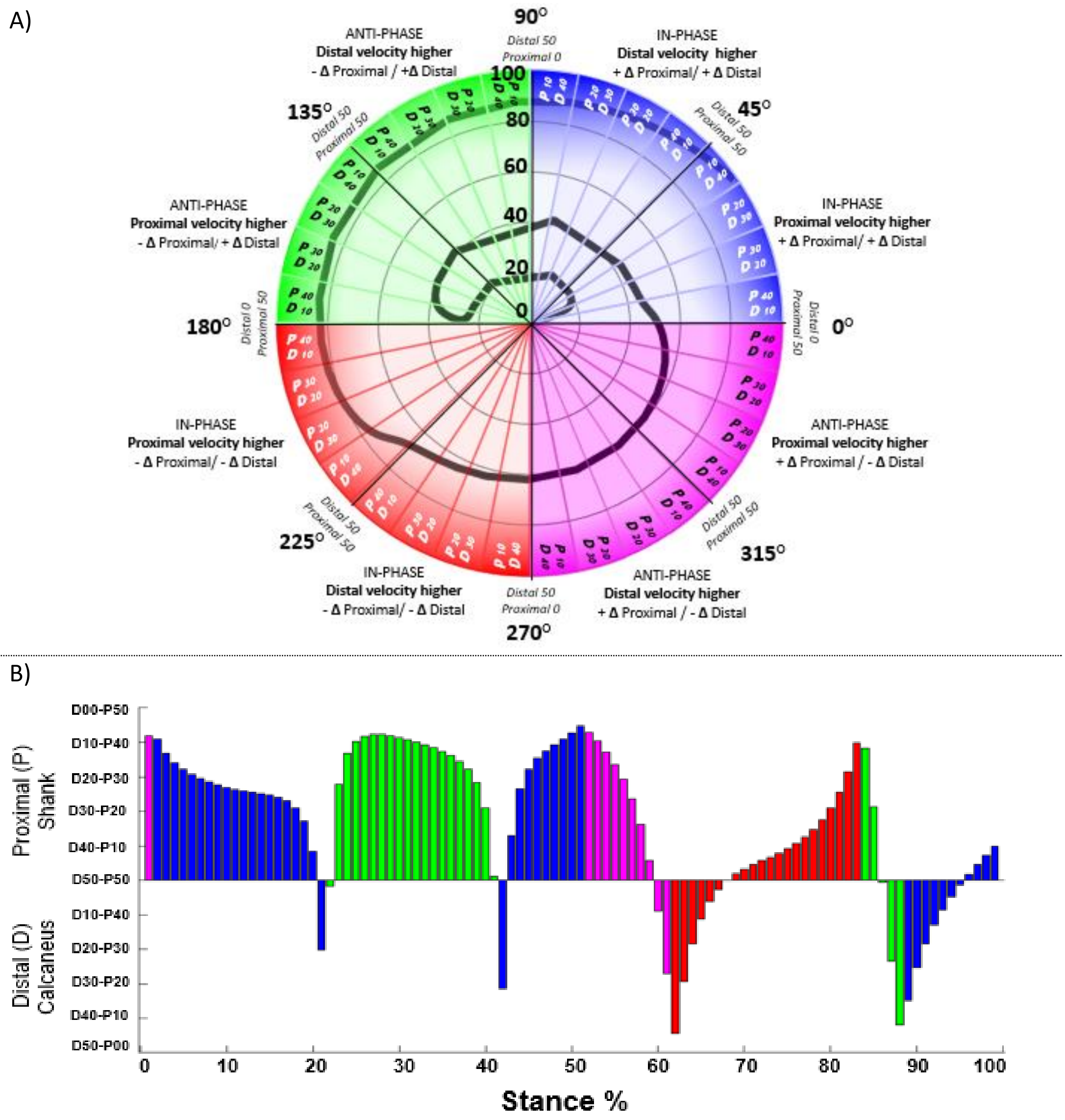


FIGURE 6. 3-3: An example of a relationship plot for the shank/calcaneus velocities. The arrows indicate the direction of movement. The inset explains the coupling angle between two successive points.

The coupling angles were described using modifications to the coordination pattern classifications as previously defined (Needham et al., 2015). A change in velocity (Δvelocity) was either positive or negative. Obeying the right-hand rule, a positive change in velocity for a segment indicated the change of directional rotation around the axis towards the positive direction (anti-clockwise). For example, a positive change in velocity for ShankZ indicated either an acceleration in a positive direction around the axis (increasing internal rotation velocity) or deceleration of the negative rotational direction around the axis (decreasing external rotation velocity). The opposite would achieve a negative change in velocity. In-phase coordination patterns were present when both segments had a concurrent positive or concurrent negative change in velocities, and anti-phase coordination when the two segments showed an opposite change in velocities (Figure 6.3-4 and Table 6.3-1).

The magnitude of the segmental rotation and contribution of each segment to the coordination pattern was calculated using the method described in previous research (Needham et al., 2015). Each quadrant (90°) was represented by 100 radians and can, therefore, could easily converted to a percentage. The contribution of a segment is quantified as a percentage of 50 (Figure 6.3-4 B). For example, equal contribution (moving at the same velocity) at a coupling angle of 45°, 135°, 225°, and 315° was expressed as Distal 50 – Proximal 50. The recommended method of displaying the coupling angle classifications (Needham, Naemi, Hamill, & Chockalingam, 2020) was modified to facilitate interpretation and the coordination pattern graphs display the proximal segment (shank) at the top and the distal segment (calcaneus) at the bottom (Figure 6.3-4 B).



Key:

IN-PHASE Shank +Δvelocity Calcaneus +Δvelocity	**ANTI-PHASE** Shank -Δvelocity Calcaneus +Δvelocity	**IN-PHASE** Shank -Δvelocity Calcaneus -Δvelocity	**ANTI-PHASE** Shank +Δvelocity Calcaneus -Δvelocity

FIGURE 6. 3-4: A) Polar plot of the coupling angle and classification of coordination patterns between the proximal (shank) and distal (calcaneus) segments (Adapted (Needham, Naemi, & Chockalingam, 2015)). The proximal-distal (P-D) numbering inside the wedges represents the contribution of each segment to the coordination pattern. The stance phase is represented by the radius in the polar plot. Initial foot strike is represented by the 0 R-tick (middle of the polar plot) and toe off by 100 R-tick. B) The coordination pattern displayed as a bar-graph with the proximal (shank) segment on the top half and the distal (calcaneus) segment on the bottom half.

TABLE 6. 3-1: Coordination pattern classifications. Adapted (Needham et al., 2015).

Coordination pattern	Coupling angle	Coupling direction
In-phase (*Proximal Δvelocity larger*)	0° to 45° (blue – top) 180° to 225° (red – top)	Both segments positive Δvelocity (0° to 45°), Both segments negative Δvelocity (180° to 225°)
In-phase (*Distal Δvelocity larger*)	45° to 90° (blue – bottom) 225° to 270° (red – bottom)	Both segments positive Δvelocity (45° to 90°), Both segments negative Δvelocity (225° to 270°)
Anti-phase (*Distal Δvelocity larger*)	90° to 135° (green – bottom) 270° to 315° (magenta – bottom)	Distal positive Δvelocity and proximal negative Δvelocity (90° to 135°) Distal negative Δvelocity and proximal positive Δvelocity (270° to 315°)
Anti-phase (*Proximal Δvelocity larger*)	135° to 180° (green – top) 315 to 360° (magenta – top)	Proximal positive Δvelocity and distal negative Δvelocity (90° to 135°) Proximal negative Δvelocity and distal positive Δvelocity (270° to 315°)
In-phase (*Equal Δvelocity*)	45° and 225°	Distal and proximal positive Δvelocity (45°) Distal and proximal negative Δvelocity (225°)
Anti-phase (*Equal Δvelocity*)	135° and 315°	Proximal negative Δvelocity and distal positive Δvelocity (135°) Proximal positive Δvelocity and distal negative Δvelocity (315°)
Proximal Δvelocity only	0°, 360°, and 180°	Proximal positive Δvelocity only (0°/360°) Proximal negative Δvelocity only (180°) (distal segment does not move)
Distal Δvelocity only	90 and 270°	Distal positive Δvelocity only (90°) Distal negative Δvelocity only (270°) (the proximal segment does not move)

Note: Positive Δvelocity for a segment indicates the change of directional rotation towards the positive direction (anti-clockwise) around the axis (e.g., accelerating shank internal rotation or decelerating shank external rotation). Negative Δvelocity indicates a change of directional rotation towards the negative direction (clockwise) around the axis (e.g., accelerating shank external rotation or decelerating shank internal rotation).

6.4. RESULTS

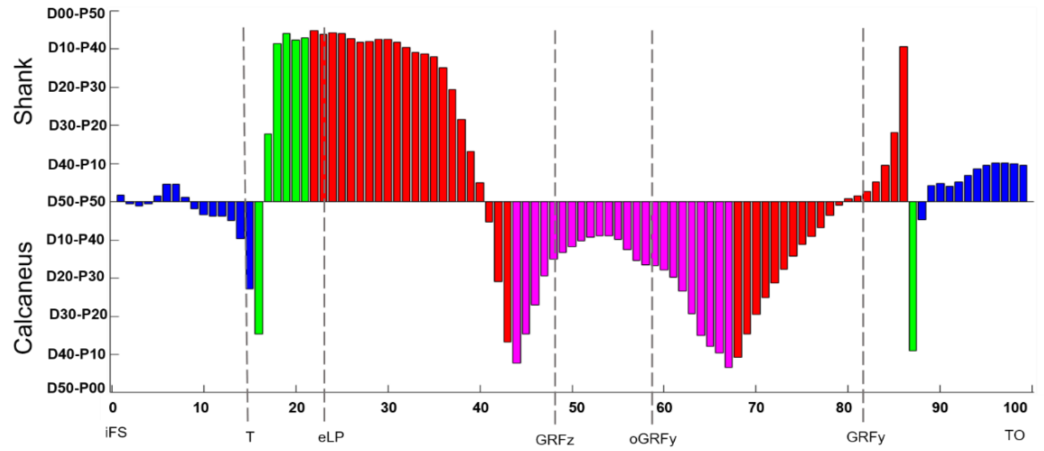
To facilitate interpretation of the coordination patterns, the average of segmental velocities was used to calculate the coupling angles (Equation 6.3-3). The coordination pattern of the individual athlete may vary from the average coordination pattern presented here. Segmental coupling relationships for all athletes can be found in the supplementary material (Appendix C).

6.4.1 AVERAGE COORDINATION PATTERNS

6.4.1.1 SHANKZ/CALCANEUSX

Mean coupling angles for ShankZ/CalcaneusX are shown in Figure 6.4-1. During the first 15 % of stance, the shank and calcaneus velocities were similar in direction and magnitude. The shank (proximal (P)) and calcaneus (distal (D)) segments contributed mostly in equal parts to the coordination pattern (D≈50:P≈50 to D≈20:P≈30). Within the loading phase, the shank first showed a change in velocity followed by the change in velocity of the calcaneus, suggesting a proximal-distal coupling effect. This apparent proximal-distal coupling effect was not observed for the remainder of the stance phase.

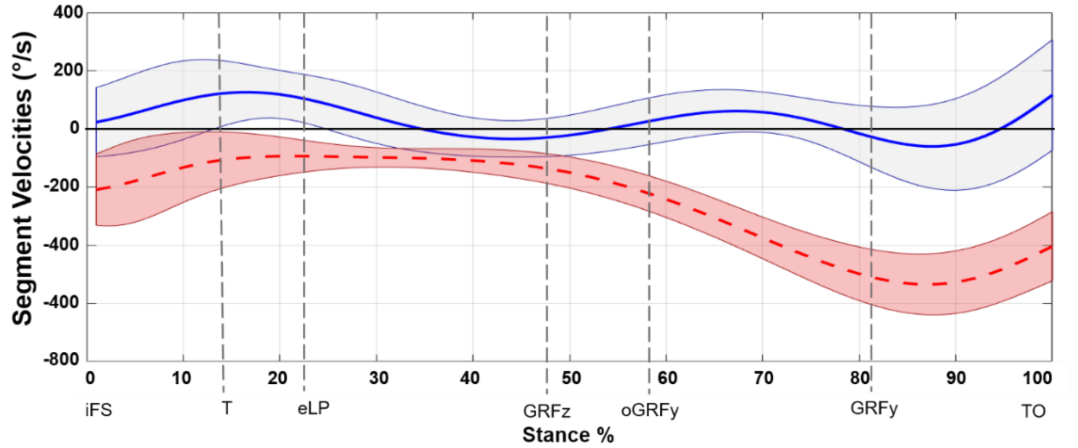
A)



Key:

IN-PHASE Shank +Δvelocity Calcaneus +Δvelocity	**ANTI-PHASE** Shank -Δvelocity Calcaneus +Δvelocity	**IN-PHASE** Shank -Δvelocity Calcaneus -Δvelocity	**ANTI-PHASE** Shank +Δvelocity Calcaneus -Δvelocity

B)



Key:

Shank(+IR/-ER)/ velocity

Calcaneus (+EXT/-FLX) velocity

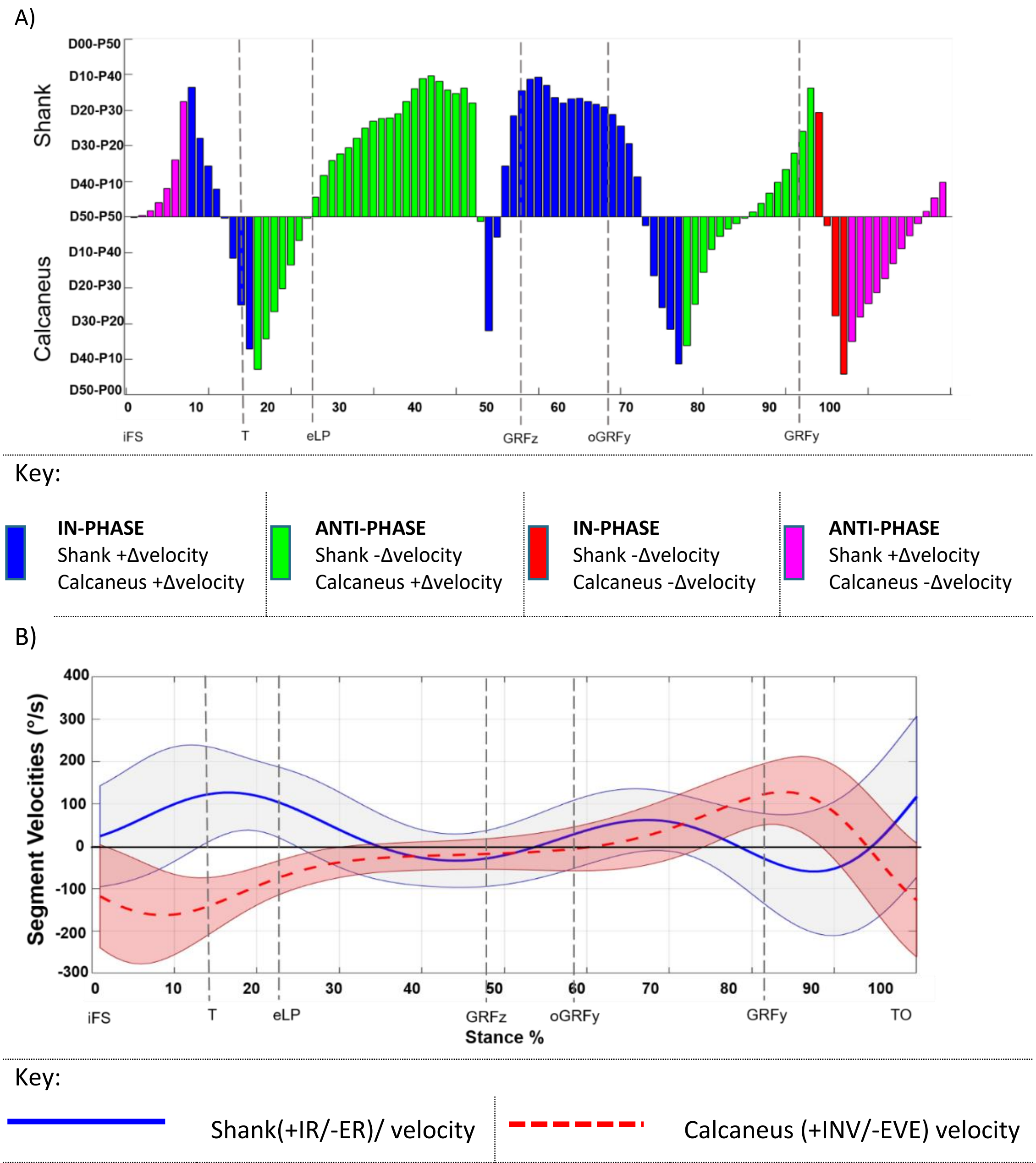
Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force, TO = toe off.

FIGURE 6. 4-1: A) Average ShankZ (+IR/-ER)/CalcaneusX (+EXT/-FLX) coordination pattern and B) Average ShankZ (+IR/-ER)/ CalcaneusX (+EXT/-FLX) segmental velocities.

6.4.1.2 ShankZ/CalcaneusY

From initial foot strike, an average anti-phase coordination pattern (shank internal rotation velocity increased – movement towards positive direction - while calcaneus eversion velocities increased – movement towards negative direction) was observed. Calcaneus eversion velocity decreased from ≈7 % of stance, while shank internal rotation velocity continued to increase, creating the average in-phase coordination pattern. The average shank internal rotation velocity only started to decrease at ≈15 % (Figure 6.4-2).

The duration of the in-phase accelerating shank internal rotation/decelerating calcaneus eversion coordination pattern varied between athletes (see supplementary material - (Appendix C)). A similar coupling sequence was observed towards the end of the stance phase, with the calcaneus change in segmental velocity preceding the shank change in segmental velocity. Thus, during the loading and propulsion phase, a positive change in velocity (calcaneus inversion) coincided with a negative change in velocity for the shank (external rotation), and a negative change in velocity (calcaneus eversion) coincided with a positive change in velocity for the shank (internal rotation).



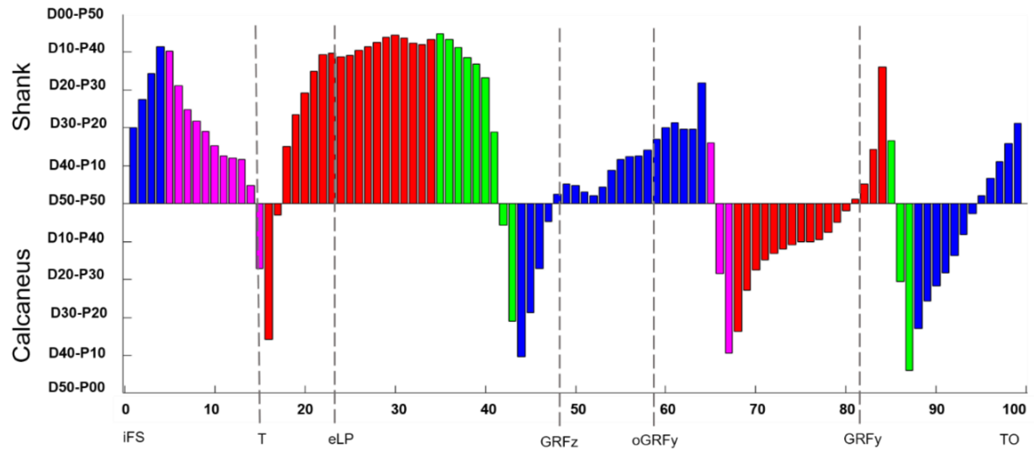
Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force, TO = toe off.

FIGURE 6. 4-2: A) Average ShankZ (+IR/-ER)/CalcaneusY (+INV/-EVE) coordination pattern and B) Average ShankZ (+IR/-ER)/ CalcaneusY (+INV/-EVE) segmental velocities.

6.4.1.3 ShankZ/CalcaneusZ

The average CalcaneusZ change in velocity occurred at ≈5 % of the stance (decrease of adduction velocity), while shank internal rotation velocity increased (Figure 6.4-3). The negative CalcaneusZ velocity continued to accelerate into abduction during the loading phase. Shank internal rotation velocity decreased at ≈15 % of the stance. Thus, by the end of the loading phase (at ≈22 %), most athletes displayed an in-phase negative change in velocity (shank external rotation/calcaneus abduction) coordination pattern. A similar pattern of shank rotational change in velocity following calcaneus change in velocity was observed for the remainder of the stance phase. A change in velocity in the calcaneus towards adduction preceded a change in velocity for the shank towards internal rotation, and abduction rotations at the calcaneus preceded external rotations at the shank.

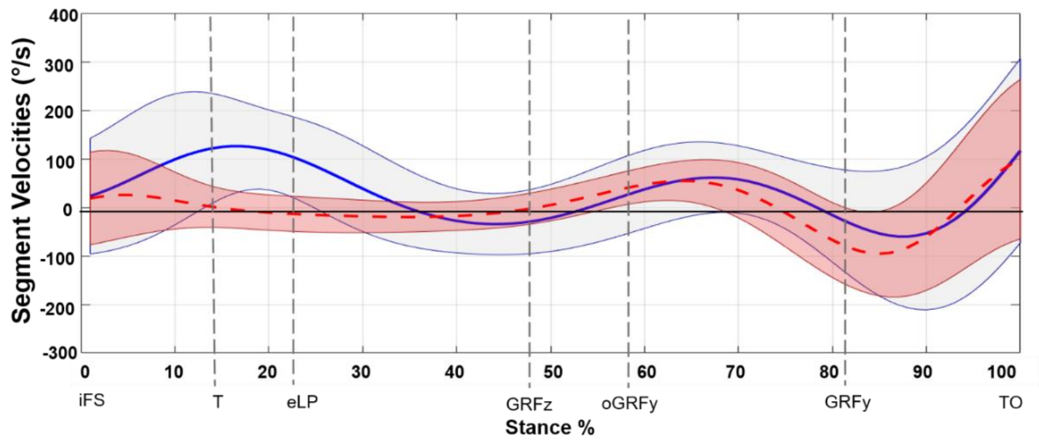
A)



Key:

IN-PHASE Shank +Δvelocity Calcaneus +Δvelocity	**ANTI-PHASE** Shank -Δvelocity Calcaneus +Δvelocity	**IN-PHASE** Shank -Δvelocity Calcaneus -Δvelocity	**ANTI-PHASE** Shank +Δvelocity Calcaneus -Δvelocity

B)



Key:

Shank(+IR/-ER)/ velocity | Calcaneus (+ADd/-ABd) velocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force, TO = toe off.

FIGURE 6. 4-3: A) Average ShankZ (+IR/-ER)/CalcaneusZ (+ADd/-ABd) coordination pattern and B) Average ShankZ (+IR/-ER)/CalcaneusZ (+ADd/-ABd) segmental velocities

6.5. DISCUSSION

The objective of this study was to test the hypothesis of a distal-proximal coupling relationship between the calcaneus and transverse axis rotation of the shank segment during unexpected change of direction tasks associated with ACL injury risk.

We observed distal-proximal coupling relationships between frontal-calcaneus and transverse plane shank as well as transverse- calcaneus and transverse plane shank rotations. The calcaneus eversion at iFS and ensuing coupling effect between calcaneus inversion and shank external rotation noted here were similar to previous straight-line locomotion studies (Chan & Rudins, 1994; Eslami, Begon, Farahpour, & Allard, 2007; Nester, Jarvis, Jones, Bowden, & Liu, 2014; Pohl et al., 2007). Calcaneus adduction has also previously been correlated with shank internal rotation during straight-line running (Fischer, Willwacher, Hamill, & Brüggeman, 2017). In addition, we observed a lag between transverse plane shank and transverse plane calcaneus rotations (Figure 6.4-3). The lag we observed here between calcaneus adduction/abduction, talus adduction/abduction, and shank internal/external rotation was also described in earlier work (Huson, 1991).

A distal-proximal coupling relationship between the calcaneus and shank segment may suggest the possibility of calcaneus manipulation to modulate the transfer of rotations to the knee joint via the shank, aiding in mitigating ACL injury risk. The link between foot pronation, shank internal rotation, and increased ACL injury risk has been made before (Beckett, Massie, Bowers, & Stoll, 1992; McLean et al., 2004). However, these studies were retrospective and as such the causal relationship between foot pronation and ACL injury risk is at best speculative.

Furthermore, in these studies, it was also unclear whether the movement at the hip and knee resulted in foot pronation as a compensatory function for internal tibial rotation (proximal-distal coupling) or whether foot pronation and calcaneus eversion contributed to internal tibial rotation (distal-proximal coupling). The current study suggests that tibial rotation (at least a component thereof) is a function of frontal and/or transverse plane calcaneal rotations.

Controlling frontal plane movement of the calcaneus is mostly achieved through orthotics and shoe design (Nunan & Walls, 2017). However, these external modes of calcaneus control are often only effective during a heel strike action, which is rare in high intensity change of direction tasks (Losito, 2017). The transverse plane coupling between the calcaneus and shank is reportedly dictated by anatomical properties like the elasticity of the ligament fibres and may therefore play a role in the coupling relationship of the ankle joint (Huson, 1991). However, passive structures are not necessarily the only components responsible for ShankZ/CalcaneusZ coupling. Neuromuscular control of the calcaneus may be possible as experimental studies have shown that the frontal plane movement at the subtalar joint (which includes extension-eversion-abduction movement of the calcaneus) can be concentrically controlled by the fibularis (longus and brevis) and eccentrically by the tibialis posterior (Klein, Mattys, & Rooze, 1996). In addition, stimulation of intrinsic muscles (i.e., abductor hallucis, flexor digitorum brevis, and quadratus plantae) reportedly caused calcaneus inversion under load (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014). Thus, strengthening these extrinsic and intrinsic foot muscles may be important in decreasing the transfer of rotational knee motion during dynamic tasks, potentially aiding the decrease of ACL injury risk. To this point, a study that tested the effect of combining the toe spread exercise, which stimulates the abductor hallucis muscle, and hip external rotation exercises, on navicular drop found a larger decrease in navicular drop for the group that did both foot and hip exercises compared to the group that performed only hip exercises (Goo, Kim, & Kim, 2016). Thus, including exercises to strengthen the extrinsic (fibularis and tibialis posterior) and intrinsic (abductor hallucis, flexor digitorum brevis, quadratus plantae) muscles that control frontal and transverse plane calcaneus rotation may expand comprehensive ACL injury prevention strategies.

The current study has a few limitations worth noting. Only female court sport athletes were recruited as this cohort is at a higher risk for sustaining ACL injury (Gamada, 2014; Shultz et al., 2015). Furthermore, the tasks were performed barefoot to simplify the placement and observation of reflective markers on the foot. Thus, the generalization of findings beyond this cohort and footwear condition is uncertain. The study was limited to investigating calcaneus and shank coupling even though other intersegmental couplings may influence calcaneus and/or shank coordination patterns (Dubbeldam, Nester, Nene, Hermens, & Buurke, 2013). We also acknowledge that fatigue may influence the coupling relationship between segments, which can influence lower limb biomechanics increasing ACL injury risk (Brüggeman, 1996). Furthermore, anatomical variations between athletes may result in different coupling strategies. However, the aim of this study was to focus on the movement relationship between the calcaneus and shank in healthy athletes.

To establish prospective coordination patterns, it was necessary to use the average motion over all trials. Individual differences in calcaneus and shank movement were seen (note the variability in the segment velocities from iFS to T shown in Figures 6.4-2 and 6.4-3), which may influence ACL injury risk for the athletes. The differences, however, were primarily seen in the timing of the change in coordination pattern, not in the sequence of the coupling angle changes (see supplementary material - Appendix C). As such, the current study demonstrated that changes in calcaneal velocity can influence the internal/external velocity changes of the shank. A future randomised controlled intervention study needs to be undertaken to test this hypothesis. Finally, we acknowledge that the vector coding technique is sensitive to small changes in segment movements (Chang et al., 2008). To minimize errors and ensure the reliability of segmental movements, a reliable foot model was used, and the same trained investigator applied markers on all individuals (Caravaggi et al., 2011; Leardini et al., 2007).

6.6. CONCLUSION

This project established distal-proximal coupling in the ShankZ/CalcaneusY and ShankZ/CalcaneusZ coordination patterns. The change in calcaneal velocity from eversion towards inversion and calcaneal adduction to abduction may have slowed the internal rotational velocity of the shank within the loading phase. As ACL injury is likely to occur during the loading phase (Brown, Brughelli, & Hume, 2014; Koga et al., 2010), controlling frontal and transverse plane calcaneus velocities may influence the internal shank rotation component linked to ACL injury risk. The phasic lag between motions of the calcaneus and shank is possibly related to the elastic properties of the horizontal fibres of the ankle ligaments (Huson, 1991). However, previous studies have shown that neuromuscular training can alter calcaneus and shank kinematics and may play a role in segmental coordination patterns in addition to passive structures (Fraser, Feger, & Hertel, 2016; Kelly et al., 2014). From this study it seems plausible that manipulating the calcaneus may modulate the transfer of distal-proximal rotation via the shank to the knee.

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CHAPTER 7

DESCRIBING THE COUPLING RELATIONSHIP BETWEEN FOOT SEGMENTS DURING UNANTICIPATED CHANGES OF DIRECTION: THEORETICAL IMPLICATIONS FOR PROPHYLACTIC LATERAL ANKLE SPRAIN STRATEGIES

PUBLICATION

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7.1. ABSTRACT

BACKGROUND AND OBJECTIVES

Stability to the ankle joint is passively provided by the joint congruity and ligaments, and actively by the muscles acting on the foot. The ankle joint is most stable when loaded and dorsiflexed. However, during unanticipated changes of direction, typical in court sports, the foot is often in a vulnerable unloaded, plantarflexed position, increasing lateral ankle sprain (LAS) risk. An athlete's risk for re-injury and developing long term dysfunction increases significantly after sustaining an acute LAS. Research has shown that stability of the forefoot and controlling rearfoot movement to avoid excessive ankle inversion and adduction play a role in LAS injury risk reduction. However, information regarding the coupling relationship between the forefoot (hallux and metatarsal segments) and the rearfoot (calcaneus segment) during unanticipated changes of direction is lacking. We aim to test the hypothesis that a distal-proximal coupling relationship exists between the forefoot and rearfoot during unanticipated change of direction.

METHOD

The coupling angles between sagittal plane hallux, tri-planar metatarsal, and frontal- and transverse plane calcaneus movement, respectively, were calculated with a modified vector coding technique which used segmental velocities in a local, anatomical reference frame instead of segmental angles in a global reference frame.

RESULTS

Coupling relationships revealed anti-phase movement (uncoupled movement) between sagittal- metatarsal- and frontal plane calcaneus movement throughout stance. During loading, sagittal- and frontal plane metatarsal acceleration/deceleration coincided with frontal-transverse plane calcaneus acceleration/deceleration, respectively. The remainder of the braking phase was characterised by calcaneus eversion deceleration. During propulsion, the hallux and metatarsal segments increased flexion velocity in response to calcaneus inversion and adduction acceleration.

CONCLUSIONS

As the forefoot was the only point of contact during stance, the coordinated movement between segments were most likely due to neuromuscular control, rather than anatomical congruity. Strengthening intrinsic and extrinsic foot muscles may thus contribute to foot and ankle stiffness and stability, complimenting current prophylactic LAS strategies.

KEY WORDS

lateral ankle sprain, injury prevention, multi-segment foot coordination, vector coding technique, unanticipated cutting

7.2. INTRODUCTION

Lateral ankle sprain (LAS) injuries are common in females competing in sports with high-intensity, abrupt, unanticipated changes of direction (Doherty et al., 2014; Fong, Hong, Chan, Yung, & Chan, 2007). Preventing first-time LAS injuries should be a high priority, as an athlete's risk for re-injury and developing chronic ankle instability increases significantly after sustaining an acute LAS (Gribble et al., 2016). Injury to any of the three lateral ankle ligaments (anterior talofibular, calcaneofibular, and posterior talofibular ligaments) arises from rapidly developing, unrestrained inversion and/or supination velocities and moments around the foot-shank complex (Ferran & Maffulli, 2006; Fong, Chan, Mok, Yung, & Chan, 2009b; Fong, Ha, Mok, Chan, & Chan, 2012; Gehring, Wissler, Mornieux, & Gollhofer, 2013a; Gribble et al., 2016; Kristianslund, Bahr, & Krosshaug, 2011; Panagiotakis, Mok, Fong, & Bull, 2017).

The foot and ankle complex is most stable in dorsiflexion as the articulation surfaces of the various segments are more congruent, limiting excessive inversion and adduction of both the talocrural and subtalar joints (Duerinck, Hagman, Jonkers, Van Roy, & Vaes, 2014; Stormont, Morrey, An, & Cass, 1985). The lateral ankle ligaments provide further passive stability to the ankle, whereas the intrinsic and extrinsic muscles acting on the foot offer dynamic stability (Hertel, 2002; Stephens & Sammarco, 1992). As the calcaneus is typically off the floor during cutting and shuffling movements, the lateral stability provided by subtalar and talocrural joints in dorsiflexion, is unattainable. Stability of the foot, therefore, needs to be established in the forefoot while the rearfoot movement is controlled to limit rotational velocities. Understanding how the hallux and metatarsal segments respond to or influence calcaneus inversion and adduction velocities, while maintaining functional performance during a change of direction task, should provide valuable information to develop prophylactic LAS strategies. Previous research has revealed that complex rotational couplings exist between the hallux, metatarsals, midfoot, and calcaneus segments (Dubbeldam, Nester, Nene, Hermens, & Buurke, 2013; Nester, Jarvis, Jones, Bowden, & Liu, 2014; Pohl, Messenger, & Buckley, 2007). However, these studies were mainly on straight line walking or running tasks where a heel foot strike pattern was common. Pohl and Buckley (2008) found that the foot strike angle has an influence on foot intersegmental coupling patterns. Furthermore, changing the direction of locomotion also changes the function and control of certain muscles acting on the foot (Zelik, La Scaleia, Ivanenko, & Lacquaniti, 2015) and conceivably also segmental movements. The previous findings of intersegmental coordination patterns may therefore not be appropriate in discerning prophylactic LAS strategies.

Analysing the interaction and coordination patterns between foot segments by means of segmental coupling provides unique insights into segmental rotations and their interrelationship. More conventional kinematic techniques only provide information about a single parameter (e.g., angle, velocity, or displacement) of a given joint or segment at a certain point in time. Other functional data analysis methods may analyse the function of a joint across the whole range of motion considering the time-dependency of biomechanical data (Hébert-Losier et al., 2015). However, these methods do not discern between the unique contribution and interaction of each segment to the observed joint movement. Non-linear, continuous relative phase or vector coding techniques are often used to study the dynamic coordination between segments over the entire stance phase (Glazier, 2006; Hamill, Haddad, & McDermott, 2000; Hamill, van Emmerik, Heiderscheit, & Li, 1999). Recently, a modified vector coding technique has been proposed that highlights the contribution of each individual segment to the coordination pattern (Chang, Van Emmerik, & Hamill, 2008; Needham, Naemi, & Chockalingam, 2015). The aim of this study is to use this modified vector coding technique to test the hypothesis of a distal-proximal coupling relationship between the calcaneus/hallux and calcaneus/metatarsal segments by describing and quantify the coupling relationship during an unanticipated change of direction task.

7.3. METHODS

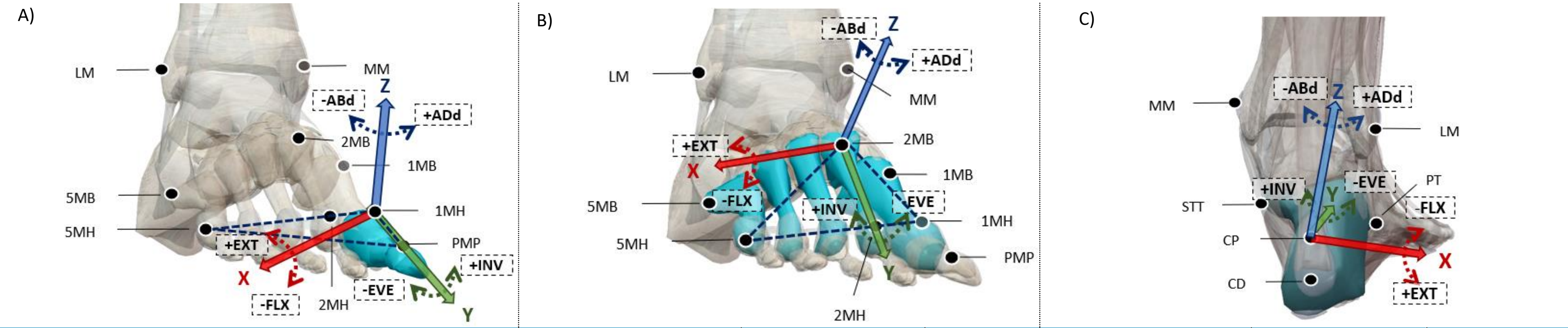
7.3.1 PARTICIPANTS

Eighteen skilled court sport (netball, volleyball, tennis, badminton, and squash) female athletes (mean ± standard deviation; age: 17.4 ± 1.7 years; height: 1.68 ± 0.06 m; mass: 65.53 ± 8.96 kg) took part in this study. All athletes had been injury-free for the six months preceding the experiment, were pain-free, and had no diagnosed musculoskeletal or neurological condition affecting movement or foot function. All athletes gave written informed consent with ethical approval received from the University Human Ethics Committee (SOA 16/77).

7.3.2 DATA COLLECTION

7.3.2.1 MARKER PLACEMENT

The motion of the hallux, metatarsal, and calcaneus segments was tracked by spherical retro-reflective markers. The 8 mm markers were placed on landmarks according to a previously published multi-segmental foot model (Leardini et al., 2007a). The landmarks used to define the hallux, metatarsal, and calcaneus segments and their local reference frames are presented in Figure 7.3-1. A ten camera (six Oqus 3-series and four Oqus 7-series) Qualisys system (QTM software v2.17; Qualisys, Gothenburg, Sweden) sampling at 200 Hz was used to collect kinematic data. A Kistler, 569x DAQ force platform (Winterhur, Switzerland) was used to collect kinetic data at a sampling rate of 200 Hz.



Landmarks	Local reference frame	Rotation
PMP: distal, dorsal point of the head of the proximal phalanx of the hallux	**Z-axis**: vector normal to the plane defined by the PMP, 1MH, and 5MH, and originating at 1MH	+adduction (+ADd) -abduction (-ABd)
1MH: dorso-medial head of the first metatarsal 5MH: dorso-lateral head of the fifth metatarsal	**Y-axis:** line joining 1MH and PMP, normal to the Z-axis and parallel to the plane defined by PMP, 1MH, and 5MH	+inversion (+INV) -eversion (-EVE)
	X-axis: orthogonal to the Z and Y axes	+extension (+EXT) -flexion (-FLX)

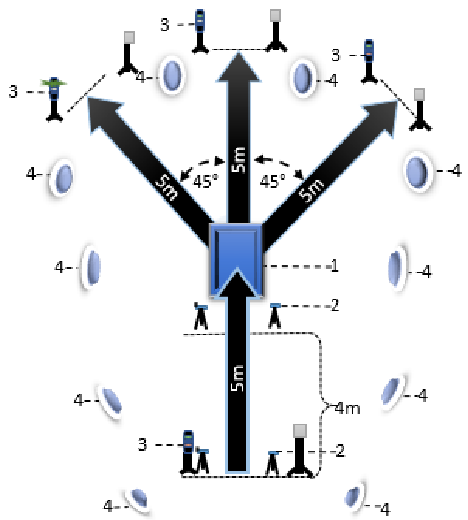
Landmarks	Local reference frame	Rotation
1MB: dorso-medial base of the first metatarsal 1MH: dorso-medial head of the first metatarsal 2MB: dorsal base of the second metatarsal 2MH: dorsal head of the second metatarsal 5MB: dorso-lateral base of the fifth metatarsal 5MH: dorso-lateral head of the fifth metatarsal	**Y-axis**: line joining 2MB and 2MH projected onto the plane defined by 2MB, 1MH, and 5MH, and originating at 2MB	+inversion (+INV) -eversion (-EVE)
	X-axis: normal to the Y-axis and parallel to the plane defined by 2MB, 1MH, and 5MH, originating at 2MB	+extension (+EXT) -flexion (-FLX)
	Z-axis: orthogonal to the X and Y axes	+adduction (+ADd) -abduction (-ABd)

Landmarks	Local reference frame	Rotation
STT: **s**ustentaculum tali PT: peroneal tubercle CD: distal calcaneus CP: proximal calcaneus	**Y-axis:** line between CP and the midpoint between STT and PT, and originating at CP	+inversion (+INV) -eversion (-EVE)
	X-axis: line normal to the Y- axis and parallel to the plane defined by CP, PT, and STT, originating at CP	+extension (+EXT) -flexion (-FLX)
	Z-vertical axis: orthogonal to the X and Y axes, originating at CP	+adduction (+ADd) -abduction (-ABd)

FIGURE 7. 3-1: Bony landmarks used to define the right A) hallux, B) metatarsal, and C) calcaneus segments and the orientation of the respective local reference frames (Leardini et al., 2007; Portinaro et al., 2014). Images courtesy of Visible Body (Argosy Publishing, 2019).

7.3.2.2 EXPERIMENTAL PROTOCOL

Athletes were instructed to complete 45° unanticipated cutting tasks on the force platform running barefoot. For the purpose of this study, trials were analysed when force platform foot strike was made with the dominant leg, as that limb is thought to be more susceptible to injury (Gamada, 2014). Dominance was defined as the leg chosen to step up onto a 45 cm high step-up box (Hoffman, Schrader, Applegate, & Koceja, 1998). The approach speed (4.5 ± 0. 47 m·s^{-1}) of the 5 m run-up to the force platform was monitored by a photocell timing system (Brower Timing System, USA) to ensure comparison of the data between athletes (Spiteri, Hart, & Nimphius, 2014). One set of traffic lights (Smart Speed, Fusion Sport, Queensland, Australia) flashed 0.7 sec after the 5 m approach run was initiated. Athletes were required to perform five successful trials for each randomly indicated direction to maintain the unanticipated demand of the task. A trial was deemed successful if: 1) the approach run was within 10 % of the desired 4.5 m·s^{-1} approach speed, 2) the force platform was struck with the dominant foot, and 3) the cut was performed towards the indicated direction.



1. Force platform – identify initial foot strike and toe-off events
2. Brower timing gates – monitor approach speed
3. SmartSpeed timing gates – random flashing light to indicate directional change
4. 10 Qualisys cameras – motion capture

FIGURE 7. 3-2: Experiment set up.

7.3.3 DATA PROCESSING

The analogue GRF signals were filtered using a fourth-order low-pass Butterworth filter with a cut-off frequency of 25 Hz. The start of the stance phase was recorded from when the GRFz exceeded 20 N (initial foot strike, iFS) and ended when the GRFz fell below 20 N (toe off, TO) (Pfile et al., 2013). The GRFz impact transient was identified as the rapidly developing spike in the GRFz curve, occurring within the first 50 ms of stance (Lieberman et al., 2010). The loading phase was defined as the period from iFS to the average occurrence of the first trough in the GRFz data (0-22 % of stance). From the GRFy curve, the anterior-posterior braking phase (negative GRFy), propulsion phase (positive GRFy), and peak propulsion force were identified.

The trajectories from the markers were interpolated in QTM, allowing a maximum of 10 frames (50 ms) for gap filling. The signals were exported to Visual 3D™ (v5.0, C-Motion, USA) and smoothed using a fourth-order low-pass Butterworth filter with a 10 Hz cut-off frequency, time scaled, and normalized to 100 % of the stance phase. The foot strike angle was calculated as the angle between the foot segment and the virtually created lab floor. The foot-to-floor angle at iFS of each trial was subtracted from the foot-to-floor angle obtained from the calibration trial of each athlete, where a negative and positive angle represents a forefoot and rearfoot strike, respectively. The segment velocities, relative to the lab and expressed in the segment's coordinate system were also determined using Visual 3D™ . The coupling angles were calculated from the segment velocity signals after they were exported to MATLAB (R2018a, Mathworks, MA USA). The GRF events throughout the stance phase was identified in MATLAB. Diagrams and plots were also created in MATLAB.

7.3.4 Calculating the coupling angle

The calcaneus-hallux and calcaneus-metatarsal coordination patterns were analysed by adapting the modified vector coding system described by Chang et al. (2008). Segmental rotational velocities in a local reference frame were used, as opposed to the more commonly used segment positional angles in the global reference frame. When the body moves in 3D space, the local reference frame of the individual segment is no longer aligned with the global reference frame. The analysis of relative motion between individual segments becomes problematic. In addition, the local reference system of individual segments cannot be used to calculate the angular position as the angular position in the local reference frame will always, by definition, be the same. Rotational velocities, which do not need to be measured relative to a secondary reference frame, were therefore used to calculate the coupling angle between segments. The segmental reference systems are displayed in Figure 7.3-1.

The X-Y-Z Cardan sequence, corresponding to the joint coordination system, interpreted segmental rotations. Rotations in the sagittal plane around the X-axis are interpreted as extension/flexion, in the frontal plane around the Y-axis as inversion/eversion, and in the transverse plane around the Z-axis as adduction/abduction. Rotational velocity relationship plots were created for all trials for all coupling relationships between the calcaneus/hallux and calcaneus/metatarsal of interest (Table 7.3-1).

TABLE 7. 3-1: The coupling relationships between the calcaneus/hallux and calcaneus/metatarsal of interest.

Coordination with calcaneus inversion/eversion (Y-axis)	**Coordination with calcaneus adduction/abduction (Z-axis)**
Calcaneus Y-axis/Hallux X-axis (CalcaneusY/HalluxX)	Calcaneus Z-axis/Hallux X-axis (CalcaneusZ/HalluxX)
Calcaneus Y-axis/Metatarsal X-axis (CalcaneusY/Metatarsal X)	Calcaneus Z-axis/Metatarsal X-axis (CalcaneusZ/Metatarsal X)
Calcaneus Y-axis/Metatarsal Y-axis (CalcaneusY/Metatarsal Y)	Calcaneus Z-axis/Metatarsal Y-axis (CalcaneusZ/Metatarsal Y)
Calcaneus Y-axis/Metatarsal Z-axis (CalcaneusY/Metatarsal Z)	Calcaneus Z-axis/Metatarsal Z-axis (CalcaneusZ/Metatarsal Z)

The coupling angle (γ) was inferred for each instant (i) during the stance phase of each trial(j) (Figure 7.3-3). The coupling angle was the vector linking two successive time points relative to the right horizontal between the proximal $(\omega_{P(i)}, \omega_{P(i+1)})$ and distal $(\omega_{D(i)}, \omega_{D(i+1)})$ segmental velocities (Equation. 7.3-1 & 2) (Chang et al., 2008; Needham, Naemi, & Chockalingam, 2014).

$$\gamma_{ij} = Atan\left(\frac{\omega_{D(j,i+1)} - \omega_{Dj.i}}{\omega_{P(j,i+1)} - \omega_{Pj.i}}\right) \cdot \frac{180}{\pi} \qquad \boldsymbol{\omega_{P(j,i+1)} - \omega_{Pj.i} > 0}$$

EQUATION 7. 3-1: Coupling angle when segment velocity at a certain point in time is larger than previous instance.

$$\gamma_{ij} = Atan\left(\frac{\omega_{D(j,i+1)} - \omega_{Dj.i}}{\omega_{P(j,i+1)} - \omega_{Pj.i}}\right) \cdot \frac{180}{\pi} + 180 \qquad \boldsymbol{\omega_{P(j,i+1)} - \omega_{Pj.i} < 0}$$

EQUATION 7. 3-2: Coupling angle when segment velocity at a certain point in time is smaller than previous instance.

The coupling angle was corrected to represent a value between 0° and 360° (Equation 7.3-3) (Chang et al., 2008; Needham et al., 2014).

$$\gamma_{j,i} = \begin{cases} \gamma_{i,j} + 360 & \gamma_{i,j} < 0 \\ \gamma_{i,j} & \gamma_{i,j} \geq 0 \end{cases}$$

EQUATION 7. 3-3: Corrected coupling angle.

Mean $\overline{\gamma_i}$ was determined by calculating the average horizontal and vertical components at each instant for each trial and then for all athletes using circular statistics (Batschelet, 1981).

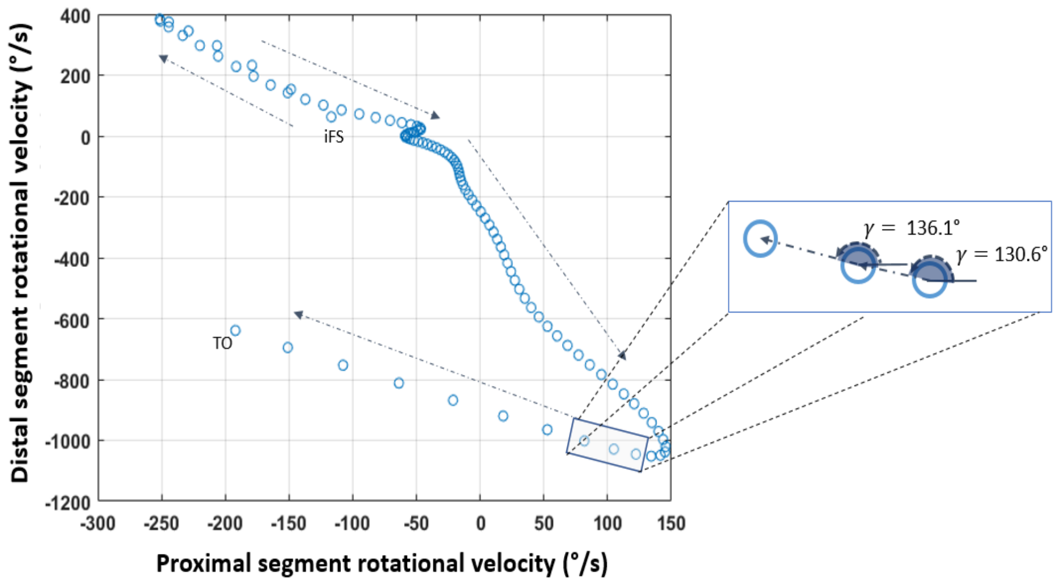


FIGURE 7. 3-3: An example of a relationship plot between the calcaneus and metatarsal rotational velocities. The arrows indicate the direction of the movement from initial foot strike (iFS) to toe-off (TO). The inset explains the coupling angle between two successive points in time.

The classification of the coupling angle in this study is similar to other studies where the coordination pattern was described in a continuous manner (Needham et al., 2015). However, the classification system was modified to describe segmental rotational velocities. Specifically, the change in the rotational velocity (Δvelocity) is either positive or negative. The right-hand rule is used to describe a positive change in velocity as a change in rotational direction towards the positive direction around the axis (anti-clockwise). For example, a positive change in velocity for CalcaneusY would be achieved by either an acceleration in a positive direction around the axis (e.g., accelerating calcaneus inversion velocity) or deceleration of the negative rotational direction around the axis (e.g., decelerating calcaneus eversion velocity). The opposite would reflect a negative change in velocity.

The coordination between segments was in-phase when both segments had simultaneous positive or negative change in velocities and anti-phase when the change in velocities had opposing signs (Table 7.3-2). The magnitude of each segment's rotational velocity and subsequent contribution to the coordination pattern was classified and depicted based on prior work (Needham et al., 2015; Needham, Naemi, Hamill, & Chockalingam, 2020), and is exemplified in Figure 7.3-4. Each quadrant (90°) represents 100 radians and can be easily converted to a percentage. The individual contribution of a segment is denoted by a percentage over 50. For example, when the distal segment is moving and the proximal segment has a 0°/s velocity (at coupling angle 90° or 270°), the contribution is expressed as Distal 50 – Proximal 0. The wedges in each quadrant have proximal-distal numbering representing the segmental contributions to the coordination pattern. The display of coordination patterns in the results section was modified so that the proximal segment is represented by the top part of the coordination pattern graph and the distal segment by the bottom.

TABLE 7. 3-2: Coordination pattern classifications. Adapted from (Needham et al., 2015).

Coordination pattern	Coupling angle	Coupling direction
In-phase (*Proximal Δvelocity larger*)	0° to 45° (blue – top) 180° to 225° (red – top)	Both segments positive Δvelocity (0° to 45°), Both segments negative Δvelocity (180° to 225°)
In-phase (*Distal Δvelocity larger*)	45° to 90° (blue – bottom) 225° to 270° (red – bottom)	Both segments positive Δvelocity (45° to 90°), Both segments negative Δvelocity (225° to 270°*)*
Anti-phase (*Distal Δvelocity larger*)	90° to 135° (green – bottom) 270° to 315° (magenta – bottom)	Distal positive Δvelocity and proximal negative Δvelocity (90° to 135°) Distal negative Δvelocity and proximal positive Δvelocity (270° to 315°)
Anti-phase (*Proximal Δvelocity larger*)	135° to 180° (green – top) 315 to 360° (magenta – top)	Proximal positive Δvelocity and distal negative Δvelocity (90° to 135°) Proximal negative Δvelocity and distal positive Δvelocity (270° to 315°)
In-phase (*Equal Δvelocity*)	45° and 225°	Distal and Proximal positive Δvelocity (45°) Distal and Proximal negative Δvelocity (225°)
Anti-phase (*Equal Δvelocity*)	135° and 315°	Proximal negative Δvelocity and distal positive Δvelocity (135°) Proximal positive Δvelocity and distal negative Δvelocity (315°)
Proximal Δvelocity only	0°, 360°, and 180°	Proximal positive Δvelocity only (0°/360°) Proximal negative Δvelocity only (180°) (distal segment does not move)
Distal Δvelocity only	90° and 270°	Distal positive Δvelocity only (90°) Distal negative Δvelocity only (270°) (proximal segment does not move)

Notes. Positive Δvelocity for a segment indicates the change of directional rotation towards the positive direction (anti-clockwise) around the axis (e.g., accelerating calcaneus inversion or decelerating calcaneus eversion). Negative Δvelocity indicates a change of directional rotation towards the negative direction (clockwise) around the axis (e.g., accelerating calcaneus eversion or decelerating calcaneus inversion). Segmental contribution to the coupling angle is presented in Figure 7.3-4.

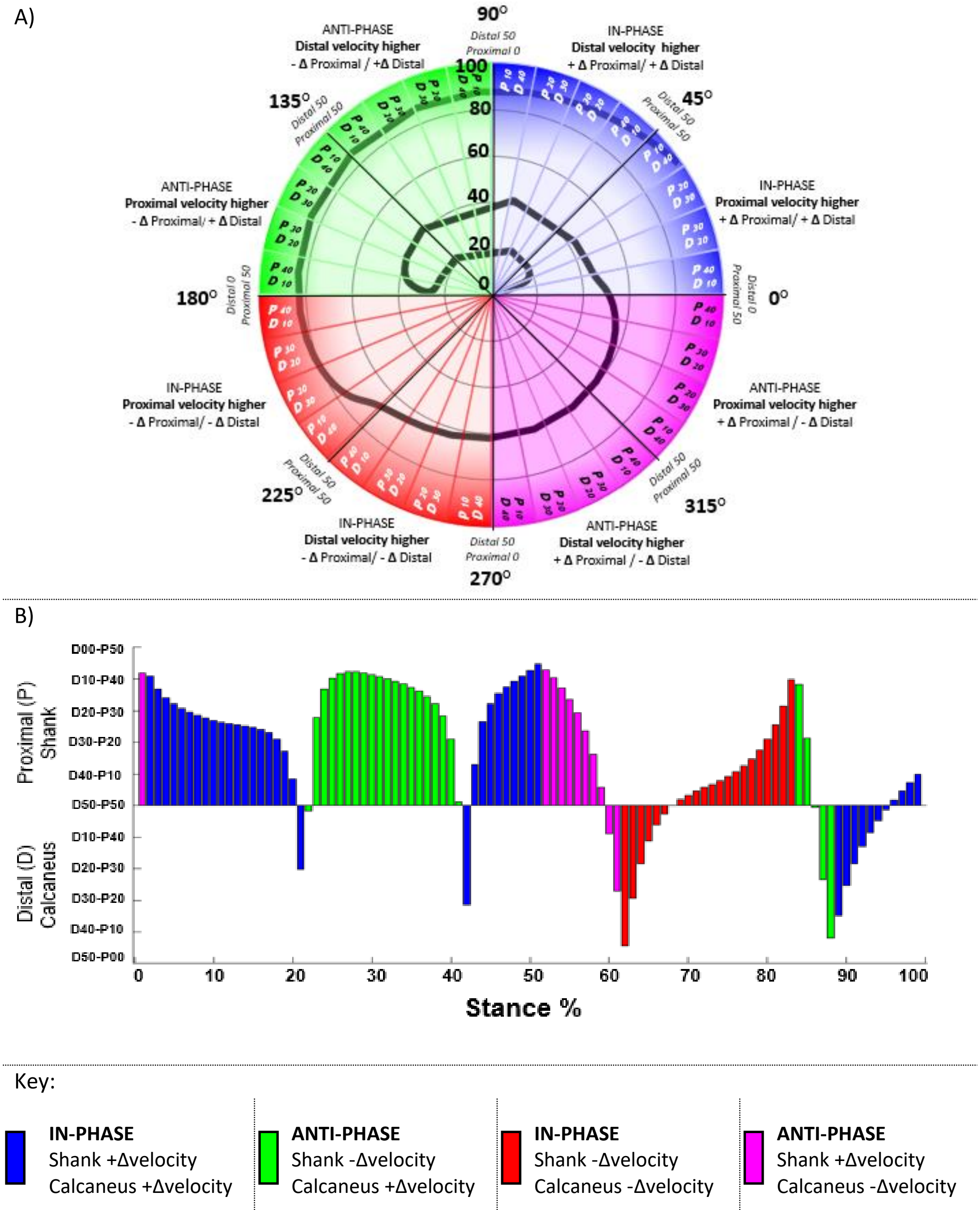


FIGURE 7. 3-4: A) Polar plot of the coupling angle and classification of coordination patterns between the proximal (calcaneus) and distal (metatarsals / hallux) segments (Adapted (Needham et al., 2015)). The proximal-distal (P-D) numbering inside the wedges represents the contribution of each segment to the coordination pattern. The stance phase is represented by the radius in the polar plot. Initial foot strike is represented by the 0 R-tick (middle of the polar plot) and toe off by 100 R-tick. B) The coordination pattern displayed as a bar-graph with the proximal (calcaneus) segment on the top half and the distal (metatarsals / hallux) segment on the bottom half.

7.4. RESULTS

7.4.1 AVERAGE COORDINATION PATTERNS INFLUENCING CALCANEUS INVERSION/EVERSION

In the majority of trials, a negative foot-to-floor angle (-3.53 ± 8.11◦) was observed, indicating a forefoot strike at iFS (Figure 7.4-1). In the frontal plane, the metatarsal segment changed from its initial MetatarsalY eversion acceleration to deceleration at ≈5 % of stance, preceding the CalcaneusY eversion acceleration to eversion deceleration at ≈9 % of stance. The MetatarsalX positive change in velocity (accelerating metatarsal extension) occurred at ≈11 % of stance following CalcaneusY negative change in velocity (accelerating calcaneus eversion) into MetatarsalX negative change in velocity (decelerating metatarsal extension).

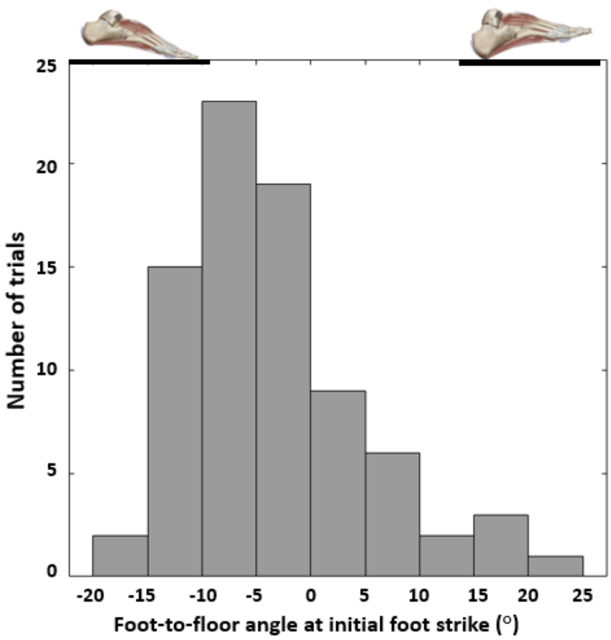
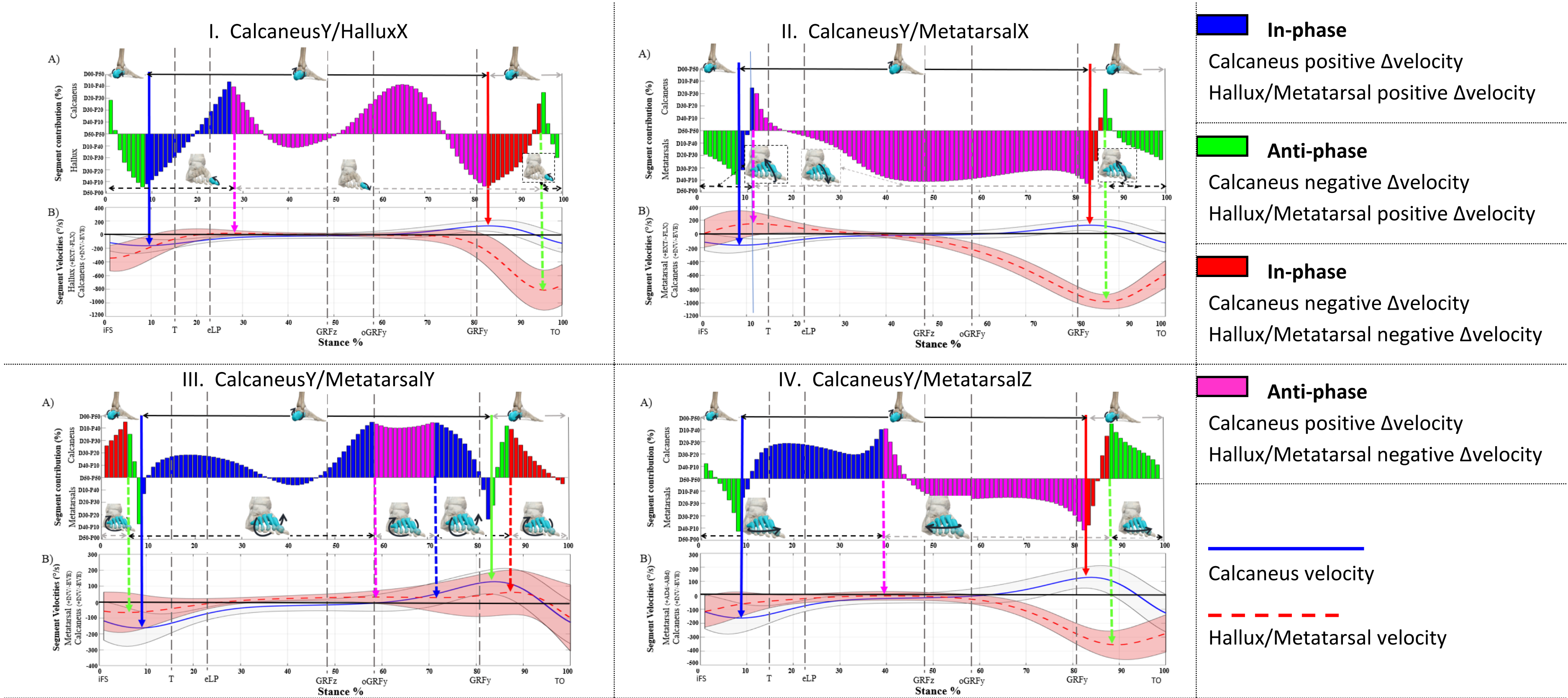


FIGURE 7. 4-1: Histogram displaying the initial foot strike angle. A negative angle indicates a forefoot strike. A positive angle indicates a rearfoot strike.

After peak GRFz was reached, calcaneus positive change in velocity (calcaneus inversion acceleration) contributed the most to the CalcaneusY/MetatarsalY coordination pattern. The average MetatarsalY segment rotational velocity displayed large variability between athletes during the propulsion phase. The CalcaneusY/MetatarsalY and CalcaneusY/MetatarsalX coordination patterns maintained their close coupling relationship between peak GRFy (≈81 % of stance) and TO, as observed during the loading phase from iFS to the average occurrence of the first trough (≈22 % of stance) in the GRFz data. No coupling effect was discernible in CalcaneusY/HalluxX and CalcaneusY/MetatarsalZ coordination patterns (Figure 7.4-2. III and IV).



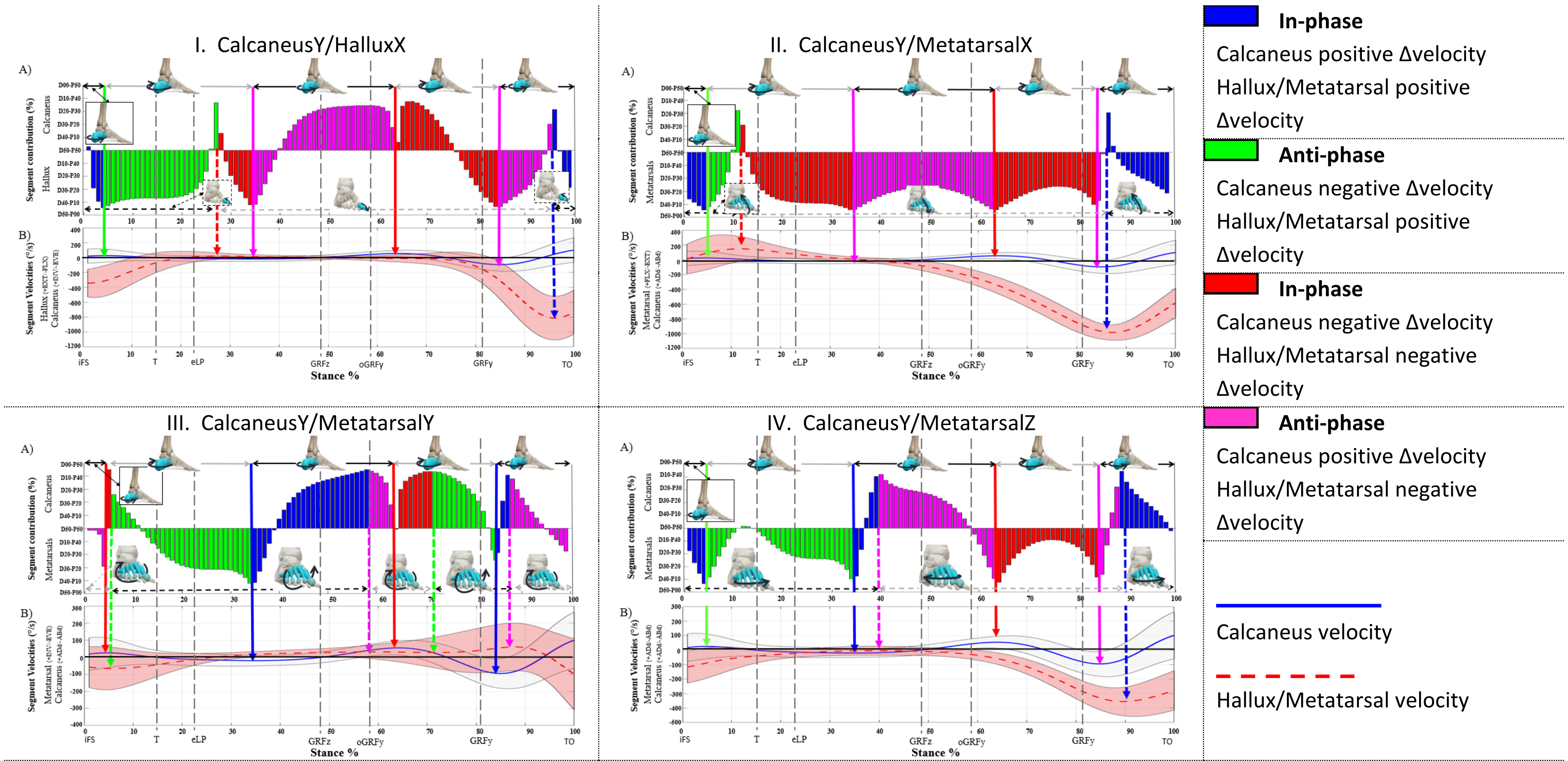
Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force, TO = toe off.

FIGURE 7. 4-2: The average I) CalcaneusY (+INV/-EVE)/HalluxX (+EXT/-FLX), II) CalcaneusY (+INV/-EVE)/MetatarsalX (+EXT/-FLX), III) CalcaneusY (+INV/-EVE)/MetatarsalY (+INV/-EVE), IV) CalcaneusY (+INV/-EVE)/MetatarsalZ (+ADd/-ABd) coordination patterns. Graphs A) display the coupling angle with the contribution of the proximal segment on the top half and distal segment on the bottom half. Graphs B) display segmental velocity. The shaded areas indicate the standard deviation of the segment velocities. The illustrations (calcaneus lateral view and hallux, and metatarsal frontal-lateral view) indicate the change in velocity (segmental accelerations or deceleration). Images courtesy of Visible Body (Argosy Publishing, 2019).

7.4.2 Average coordination patterns influencing calcaneus adduction/abduction

The CalcaneusZ rotational velocity fluctuated around 0°/s from iFS to peak GRFz. The hallux and metatarsal segments were the main contributors to the coordination patterns with CalcaneusZ, the hallux and metatarsal segments displaying larger rotational velocities than CalcaneusZ velocity during the first 40 % of stance (Figure 7.4-3). The CalcaneusZ and HalluxX Δvelocities did not seem to influence each other during the loading phase (Figure 7.4-3 I).

During the last ≈40 % of stance (propulsion phase), HalluxX, MetatarsalX, and MetatarsalZ displayed mostly in-phase coordination with CalcaneusZ and the velocity curves had similar acceleration/deceleration patterns (Figure 7.4-3. I, II & IV). The HalluxX, MetatarsalX, and MetatarsalZ segments moved at a higher velocity than CalcaneusZ during this period. MetatarsalY rotational velocity exhibited large variability between athletes during propulsion in the CalcaneusZ/MetatarsalY coordination pattern (Figure 7.4-3. III).



Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force, TO = toe off.

FIGURE 7. 4-3: The average I) CalcaneusZ (+ADd/-ABd)/HalluxX (+EXT/-FLX), II) CalcaneusZ (+ADd/-ABd)/MetatarsalX (+EXT/-FLX), III) CalcaneusZ (+ADd/-ABd)/MetatarsalY (+INV/-EVE), IV) CalcaneusZ (+ADd/-ABd)/MetatarsalZ (+ADd/-ABd) coordination patterns. Graphs A) display the coupling angle with the contribution of the proximal segment on the top half and distal segment on the bottom half. Graphs B) display segmental velocity. The shaded areas indicate the standard deviation of the segment velocities. The illustrations (calcaneus lateral view and hallux, and metatarsal frontal-lateral view) indicate the change in velocity (segmental accelerations or deceleration). Images courtesy of Visible Body (Argosy Publishing, 2019).

7.5. DISCUSSION

The aim of the project was to test the hypothesis of a distal-proximal coupling relationship between the calcaneus/hallux and calcaneus/metatarsal segments. The coupling relationship between the forefoot and rearfoot segments were thus described and quantified during an unanticipated change of direction task. An athlete is at risk for sustaining a LAS injury when high-velocity inversion and adduction occur around the talocrural and/or subtalar joint (Fong et al., 2012; Gehring et al., 2013a; Kristianslund et al., 2011). In the current study, the maximum CalcaneusY eversion velocity was observed during the loading phase (first ≈20 % of stance) and maximum CalcaneusY inversion and CalcaneusZ adduction velocities were observed during the propulsion. For this reason, the interpretation and analysis of the coordination patterns will mainly focus on the CalcaneusY couplings during the first ≈20 % of stance, and both CalcaneusY and CalcaneusZ couplings during the last ≈60 to 90 % of stance.

Within the loading phase, the CalcaneusY, MetatarsalX, and MetatarsalY change in velocity followed in close succession, indicating a possible strong coupling relationship between CalcaneusY/MetatarsalX and CalcaneusY/MetatarsalY coordination patterns within this period. Similarly, previous research investigating intersegmental foot coordination patterns during straight line running and walking also found strong coupling relationships between CalcaneusY/MetarsalX (Dubbeldam et al., 2013; Pohl & Buckley, 2008; Pohl et al., 2007). However, these studies also found strong coupling between CalcaneusY/HalluxX and CalcaneusY/MetatarsalZ coordination and, in contrast to our findings, weak or no coupling between CalcaneusY/MetatarsalY (Dubbeldam et al., 2013; Pohl & Buckley, 2008; Pohl et al., 2007) and CalcaneusZ/MetatarsalY (Pohl & Buckley, 2008) was observed. The conflicting findings may be as a result of differences in tasks as intersegmental coupling patterns as well as the function and control of certain muscles acting on the foot changes according to demand (Pohl & Buckley, 2008; Zelik et al., 2015). The lack of a strong coupling relationship in the coordination HalluxX/CalcaneusY and HalluxX/CalcaneusZ is surprising as hallux extension is linked to formation of the medial longitudinal arch (MLA) via the windlass mechanism (Hicks, 1954). Our observation of a lack or at least a very weak coupling relationship between HalluxX and CalcaneusY-CalcaneusZ movement may indicate the presence of neuromuscular control of the calcaneus as the forefoot stabilises the foot, as suggested in other research (Nester et al., 2014).

Our observation of positive change in velocity in-phase CalcaneusY/MetatarsalY coordination pattern indicates a simultaneous movement of the calcaneus and metatarsal segments towards inversion acceleration. If forefoot inversion acceleration is not eccentrically controlled, the centre of pressure may shift to the lateral border in the forefoot, causing the forefoot to lose contact with the ground, increasing LAS risk. This observation correlated with previous research that found eccentric deficits in the invertors (tibialis anterior and tibialis posterior) increased ankle instability, possibly increasing LAS risk (Munn, Beard, Refshauge, & Lee, 2003). The in-phase CalcaneusY/MetatarsalY coordination observed during ≈10 % to ≈60 % of the braking phase, indicates a foot moving as a single unit. Nester (2009) suggested that smaller segmental movements are an indication of a stiffer foot and that foot stiffness is a result of increased muscle contraction needed to control segmental movement during higher load situations such as running (and conceivably change of direction tasks). Interestingly, a recent study found intrinsic foot muscle activity decreased during the contact time of a hopping activity while midfoot stiffness increased with an increase in the frequency of the hopping activity. The authors hypothesised that pre-activation of both extrinsic (medial gastrocnemius and soleus) and intrinsic (abductor hallucis) were responsible for the observed midfoot stiffness (Kessler, Lichtwark, Welte, Rainbow, & Kelly, 2020). Thus, the in-phase coordination pattern, resulting from calcaneus eversion deceleration (positive change in velocity) and metatarsal inversion acceleration (positive change in velocity), may indicate neuromuscular control. The intrinsic muscles, abductor hallucis and flexor digitorum brevis, spans the midfoot joints, and could therefore be responsible for the observed in-phase coupling between frontal plane calcaneus and metatarsal segments. Indeed, previous research found that stimulation of the intrinsic muscles, abductor hallucis and flexor digitalis brevis caused calcaneal inversion under loaded conditions (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014). Thus, given that the observed calcaneus positive change in velocity (movement towards calcaneus inversion) was the result of calcaneus eversion deceleration. It is likely that eccentric muscle activity of the intrinsic muscles was responsible for calcaneus eversion deceleration.

A strong coupling relationship in the CalcaneusY/MetatarsalX coordination pattern also emerged within the loading phase and continued throughout stance. Here, metatarsal extension acceleration corresponded with calcaneus eversion acceleration and metatarsal flexion acceleration with calcaneus inversion acceleration. These findings are similar to other studies where foot strike angle and speed of straight-line running and walking varied (Pohl & Buckley, 2008; Pohl et al., 2007). These studies reported a stronger CalcaneusY/MetatarsalX coupling relationship in running compared to walking trials (Pohl et al., 2007). The authors attributed their findings to the ability of the increased soft tissue stiffness around the subtalar joint during running compared to walking and the congruity of the bones of the midfoot, which allows the transfer movement between the segments and the increased soft tissue stiffness around the subtalar joint during running compared to walking (Pohl & Buckley, 2008; Pohl et al., 2007). A recent study confirmed the stiffening of the foot during the push-off phase in walking trials was the result of ankle plantar flexors and plantar intrinsic foot muscle activity (Farris, Birch, & Kelly, 2020). Our use of segmental velocities confirmed the creation of foot stiffness via the forefoot in the sense that MetatarsalX velocity was higher than CalcaneusY velocity. The higher metatarsal velocities possibly indicating the activity of extrinsic and intrinsic ankle plantar and metatarsal flexors which then created distal-proximal stiffness in the foot manifesting as calcaneus inversion/eversion movement.

Previous studies also found strong coupling relationships between CalcaneusY and HalluxX (Dubbeldam et al., 2013) and CalcaneusY and MetatarsalZ coordination patterns in straight line walking and running (Dubbeldam et al., 2013; Pohl & Buckley, 2008; Pohl et al., 2007), which cannot be confirmed here. However, as earlier researchers pointed out, the absence of a coupling relationship may highlight the role of neuromuscular control in dynamic movements (Nester et al., 2014). Calcaneus inversion and/or adduction velocity (and LAS injury risk) is thought to increase when the hallux and metatarsal segments are unable to maintain ground contact (Fong, Hong, et al., 2009a; Gehring et al., 2013a; Huson, 1991; Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005). Maintaining ground contact with a medial centre of pressure implies flexion-eversion of the forefoot (Huson, 1991). The hallux and metatarsal segments responded to the calcaneus inversion and adduction acceleration with rapid hallux flexion, metatarsal flexion, and metatarsal abduction accelerations at the onset of the propulsion phase (≈60 % of stance). However, the expected metatarsal eversion acceleration was not observed; rather, large variability in MetatarsalY velocities was present (Figure 7.4-2 III and Figure 7.4-3 III). The acceleration in hallux and metatarsal flexion can possibly be explained by the fact that when the body weight is carried on the forefoot, the toe flexors and other intrinsic muscles act to resist extension of the toes and stabilise the first metatarsophalangeal and metatarso-cuneiform joints under increasing load (Huson, 1991; Reeser, Susman, & Stern Jr, 1983). Furthermore, the ankle plantar flexors and intrinsic muscles responsible for hallux and metatarsal flexion may play an important role in creating not only forefoot stiffness but also stiffness of the whole foot (Farris et al., 2020). Forefoot stiffness and stability can assist in decreasing LAS risk by enabling the calcaneus to move freely and align with the shank to preserve lateral ankle stability (Fraser, Feger, & Hertel, 2016; Graf & Stefanyshyn, 2012).

During the propulsion phase, the forefoot is also responsible for providing the stiffness and the stability needed as the heel is rising (Donatelli, 1985; Duerinck et al., 2014). We observed metatarsal abduction acceleration, which may suggest that the foot is moving away from metatarsal adduction that is typically associated with foot stiffness (Kelly et al., 2014). This finding may imply that forefoot stiffness for propulsion was achieved in our study by flexion of the hallux and metatarsal segments alone. The transverse plane movement of the metatarsal segment may simply indicate the function of metatarsal abduction acceleration as a means of propelling the body towards the changing direction. Similarly, the large metatarsal frontal plane velocity variation between athletes during propulsion suggests that different strategies were used at an individual level for propulsion, potentially placing some athletes at a higher risk for sustaining LAS injury. It thus seems worthwhile to investigate the effect of an intrinsic and extrinsic foot muscle strengthening programme on LAS injury mechanisms not only at a group level, but at an individual level.

A shortcoming of this study was the averaging of the segmental velocities, as large variability was especially observed in the frontal plane metatarsal velocities. The use of this statistical method was needed to establish prospective coordination patterns, but interpretation of the coordination patterns should be approached with caution in this particular instance. It is also known that the vector coding technique is sensitive to small changes in segment movements (Chang et al., 2008). To compensate for these known concerns and ensure reliable kinematic data, a multi-segmental foot model with high reliability was used and markers were applied to all athletes by the same trained investigator (Caravaggi et al., 2011). Lastly, no electromyographic investigations were undertaken as part of this study; hence, muscular function is only inferred from the kinematic results and other research studies.

7.6. CONCLUSION

The forefoot was the first point of contact during unanticipated changes of direction and the calcaneus remained off the floor during the whole stance phase. The coordination patterns reported here were overall dissimilar to previous straight line running and walking studies. In the loading phase, we observed close coupling between CalcaneusY/MetatarsalX, CalcaneusY/MetatarsalY, and CalcaneusZ/MetatarsalY. CalcaneusZ velocity was close to 0°/s and was not considered to increase LAS risk here. The Metatarsal segment velocity was higher after the initial foot strike, possibly indicating the importance of the forefoot maintaining contact with the ground, stiffening, and stabilising the foot. Calcaneus eversion deceleration was observed during the braking phase, possibly indicating neuromuscular control of the segment through eccentric muscle contractions. Consistent coupling between MetatarsalX and CalcaneusY was observed for the whole stance phase confirming previous study findings and highlighting the importance of forefoot stiffness and stability.

During propulsion, calcaneus inversion and adduction reached their respective maximum velocities. Suggesting a distal-proximal coupling relationship. However, the forefoot did not follow the calcaneus change in velocity, but instead increased hallux and metatarsal flexion accelerations. The flexion accelerations of the forefoot and hallux was likely a function of the active and passive structures acting in unison to increase foot stiffness and stability. The expected acceleration towards metatarsal eversion (to remain medial centre of pressure in the forefoot) and metatarsal adduction (associated with forefoot stability) was not observed.

Instead, large MetatarsalY variability and metatarsal abduction acceleration were observed. We suggest that different propulsion strategies at an individual level may increase LAS in certain athletes and that strengthening the muscles acting on the foot may assist in decreasing the risk for LAS, both of which require further investigation.

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CHAPTER 8

THE EFFECT OF A 16-WEEK FOOT-MUSCLE SPECIFIC STRENGTH TRAINING INTERVENTION PROGRAMME ON FOOT FUNCTION, NON-CONTACT ANTERIOR CRUCIATE LIGAMENT INJURY AND LATERAL ANKLE SPRAIN RISK FACTORS: PROOF OF CONCEPT

8.1. INTRODUCTION

Successful injury prevention strategies need to statistically and clinically reduce the risk factors associated with the injury mechanism(s) (Engebretsen & Bahr, 2009). Non-contact anterior cruciate ligament rupture (ACLR) and lateral ankle sprain (LAS) injury prevention research has identified the risk factors associated with these lower limb injuries and current injury prevention programmes are relatively successful in decreasing injury occurrence (Noyes & Barber-Westin, 2014; Schiftan, Ross, & Hahne, 2015). Successful injury prevention programmes have a multifaceted approach (Shultz et al., 2015) of which strength training form an important component (Mehl et al., 2018). Resistance training included in ACLR prevention programmes mainly focus on strengthening the musculature above the knee (Mehl et al., 2018). The strength training component of an LAS injury prevention programme mainly involve strengthening the muscles crossing the ankle joint (mainly extrinsic muscles) (Kaminski et al., 2013). However, recent lower limb injury prevention researchers proposed a ‘bottom-up’ approach which includes strengthening the muscles crossing the ankle as well as the foot joints, thus both extrinsic and intrinsic foot muscles (Nigg, Baltich, Federolf, Manz, & Nigg, 2017). At present it is unclear whether a 'bottom-up' approach would be effective in ACLR and LAS injury prevention.

The effectiveness of an intervention programme is often only apparent with longitudinal studies (Cook & Ware, 1983). Biomechanical data can provide valuable interim information regarding the effectiveness of an intervention programme. It is thus important to first test the validity hypothesis of a 'bottom-up' approach by establishing the possible influence of the rearfoot on the shank (ACLR risk) and forefoot on the rearfoot (LAS risk). In chapters six we found that in females, a distal-proximal coupling relationship exists between frontal- and transverse plane calcaneal movement and transverse plane shank movement. The distal-proximal coupling between the calcaneus and transverse shank rotations could indicate the transfer of potentially injurious rotational forces from the foot to the knee during unanticipated change of direction tasks. Similarly, in chapter seven we found that the forefoot may play a role in providing stiffness and stability during change of direction tasks, potentially decreasing LAS risk. Functional anatomy and previous research has shown that the calcaneus, the forefoot and the arches of the foot are under neuromuscular control, creating foot stiffness (Farris, Birch, & Kelly, 2020; Kessler, Lichtwark, Welte, Rainbow, & Kelly, 2020). It thus seems possible that modulating segmental movement of the foot may, in theory, influence risk factors for both the ACLR and LAS injury. However, it is still unclear whether training the foot muscles would make a difference to movement of foot segments or any of the risk factors associated with ACLR and LAS during unanticipated change of direction tasks.

Motion caption and force data are often utilized to determine biomechanical adaptations to intervention strategies. However, longitudinal and motion caption research are expensive and time-consuming methods (Cook & Ware, 1983; Mullineaux & Wheat, 2017). Randomised controlled pilot studies and/or proof of concept studies are therefore valuable to test the feasibility and effectiveness of an intervention before considering its implementation as a larger scale intervention strategy (Thabane et al., 2010). Females, especially those taking part in court sports, are particularly prone to sustaining non-contact anterior cruciate ligament rupture (ACLR) and/or lateral ankle sprain (LAS) injury (Doherty et al., 2014; Gamada, 2014; Willems, Witvrouw, Delbaere, Philippaerts, et al., 2005). The risk for sustaining an ACLR and/or LAS increase during unanticipated deceleration tasks (change of direction, stopping or landing) in the presence of a large ground reaction force (GRF) that does not act through the joint's centre (Fong, Ha, Mok, Chan, & Chan, 2012; Olsen, Myklebust, Engebretsen, & Bahr, 2004; Stuelcken, Mellifont, Gorman, & Sayers, 2015). Typical lower limb biomechanical injury risk factors associated with ACLR include an extended knee, large knee valgus angle and -moment and tibial rotation (Gamada, 2014; Shultz et al., 2015; Stuelcken et al., 2015). LAS injury risk factors involve a large ankle inversion and adduction angle at initial foot strike that rapidly increases, creating a forceful inversion and/or adduction moment around the subtalar and talocrural joints (Ashton-Miller, Ottaviani, Hutchinson, & Wojtys, 1996; Chu et al., 2010; Fong, Chan, Mok, Yung, & Chan, 2009; Kristianslund, Bahr, & Krosshaug, 2011).

Current mainstream ACLR and LAS injury prevention programmes target the hip musculature (particularly extensors and abductors) and are relatively successful (McCriskin, Cameron, Orr, & Waterman, 2015; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Shultz et al., 2015; Vriend, Gouttebarge, van Mechelen, & Verhagen, 2016). In addition, prophylactic programmes intended to decrease LAS re-injury risk aim to increase the strength of the muscles controlling ankle inversion, eversion, extension, and flexion (Kaminski et al., 2013). However, previous research has found that segmental movement of the foot play a role in both ACLR and LAS injury risk. For example, excessive, dynamic rearfoot pronation and large rearfoot eversion angles have been linked to increased knee valgus angles, thus increasing ACLR injury risk (Holland, 2016; Kagaya, Fuji, & Nishizono, 2013; McLean, Lipfert, & Van Den Bogert, 2004). LAS injury risk has been linked to increased hallux extension range of motion (Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005) and forefoot instability (Fong, Hong, et al., 2009). To date, conventional ACLR and LAS injury prevention programmes do not aim to strengthen the muscles acting on the foot directly (Noyes & Barber-Westin, 2014; Taylor, Ford, Nguyen, Terry, & Hegedus, 2015).

Leading researchers in the field of foot biomechanics suggest that strengthening the muscles acting on the foot can influence foot segmental movement, subsequently stabilizing the foot and ankle, decreasing LAS risk. Increasing foot stiffness and stability also has the potential to decrease excessive movement rearfoot pronation subsequently decreasing the moments around the subtalar and talocrural joints, influencing the transfer of rotation to the knee and possibly decreasing ACLR risk (Fraser, Feger, & Hertel, 2016a, 2016b; McKeon, Hertel, Bramble, & Davis, 2014; Nigg, Baltich, Federolf, Manz, & Nigg, 2017). Furthermore, increasing foot muscle strength has been found to increase sprint as well as jumping performance (De Villiers, 2014; Hashimoto & Sakuraba, 2014). Improving athletic performance can have a profound impact on adherence to an injury prevention programme, which has often been noted as a significant factor in the success of injury prevention programmes (Riva, Bianchi, Rocca, & Mamo, 2016; Shultz et al., 2015).

A recent search of the literature could not find a prophylactic programme that investigated the effect of strengthening the foot musculature specifically on ACLR and/or LAS injury risk factors. This study aims to conduct a randomised controlled pilot study to determine proof-of-concept of foot specific strength training as part of an ACLR and LAS injury prevention strategy. The purpose is to explore the effect of strengthening the foot musculature on: 1) performance and kinetic parameters associated with change of direction tasks as well as ACLR and LAS injury risks; 2) the movement of the medial longitudinal arch (MLA) and the metatarsal anterior transverse arch (MetATA); 3) the change in segmental movement of the foot; and 4) risk factors associated with ACLR and LAS during unanticipated changes of direction. In addition, the proof-of-concept study aims to: 1) identify the biomechanical variables most likely to show adaptation as a result of the intervention; 2) identify the variables most likely to be different between participants and potentially influence the intervention effect; and 3) provide sample size guidelines for future randomised controlled trials.

8.2. METHODS

8.2.1 TRIAL DESIGN

A randomised controlled intervention study was undertaken during the 2018 competition season. Athletes were recruited from domestic clubs and high schools. The aim, duration, intervention programme, and exclusion criteria were explained to all athletes recruited. Eligible athletes gave written informed consent prior to participation. Ethical approval was granted from the University Human Ethics Committee of our institution (SOA 16/77).

Data collection took place before the start of the competitive season and intervention training (Figure 8.3-1), as well as after completion of the intervention 16 weeks later. At baseline, all athletes completed a demographics questionnaire and provided information regarding general health (Cardinal, Esters, & Cardinal, 1996), previous injury (Clarsen, Myklebust, & Bahr, 2013), and current sports participation. In addition, court-sport exposure hours and lower extremity injury incidences were recorded by the athletes and was submitted weekly for the duration of the intervention.

8.2.2 PARTICIPANTS

Female athletes experienced in court sports (netball, volleyball, racquet sports, and futsal) and playing in a local competitive league were recruited. All athletes were injury-free and pain-free six months preceding the pre-intervention test and had no musculoskeletal or neurological condition affecting movement or foot function. Athletes were excluded if they had a body mass index (BMI) higher than 30, were pregnant, or had been pregnant. The athletes were matched for sport and BMI

(Table 8.2-1), and then randomly assigned to either the training group (TG) or control group (CG) (Figure 8.2-1).

TABLE 8. 2-1: The mean (SD) of athletes' demographic data at pre-intervention (n = 24).

	Control group	**Training group**	
	Completed Intervention (n = 10)	**Completed Intervention (n = 8)**	**p-value**
Age (years)	17.50 (1.43)	17.25 (1.92)	0.77
Mass (kg)	65.5 (10.3)	62.86 (6.07)	0.72
Height (cm)	1.69 (0.06)	1.66 (0.07)	0.50
Body mass index (kg/m^2)	23.03 (3.33)	22.73 (1.98)	1.00
Court-sport exposure (min/week)	213.72 (72.49)	283.15 (64.54)	0.86

Note: p-value was calculated between athletes completing the intervention (CG = 10 and TG = 8).

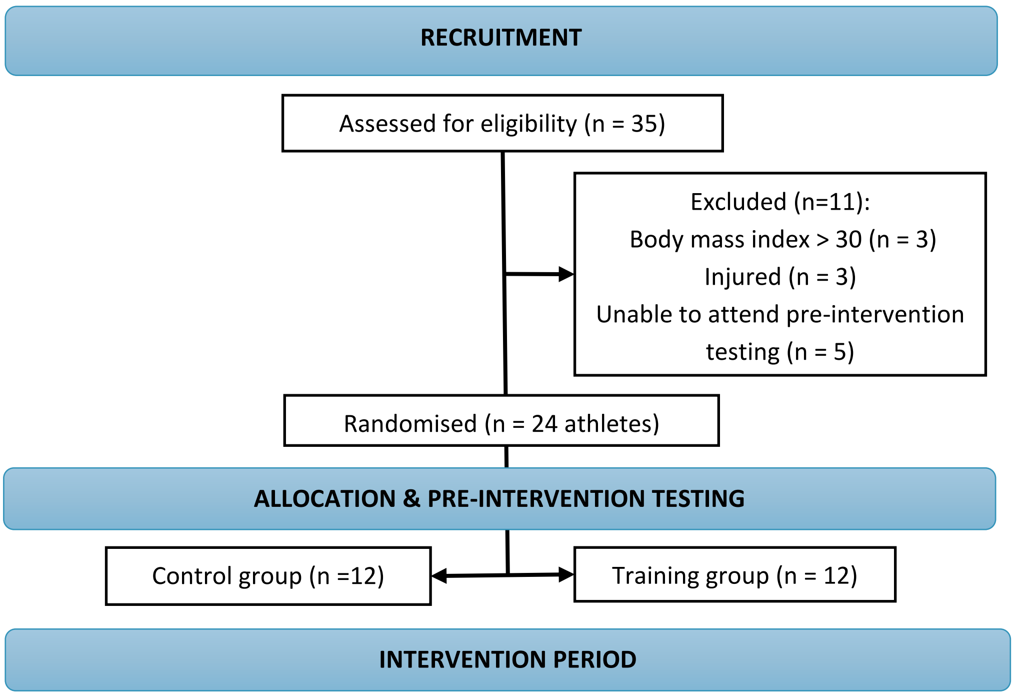

FIGURE 8. 2-1: Flowchart of the selection process and the number of athletes excluded and randomly allocated to the control or training group

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8.2.3 INTERVENTION

TG underwent a progressive 16-week intervention programme (Table 8.2-2) strengthening the muscles acting on the foot in addition to normal sports training. CG performed common sports training only and was not aware of the specific exercises performed by TG. TG performed the foot and lower leg strengthening exercises three times per week; each session lasted 20 to 25 minutes. The researcher supervised one session each week. TG received a weekly email with a link to the two unsupervised sessions of that week. The linked document included a Graphics Interchange Format animation (GIF) and written instructions for every exercise. On completion of the session, a 'finish' button was clicked, and an email sent to the researcher. In this way compliance was monitored.

The intervention programme was aimed at improving foot mobility, control, and strengthening of the muscles acting on the foot. The activities were mostly designed to enable home-based training and limit the use of machine-based strengthening equipment (the researcher provided the resistance bands and exercise balls that were necessary to complete some of the exercises). The mobility exercises aimed to increase the range of motion of the hallux, metatarsals and midfoot. Creating mobility in the fore- and midfoot regions is not only important for normal gait, but also for assisting shock attenuation during deceleration tasks (Holowka, O'Neill, Thompson, & Demes, 2017; Huson, 1991). Increasing the extension range of motion of the first metatarsal phalangeal joint (hallux) is important for normal gait as well as medial longitudinal arch function (Hicks, 1954; Nester, Jarvis, Jones, Bowden, & Liu, 2014). Foot mobility and dissociation exercises (the ability to move one body part in isolation from another) (Fernandes-de-las-

Penas, Cleland, & Dommerholt, 2016) were included to increase the athlete's awareness of the foot as a multisegmental structure. It is thought that increasing awareness (proprioception) facilitates the activation of skeletal muscles (Ribeiro & Oliveira, 2011). The exercises that were included in the strengthening component of the intervention protocol were based on functional anatomy (see Chapter 2), previous research investigating muscle activation (Gooding, Feger, Hart, & Hertel, 2016), and a foot muscle specific intervention aimed at preventing falls in the elderly (Mickle, Caputi, Potter, & Steele, 2016) (see Table 8.2-3).

The programme increased in complexity of the exercises, intensity, and endurance over four phases (I-IV). The main aim of Phase I was to familiarise the athletes with the exercises and educate them about foot function. The exercises were simple tasks to improve motor learning and ensure activation of the muscles the exercises were intended to train. Additionally, the exercises were non-weight bearing and a light resistance flexiband was used. In Phase II exercise progressions were achieved by increasing resistance. The exercises were performed weight bearing, single leg and the flexiband resistance was progressed to medium strength where applicable. The exercises were further progressed by increasing complexity. In Phases III & IV, the exercises were progressed in similar fashion by increasing resistance to the strongest flexiband and increasing exercise complexity. A progressive strength endurance component was also introduced during phase III and maintained in Phase IV. For a complete description of the intervention programme refer to Appendix G.

TABLE 8. 2-2: 16-week progressive strengthening programme for muscles acting on the foot.

PHASE I: Week 1-4				PHASE II: Week 5-8				PHASE III: Week 9-12				PHASE IV: Week 13- 16			
Exercise	**Ss**	**Rs**	***Tempo**	**Exercise**	**Ss**	**Rs**	**Tempo**	**Exercise**	**Ss**	**Rs**	**Tempo**	**Exercise**	**Ss**	**Rs**	**Tempo**
WARM-UP AND MOBILITY															
Metatarsal opener	2	5	1-3-1-0	**Metatarsal opener**	2	5	1-3-1-0	**Metatarsal opener**	1	10	1-3-1-0	**Metatarsal opener**	1	10	1-3-1-0
Toe extension / flexion	2	5	1-3-1-0	**Toe extension / flexion**	1	10	1-3-1-0	**Toe extension / flexion**	1	10	1-3-1-0	**Toe extension / flexion**	1	10	1-3-1-0
DISSOCIATION															
Individual toe tap	2	5	1-1-1-1	**Individual toe press**	2	10	1-3-1-1	**Toe wave**	1	10	1-3-1-1	**Toe wave**	1	10	1-3-1-1
Toe splay	2	10	1-1-1-1												
STRENGTH															
Lightest strength flexi-band				**Medium strength flexi-band**				**Strongest flexi-band**							
Big toe press - seated	2	10	1-1-1-3	**L/R = Short foot and R/L = Inversion**	2	10	SF: 1-3-0-0 Inv: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Inversion**	3	10 +2/ wk	SF: 1-3-0-0 Inv: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Inversion**	3	20	SF: 1-3-0-0 Inv: 1-1-3-0
Short foot – seated	2	10	1-1-1-3												
Inversion – seated	2	10	3-0-1-2												
Eversion – seated	2	10	3-0-1-2	**L/R = Short foot and R/L = Eversion**	2	10	SF: 1-3-0-0 Eve: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Eversion**	3	10 +2/ wk	SF: 1-3-0-0 Eve: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Eversion**	3	20	SF: 1-3-0-0 Eve: 1-1-3-0
Dorsiflexion - seated	2	10	3-0-1-2												
Calf raise	2	10	3-0-1-2	**Single leg calf raises**	2	10	3-0-1-3	**Single leg hop**	3	10	1-0-3-0	**Continues single leg hop**	3	20	1-0-3-0

** Tempo: Concentric - Isometric/Stretch hold – Eccentric – Isometric/Stretch hold; Ss=Sets; Rs=Reps; SF=Short foot; Inv=Inversion; Eve=Eversion; wk=week*

TABLE 8. 2-3: The muscles targeted in the strength component of the intervention protocol (Gooding et al., 2016; Mickle et al., 2016; Mulligan & Cook, 2013).

	EXERCISES OF THE STRENGTH COMPONENT						
	Big toe press	**Toe spread**	**Short foot**	**Inversion**	**Eversion**	**Dorsiflexion**	**Calf raises and hops**
Extrinsic: Anterior compartment							
Tibialis anterior				✓		✓	
Extensor digitorum longus						✓	
Extensor Hallucis longus						✓	
Extrinsic: Superficial posterior compartment							
Plantaris							✓
Gastrocnemius							✓
Soleus							✓
Extrinsic: Deep posterior compartment							
Tibialis Posterior				✓			✓
Flexor Digitorum Longus				✓			✓
Flexor Hallucis Longus	✓			✓			✓
Extrinsic: Lateral compartment							
Fibularis Longus					✓		✓
Fibularis Brevis					✓		✓
Intrinsic: First plantar layer							
Abductor Digiti Minimi		✓	✓				
Flexor Digitorum Brevis		✓	✓				
Abductor Hallucis		✓	✓				
Intrinsic: Second plantar layer							
Quadratus Plantae		✓	✓				
Lumbricals		✓	✓				
Intrinsic: Third plantar layer							
Flexor Hallucis Brevis	✓	✓	✓				
Flexor Digiti Minimi		✓	✓				
Adductor Hallucis		✓	✓				
Intrinsic: Fourth plantar layer							
Plantar Interossei		✓	✓				

8.2.4 DATA COLLECTION AND PROCESSING

Pre- and post-intervention 3D motion and force data were collected and processed using the protocol and methods as described in Chapters 5 and 6. In addition MATLAB (R2018a, Mathworks, MA USA) was also used to determine the maximum GRFz, medial-lateral ground reaction force (GRFx), and anterior-posterior (GRFy) braking and propulsion force values. The braking phase was defined as the period when the GRFy was negative and the propulsion phase where GRFy was positive. Braking and propulsion impulses were calculated from the area under their respective GRFy curves. Transient GRFz, GRFx, and braking GRFy from the respective GRF curves identified as the rapidly developing spike in the specific GRF curve, occurring within the first 50ms of stance (Lieberman et al., 2010; Whittle, 1999). The loading rates were calculated using the magnitude of the specific transient GRF divided by the time to transient. The period between transient braking force and peak propulsion force was identified as the period of the movement where injuries are most likely to occur as maximum GRF and joint angles were observed in this period.

Tri-planar segment angles were calculated in Visual 3D™ using Euler angles and following standard X-Y-Z Cardan sequence. Maximum angles occurring between transient braking force and peak propulsive force was identified in MATLAB between the following foot segments: hallux-metatarsal, metatarsal-midfoot, midfoot-calcaneus, and calcaneus-shank. Movement in the sagittal plane was classified as extension (+EXT)/flexion (-FLX) for the foot segment angles. Sagittal plane movement for the ankle was classified as dorsiflexion (+DF)/plantarflexion (-PF). Movement of the ankle and foot segments in the frontal plane were classified as inversion

(+INV)/eversion (-EVE), and in the transverse plane as adduction (+ADd)/ abduction (-ABd).

The metatarsal anterior transverse arch (MetATA) and medial longitudinal arch (MLA) length and height were measured using the methods described in previous research (Duerinck, Hagman, Jonkers, Van Roy, & Vaes, 2014; Jenkyn, Shultz, Giffin, & Birmingham, 2010). The MetATA length was the Euclidean distance between the first- and fifth metatarsal head markers. The MetATA height is the distance between the second metatarsal head marker and its perpendicular projection onto the line between the markers at the head of the first and fifth metatarsals (Duerinck et al., 2014). The length of the MLA was the Euclidean distance between the markers of the head of the first metatarsal and the proximal calcaneus. The height of the MLA was calculated as the distance between the marker at the navicular tuberosity and its perpendicular projection onto the line between the markers at the head of the first metatarsal and the proximal calcaneus (Jenkyn et al., 2010).

The foot-to-floor angle indicated the dynamic foot strike angle between the foot and the floor (negative indicates forefoot contact with floor). The foot-to-floor angle was calculated as the sagittal plane angle between the floor and the virtual foot segment, subtracted from the foot-to-floor angle in the static trial.

8.2.5 Outcome measures

8.2.5.1 Performance and kinetic variables

Performance outcomes were derived from the cutting trial and included the approach speed, speed at initial foot strike, exit speed, and stance time (Fox, 2018; McLean et al., 2004; Spiteri et al., 2015). GRF variables are associated with performance, and both ACLR and LAS injury risk (Raina & Nuhmani, 2014; Serpell, Scarvell, Ball, & Smith, 2012; Spiteri et al., 2015). Thus, kinetic outcomes included maximum values for GRFx, GRFy braking force, GRFz, and loading rates for GRFz, GRFy and GRFx curves.

8.2.5.2 Kinematic and injury risk variables

ACLR and LAS injury risk increases as the magnitude of the resultant GRF increases (Raina & Nuhmani, 2014; Serpell et al., 2012). Thus, the following kinematic variables associated with ACLR and LAS injury risk were measured within the period between transient braking force and peak propulsion force (Table 8.2-4).

To establish adaptations to forefoot stiffness and stability, a risk factor associated with LAS injury risk (Fong, Hong, et al., 2009), the deformation of the forefoot was evaluated. Forefoot stiffness is characterised by the length and height of the MetATA (Duerinck et al., 2014). The extension range of motion of the hallux is also associated with an increased risk of LAS injury, therefore sagittal plane hallux-metatarsal angle was assessed (Willems, Witvrouw, Delbaere, De Cock, et al., 2005). Other LAS injury risk factor variables were the maximums of the ankle inversion angle and ankle adduction angle (Hollis, Blasier, & Flahiff, 1995; Panagiotakis, Mok, Fong, & Bull, 2017; Renstrom et al., 1988; Stormont, Morrey, An, & Cass, 1985), maximum ankle inversion and adduction velocity (Ashton-Miller et al., 1996; Chu et al., 2010;

Fong, Chan, et al., 2009; Kristianslund et al., 2011), and the maximum lengths of the ankle inversion and -eversion moment arms (Fong, Chan, et al., 2009).

ACL injury risk is associated with dynamic rearfoot pronation and rearfoot eversion angles (Holland, 2016; Kagaya et al., 2013; McLean et al., 2004). Thus, changes to the maximum sagittal, frontal, and transverse plane motions between the metatarsal-midfoot and midfoot-calcaneus segment angles was examined. The changes in the metatarsal-midfoot and midfoot-calcaneus segment angles also manifests as changes to the length and height of the medial longitudinal arch (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014). As movement of the calcaneus is linked to shank movement (Hicks, 1953; Huson, 1991), and transverse plane shank movement to both ACLR (Mokhtarzadeh et al., 2015) and LAS (Panagiotakis et al., 2017) injury risk, the triplanar movement of the calcaneus-shank angles were observed. Additional ACLR injury risk factor variables were the maximums of the knee valgus and extension angle, knee internal rotation velocity and length of the valgus moment arm (Gamada, 2014; Shultz et al., 2015; Stuelcken et al., 2015).

TABLE 8. 2-4: Injury risk variables and the joint it is associated with.

INJURY RISK VARAIBLES		
VARIABLE	**ACL**	**LAS**
Ground reaction force (BW)		
Peak medial-lateral		✓
Peak braking	✓	✓
Peak vertical		✓
Loading rate (BW/s)		
Medial-lateral	✓	✓
Anterior-posterior	✓	✓
Vertical	✓	✓
Foot stability (Arches maximum height and length (cm))		
MLA Height (cm)	✓	
MLA Length (cm)	✓	
MetATA Height (cm)		✓
MetATA Length (cm)		✓
Foot segmental angle maximums (°)		
Min. HalMet (+EXT/-FLX)		✓
Max. MetMid (+EXT/-FLX)	✓	
Min. MetMid (+INV/-EVE)	✓	
Min. MetMid (+ADd/-ABd)	✓	
Max.MidCal (+EXT/-FLX)	✓	
Min. MidCal (+INV/-EVE)	✓	
Min. MidCal (+ADd/-ABd)	✓	
Max. CalShk (+DF/-PF)	✓	✓
Min. CalShk (+INV/-EVE)	✓	✓
Min. CalShk (+ADd/-ABd)	✓	✓
Knee and Ankle angle (°)		
Max. Ankle frontal plane (+INV/-EVE)		✓
Max. Ankle transverse plane (+ADd/-ABd)		✓
Min. Knee frontal plane (+VAR/-VAL)	✓	
Min. Knee sagittal plane (-EXT/+FLX)	✓	
Knee and Ankle velocity (°/s)		
Max. Ankle frontal plane (+INV/-EVE)		✓
Max. Ankle transverse plane (+ADd/-ABd)		✓
Knee transverse plane (+IR/-ER)	✓	
Knee and Ankle moment arm (cm)		
Max. Ankle frontal plane (+INV)		✓
Max. Ankle frontal plane (-EVE)		✓
Min. Knee frontal (+VAR/-VAL)	✓	

Notes: MLA = medial longitudinal arch, MetATA = metatarsal anterior transverse arch; HalMet = Hallux-Metatarsal; MetMid = Metatarsal-Midfoot; MidCal = Midfoot-Calcaneus; CalShk = Calcaneus-Shank; +EXT = Extension; -FLX = Flexion; +INV = Inversion; -EVE = Eversion; +ADd = Adduction; -ABd = Abduction; +VAR = Varus; -VAL = Valgus.

8.2.6 STATISTICAL ANALYSES

Statistical analyses were conducted using SAS 9.4 (TS Level 1M6). A one-way ANOVA was used to compare age, height, mass, BMI, and minutes exposed to court-sports activity per week between groups. An unbalanced two-way analysis of variance (group x session) was used to determine the effect of the intervention programme on the outcome variables, with Tukey procedures used during post-hoc testing (**α** level set as $p < 0.05$). However, a decision was made to relax the α-level to $p \leq 0.25$ on account of a small sample size and the impact of a third factor (Thiese, Ronna, & Ott, 2016). The athletes recruited for this study are all competent athletes and a certain level of foot and lower limb muscle strength can be expected (Pohl, Messenger, & Buckley, 2007) which could obscure the effect of the foot muscle specific training intervention.

As the minimum and/or maximum segmental angle values only provide information about a single point in stance, coefficient of determination (R^2) values were calculated. R^2 is a goodness-of-fit measure and uses a fitted linear regression model to explain the variance of the post-intervention values from the linear regression model. In this way, the segmental angle adaptation from pre-intervention to post-intervention can be compared over the entire stance phase within each group. R^2 values are between 0 and 1, where a value closer to 0 indicates that the curves have changed completely. Differences in R^2 can thus indicate changes between pre- and post-intervention time points within the group.

8.2.7 SAMPLE SIZE CALCULATION

The means and standard deviation for each variable gained from the pilot study were used to calculate the sample size for a future randomised control trial. The sample size was calculated using the equation and conventional multipliers described in earlier research (Noordzij et al., 2010) (Equation 8.2-1).

$$n = \frac{2[(a+b)^2\sigma^2]}{(\mu_1 - \mu_2)^2}$$

EQUATION 8.2-1: Sample size calculation (Noordzij et al., 2010).

n = the sample size for each group
μ_1 is the mean of CG
μ_2 is the mean of TG
$\mu_1 - \mu_2$ is the difference detected between the means of the two groups
σ^2 is the variance (SD) of the CG
a is the conventional multiplier for α of 0.05 = 1.96
b is the conventional multiplier for β of 0.2, a power of 0.80 = 0.842

8.3. Results

8.3.1 Participants

From the 35 athletes recruited, 24 athletes were eligible to take part in the intervention. Two TG athletes withdrew within Phase I of the intervention (one due to unrelated hairline fracture, and one due to a change in circumstances). Four participants (two from CG and two from TG) were unable to attend the post-intervention test. Thus, 18 athletes (CG, n = 10; TG, n = 8) were assessed at both the pre- and post-intervention testing (Figure 8.3-1).

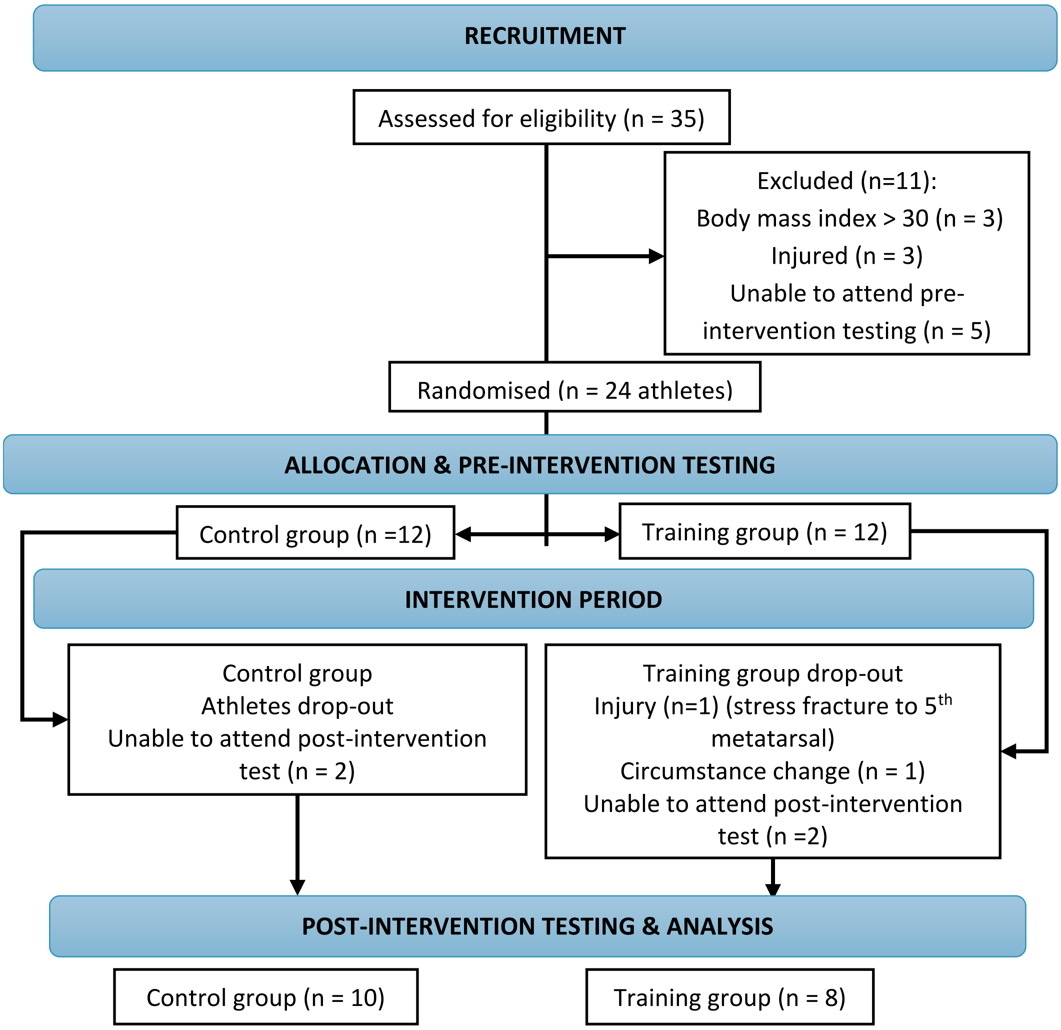


FIGURE 8. 3-1: Flowchart of the selection process and the number of athletes completing the intervention period and number of athletes analysed at post-intervention testing.

TABLE 8. 3-1: The mean (SD) demographic and court-sport exposure data of the control and training group as well as the athletes that dropped out of each group.

	Control group		Training group		
	Completed Intervention (n = 10)	Dropouts (n = 2)	Completed Intervention (n = 8)	Dropouts (n = 4)	[I]p-value
Age (years)	17.50 (1.43)	16.00 (0.00)	17.25 (1.92)	17.25 (0.83)	0.77
Mass (kg)	65.5 (10.3)	70.10 (2.90)	62.86 (6.07)	65.13 (14.23)	0.72
Height (cm)	1.69 (0.06)	1.68 (0.02)	1.66 (0.07)	1.70 (0.05)	0.50
Body mass index (kg/m^2)	23.03 (3.33)	24.87 (1.62)	22.73 (1.98)	22.27 (3.87)	1.00
Court-sport exposure (min/week)	213.72 (72.49)	395.97 (164.20)	283.15 (64.54)	245.47 (68.40)	0.86

[I]*Note: p-value was calculated between athletes completing the intervention (CG = 10 and TG = 8).*

8.3.2 Compliance

Eight of the twelve athletes (67 %) in TG completed the 16-week intervention. The mean supervised attendance was 89 % (range 69 % to 97 %) and 72 % (range 58 % to 84 %) of all the unsupervised sessions were recorded as completed. Compliance to unsupervised sessions were 58 % in phase one (range 27 % to 100 %), 79 % in phase two (range 42 % to 100 %), 84 % in phase three (range 42 % to 100 %), and 68 % in phase four (range 20 % to 100 %). The reason for participant dropout is included in Figure 8.3-1. One participant from TG withdrew within the first week of Phase II because of a clinically diagnosed stress fracture in the 5th metatarsal. The stress fracture was unlikely the result of the intervention programme as no high-impact exercises were included during Phase I of the intervention.

8.3.3 PERFORMANCE AND KINETIC VARIABLES

8.3.3.1 PERFORMANCE VARIABLES

8.3.3.1.1 EFFECT OF INTERVENTION

There were no significant differences between groups pre- to post-intervention for any performance parameter (Table 8.3-1).

8.3.3.1.2 GROUP DIFFERENCES BEFORE INTERVENTION

Stance time was significantly different between groups prior to the intervention (CG = 0.23 ± 0.02 sec, TG = 0.20 ± 0.02 sec; $p < 0.05$).

8.3.3.1.3 SAMPLE SIZE

All performance parameters were significantly different post-intervention compared to pre-intervention (Approach speed: CG = + 17 %, TG = + 14 %; $p < 0.001$, Speed at initial foot strike: CG = + 16 %, TG = + 14 %; $p < 0.001$, and exit speed: CG = + 7 %, TG = + 9 %; $p < 0.001$). The stance time for both groups was also significantly shorter (CG = - 10 %, TG = - 5 %; $p < 0.05$) post- compared to pre-intervention.

The power analysis calculations showed that in order to find a significant difference in TG pre- to post-intervention for the chosen performance parameters there needs to be a minimum sample size of N = 5 for approach speed, N = 4 for speed at initial foot strike, N = 12 for exit speed and N = 35 for stance time (Table 8.3-2).

8.3.3.2 Kinetic Variables

8.3.3.2.1 Effect of intervention

There were no significant differences between groups pre- to post-intervention for any ground reaction force or loading rate variables.

8.3.3.2.2 Group differences before intervention

There was a trend towards significant difference in the peak medial-lateral GRF between groups pre-intervention (CG = 0.75 ± 0.12 BW, TG = 0.85 ± 0.09 BW; $p = 0.10$). The loading rates were similar between groups pre-intervention.

8.3.3.2.3 Sample size

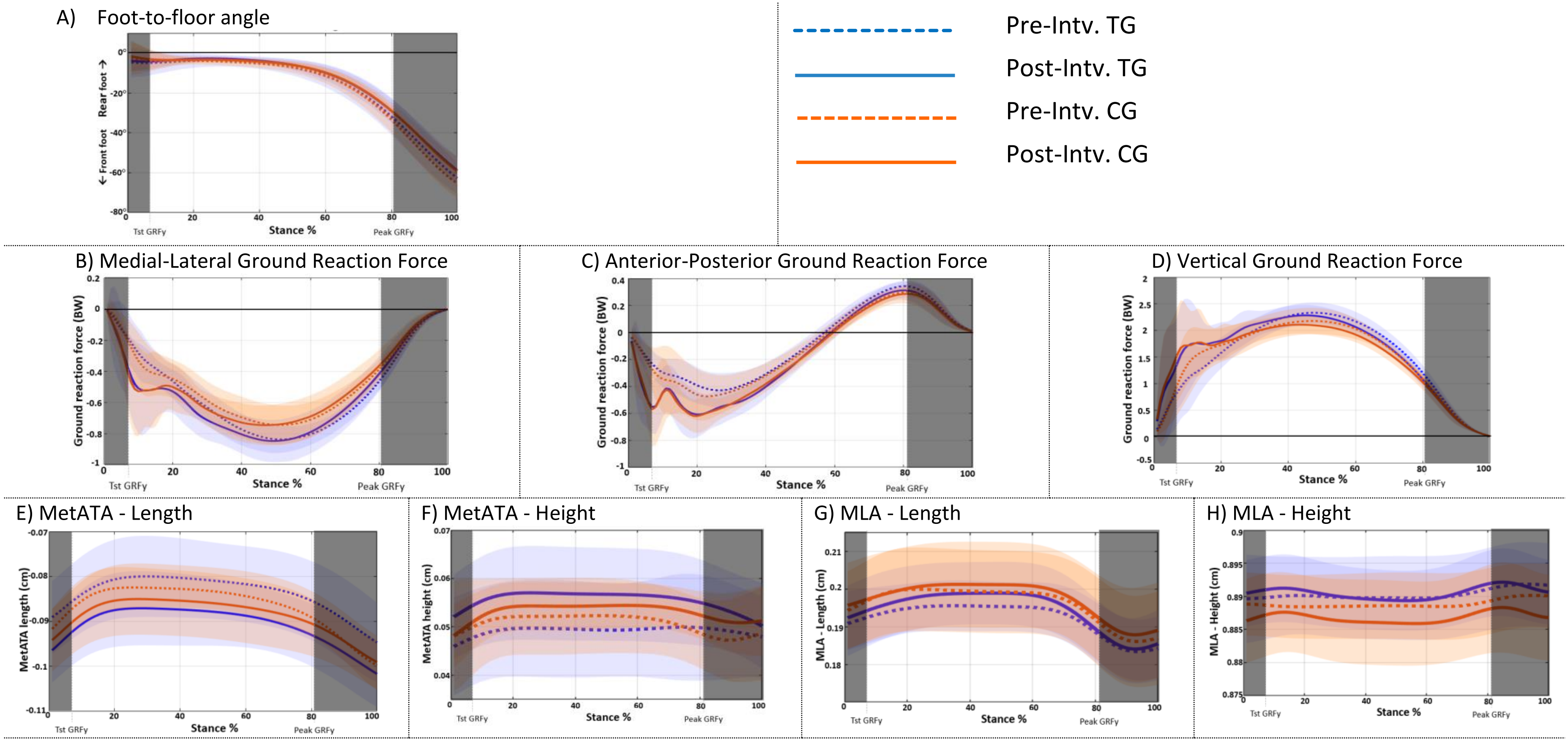
Peak braking force was significantly higher (CG = +36 %, TG = +46 %; $p < 0.001$) and vertical GRF significantly smaller (CG = -3 %, TG = -2 %; $p < 0.05$) for both groups post-intervention compared to pre-intervention. To establish a significant influence of the intervention in TG for the kinetic variables, a minimum of N = 7 in TG is needed to determine the effect on peak braking GRF, N = 475 athletes for peak medial-lateral GRF and for peak vertical GRF a minimum of N = 239 athletes is needed in TG (Table 8.3-2).

All loading rates were significantly larger post-intervention compared to pre-intervention for both groups (medial-lateral: CG = + 194 %, TG = + 189 %; $p < 0.001$, anterior-posterior: CG = + 199 %, TG = + 145 %; $p < 0.001$, vertical: CG = + 162 %, TG = + 160 %; $p < 0.001$). In order to determine the effect of the intervention on loading rates, the sample size calculation showed that a minimum N = 14 in TG is needed to determine significance in the medial -lateral loading rate, N = 13 for the vertical loading rate, and N = 5 for the anterior-posterior loading rate (Table 8.3-2 and Figure 8.3-4).

TABLE 8. 3-2: The group mean (SD) pre-intervention to post-intervention comparison of performance and selected kinetic parameters between the control and training groups. The result of the power calculation indicating the minimum TG sample size to determine significance pre- to post-intervention is reported. (As calculated using the equation in (Noordzij et al., 2010)).

	Control group (n = 10)			Training group (n = 8)			2-way ANOVA			Minimum Sample size
	Pre	Post	%	Pre	Post	%	G	S	G*S	(n TG)
Performance										
Approach speed (m/s)	3.73 (0.33)	4.36 (0.28)	17%	3.77 (0.24)	4.29 (0.37)	14%	0.88	<.00#	0.56	5
Speed at iFS (m/s)	3.65 (0.34)	4.22 (0.26)	16%	3.70 (0.21)	4.22 (0.32)	14%	0.95	<.00#	0.79	4
Exit speed (m/s)	4.17 (0.30)	4.49 (0.36)	7%	4.33 (0.27)	4.74 (0.46)	9%	0.42	<0.05*	0.52	12
Stance Time (sec)	0.23 (0.02)	0.21 (0.02)	-10%	0.20 (0.02)	0.19 (0.01)	-5%	0.03*	<0.05*	0.38	35
Ground reaction force (BW)										
Peak medial-lateral	0.75 (0.12)	0.76 (0.13)	1%	0.85 (0.09)	0.87 (0.13)	3%	0.10	0.74	0.91	475
Peak braking	0.52 (0.22)	0.71 (0.20)	36%	0.46 (0.12)	0.67 (0.17)	46%	0.35	<.00#	0.37	7
Peak vertical	2.20 (0.23)	2.13 (0.17)	-3%	2.35 (0.20)	2.30 (0.19)	-2%	0.28	0.01*	0.63	239
Loading rate (BW/s)										
Medial-lateral	0.05 (0.04)	0.15 (0.07)	194%	0.05 (0.06)	0.14 (0.09)	189%	0.86	<.00#	0.82	14
Anterior-posterior	0.06 (0.04)	0.19 (0.07)	199%	0.07 (0.04)	0.17 (0.07)	145%	0.77	<.00#	0.80	5
Vertical	0.21 (0.17)	0.55 (0.29)	162%	0.21 (0.24)	0.54 (0.36)	160%	0.77	<.00#	0.87	13

*Notes:*p < .05, #p <. 001; iFS = initial foot strike; BW = body weight; n TG = number of athletes needed in training group; G = Group; S = Session; G*S = Group*Session interaction*



Note: GRFy = Anterior-posterior ground reaction force; TstGRFy = location of the transient braking GRFy; PeakGRFy = location of the peak propulsion GRFy; TG = Training group; CG = Control group; MetATA = Metatarsal Anterior Transverse Arch; MLA = Medial Longitudinal Arch

FIGURE 8. 3-2: The mean (bold) and SD (transparent) A) Foot-to-floor angle, B)medial-lateral GRF, C) anterior-posterior GRF, D) vertical GRF, E) metatarsal anterior transverse arch height and F)-length and G) medial longitudinal arch height and H) -length values throughout stance. The maximums for each variable were identified between TstGRFy and PeakGRFy (area not blacked out).

8.3.4 Kinematic outcomes

8.3.4.1 Foot-to-floor angle

8.3.4.1.1 Effect of intervention

No interaction effect between groups pre- compared to post-intervention was observed in this pilot study.

8.3.4.1.2 Group differences before intervention

The foot-to-floor angle was not significantly different between groups pre-intervention.

8.3.4.1.3 Sample size

There was a significant decline in foot-to-floor angle for both groups (CG = - 12 %, TG - 8 %; $p < 0.05$) post-intervention compared to pre-intervention. The power analysis showed that a minimum sample size of N = 141 athletes is needed to determine a significant intervention effect in the foot-to-floor angle for TG (Table 8.3-4).

8.3.4.2 Arches

8.3.4.2.1 Medial longitudinal arch height and length

8.3.4.2.1.1 Effect of intervention

Medial longitudinal arch height R^2 values were significantly different between groups, with higher R^2 values for TG compared to CG (CG: $R^2 = 0.38 \pm 0.14$, TG: $R^2 = 0.59 \pm 0.16$; $p < 0.05$) (Table 8.3-3 and Figure 8.3-2). TG showed a trend towards a larger maximum MLA length increase from pre- to post-intervention compared to the increase observed in CG ($p = 0.13$) (Table 8.3-4).

8.3.4.2.1.2 Group differences before intervention

The MLA height and length was similar between groups pre-intervention (Table 8.3-4).

8.3.4.2.1.3 Sample size

The power calculation indicated that N = 8 athletes per group is necessary to find a statistically significant R^2 value between groups (Table 8.3-3 and Figure 8.3-2).

The pilot study revealed that maximum MLA length for the average athlete increased significantly (CG = + 1 %, TG = + 2 %; $p < 0.05$) pre-intervention to post-intervention. To determine significant change due to intervention in maximum MLA height and length, a minimum sample sizes of N = 38 and N = 279 are needed in TG, respectively (Table 8.3-4).

8.3.4.2.2 Metatarsal anterior transverse arch height and length

8.3.4.2.2.1 Effect of intervention

R^2 values showed a higher MetATA height value for TG compared to CG and trending towards significant difference between groups (CG: $R^2 = 0.71 \pm 0.21$, TG: $R^2 = 0.89 \pm 0.27$; $p = 0.08$) (Table 8.3-3). TG had a trend towards a larger increase in maximum metatarsal anterior transverse arch height (CG = +1 %, TG = +14 %; $p = 0.09$) and a larger decrease in maximum MetATA length (CG = - 5 %, TG = - 9 %; $p = 0.12$) post-intervention compared to pre-intervention than CG (Table 8.3-4).

8.3.4.2.2.2 Group differences before intervention

CG and TG had similar maximum MetATA height and length values pre-intervention (Table 8.3-4).

8.3.4.2.2.3 Sample size

Power analysis showed that N = 28 and N = 2 athletes in each group is necessary to find significance for metatarsal anterior transverse arch length and height R^2 values respectively (Table 8.3-3 and Figure 8.3-2).

A significant increase in maximum MetATA height (CG = + 1 %, TG = + 14 %; $p < 0.05$) and decrease in length (CG = - 5 %, TG = - 9 %; $p < 0.001$) for the average athlete was observed post-intervention compared to pre-intervention. The minimum number of athletes necessary in TG to determine intervention significance for maximum MetATA height and length was determined to be N = 132 for metatarsal anterior transverse arch height and N = 108 for MetATA length (Table 8.3-4).

8.3.4.3 Foot segment angles

8.3.4.3.1 Sagittal plane Hallux-Metatarsal angles

8.3.4.3.1.1 Effect of intervention

A weak trend towards smaller sagittal plane hallux-metatarsal angle R^2-values was observed for the TG compared to the CG (CG = 0.82 ± 0.24, TG = 0.69 ± 0.32; $p = 0.23$) (Table 8.3-3 and Figure 8.3-3) . No significant difference in the decrease in the minimum hallux-metatarsal extension angle between groups post-intervention compared to pre-intervention was observed (Table 8.3-4).

8.3.4.3.1.2 Group differences before intervention

There was a trend towards a significant difference in the minimum hallux-metatarsal angle between groups pre-intervention (CG = 36.30 ± 4.63°, TG = 48.94 ± 11.92°; $p = 0.10$) (Table 8.3-4).

8.3.4.3.1.3 Sample size

To establish the influence of the intervention on the R^2 values of the sagittal plane movement of the hallux-metatarsal segment N = 73 is needed in both TG and CG (Table 8.3-3 and Figure 8.3-3). All athletes displayed a significant decrease in minimum hallux-metatarsal extension angle post-intervention compared to pre-intervention (CG = - 17 %, TG = - 35 %; $p < 0.05$). A minimum sample size of in TG of

N = 7 is necessary to find a significant change in the minimum hallux-metatarsal extension angle (Table 8.3-4).

8.3.4.3.2 METATARSAL-MIDFOOT ANGLES

8.3.4.3.2.1 EFFECT OF INTERVENTION

TG displayed a trend towards lower R^2 values for metatarsal-midfoot extension/flexion angles compared to CG (CG = 0.80 ± 0.18, TG = 0.62 ± 0.30; p = 0.10) (Table 8.3-3 and Figure 8.3-3). For minimum/maximum angles, pre- to post-intervention decrease in the maximum metatarsal-midfoot flexion and -eversion angles was significantly larger for TG compared to CG (maximum metatarsal-midfoot flexion: CG = - 1 %, TG = - 9 %; $p < 0.05$, maximum metatarsal-midfoot eversion: CG = - 5 %, TG = - 24 %; $p < 0.05$) (Table 8.3-4).

8.3.4.3.2.2 GROUP DIFFERENCES BEFORE INTERVENTION

CG and TG had similar maximum and minimum (where applicable) metatarsal-midfoot angles in all planes pre-intervention (Table 8.3-4).

8.3.4.3.2.3 SAMPLE SIZE

A minimum sample size of N = 28 athletes per group is needed to find significance between the R^2 values of each group when comparing metatarsal-midfoot extension/flexion angles (Table 8.3-4 and Figure 8.3-3). The maximum metatarsal-midfoot flexion (CG = - 1 %, TG - 9 %; $p < 0.05$) and maximum metatarsal-midfoot eversion (CG = - 5 %, TG = - 24 %; $p < 0.05$) angles were significantly different for both groups post-intervention compared to pre-intervention (Table 8.3-3 and Figure 8.3-3). To establish significance of the intervention in future studies for changes to the maximum metatarsal-midfoot flexion angle a minimum N = 14 is needed in TG and for the maximum eversion angle N = 12 (Table 8.3-4).

8.3.4.3.3 *Midfoot-calcaneus angles*

8.3.4.3.3.1 Effect of intervention

TG had significantly lower transverse plane midfoot-calcaneus angle R^2 values compared to CG (CG = 0.72 ± 0.20, TG = 0.35 ± 0.15; $p < 0.05$). The sagittal and frontal plane midfoot-calcaneus angle R^2 values displayed a trend towards significantly larger values for TG compared to CG (Sagittal plane: CG = 0.85 ± 0.10, TG = 0.88 ± 0.11; p = 0.18; Transverse plane: CG = 0.65 ± 0.24, TG = 0.81 ± 0.15; p = 0.21) (Table 8.3-3 and Figure 8.3-3). TG pre-to post-intervention decrease in maximum midfoot-calcaneus flexion angle was significantly different compared to the pre- to post-intervention decrease observed for CG (CG = - 2 %, TG = - 18 %; $p < 0.05$) (Table 8.3-4).

8.3.4.3.3.2 Group differences before intervention

Pre-intervention TG had a significantly larger maximum midfoot-calcaneus eversion angle compared to CG (CG = - 12.78 ± 5.30°, TG = - 17.91 ± 5.40°; $p < 0.05$). There was also a trend towards significant difference between groups in the transverse midfoot-calcaneus angle pre-intervention. CG presented with a midfoot-calcaneus adduction angle in contrast to the midfoot-calcaneus abduction angle of TG (CG = 2.33 ± 3.80°, TG = - 3.30 ± 4.34°; p = 0.05) (Table 8.3-4).

8.3.4.3.3.3 Sample size

In the sagittal plane midfoot-calcaneus angles a minimum N = 192, frontal plane N = 23, and in the transverse plane N = 4 athletes per group is needed to find R^2 significance (Table 8.3-4 and Figure 8.3-3).

Post-intervention, there was a significant decrease in the maximum midfoot-calcaneus extension angle for both groups compared to pre-intervention (CG = - 2 %, TG = - 18 %; $p < 0.05$). There was also a significant difference in the decrease of the maximum midfoot-calcaneus flexion angle between groups post-intervention compared to pre-intervention ($p < 0.05$). Both groups also displayed a significant increase in the maximum midfoot-calcaneus adduction angle from pre- to post-intervention (CG = 14 %, TG = 118 %; $p < 0.05$). Sample size calculation showed that TG needed a minimum N = 10 athletes in future studies to determine significance in the maximum sagittal plane midfoot-calcaneus angles and N = 41 athletes in the minimum transverse plane angle (Table 8.3-4).

8.3.4.3.4 Calcaneus-shank angles

8.3.4.3.4.1 Effect of intervention

TG had a trend towards larger sagittal plane calcaneus-shank R^2 values pre- compared to post-intervention (CG = 0.73 ± 0.19, TG = 0.84 ± 0.13; $p = 0.07$) (Table 8.3-3 and Figure 8.3-3). A trend towards significantly smaller increase in the maximum calcaneus-shank eversion for TG pre- to post-intervention was observed compared to the pre- to post-intervention increases for CG (CG = + 34 %, TG = + 13 %; $p = 0.11$).

8.3.4.3.4.2 Group differences before intervention

The calcaneus-shank angles in all three planes were similar between groups pre-intervention.

8.3.4.3.4.3 Sample size

Power calculation showed that a minimum sample size N = 31 athletes per group is needed to find significance for sagittal plane calcaneus-shank R^2 value adaptations. Post-intervention, a decrease for CG and an increase for TG in the maximum calcaneus-shank dorsiflexion angle was observed, displaying a trend towards a significant change compared to pre-intervention angles (CG = - 15 %, TG = + 26 %; $p = 0.06$). Both groups also presented with an increase in the maximum calcaneus-shank eversion and -abduction angles post-intervention compared to pre-intervention (Frontal plane: CG = + 34 %, TG = + 13 %; $p < 0.05$, Transverse plane: CG = + 53 %, TG = + 15 %; $p < 0.05$). (Table 8.3-3 and Figure 8.3-3). The sample size calculation showed that in future studies, N = 18 athletes are needed to complete the intervention in TG to establish the significance of the intervention on sagittal plane calcaneus-shank angles, N = 67 athletes in TG to determine significance for frontal plane and N = 53 athletes for transverse plane calcaneus-shank angles (Table 8.3-4 and Figure 8.3-3).

TABLE 8. 3-3: The group mean (SD) pre-intervention to post-intervention comparison of foot arch and -segment variables between CG and TG. The result of the power calculation indicating the minimum TG sample size to determine significance pre- to post-intervention is reported. (As calculated using the equation in (Noordzij et al., 2010)).

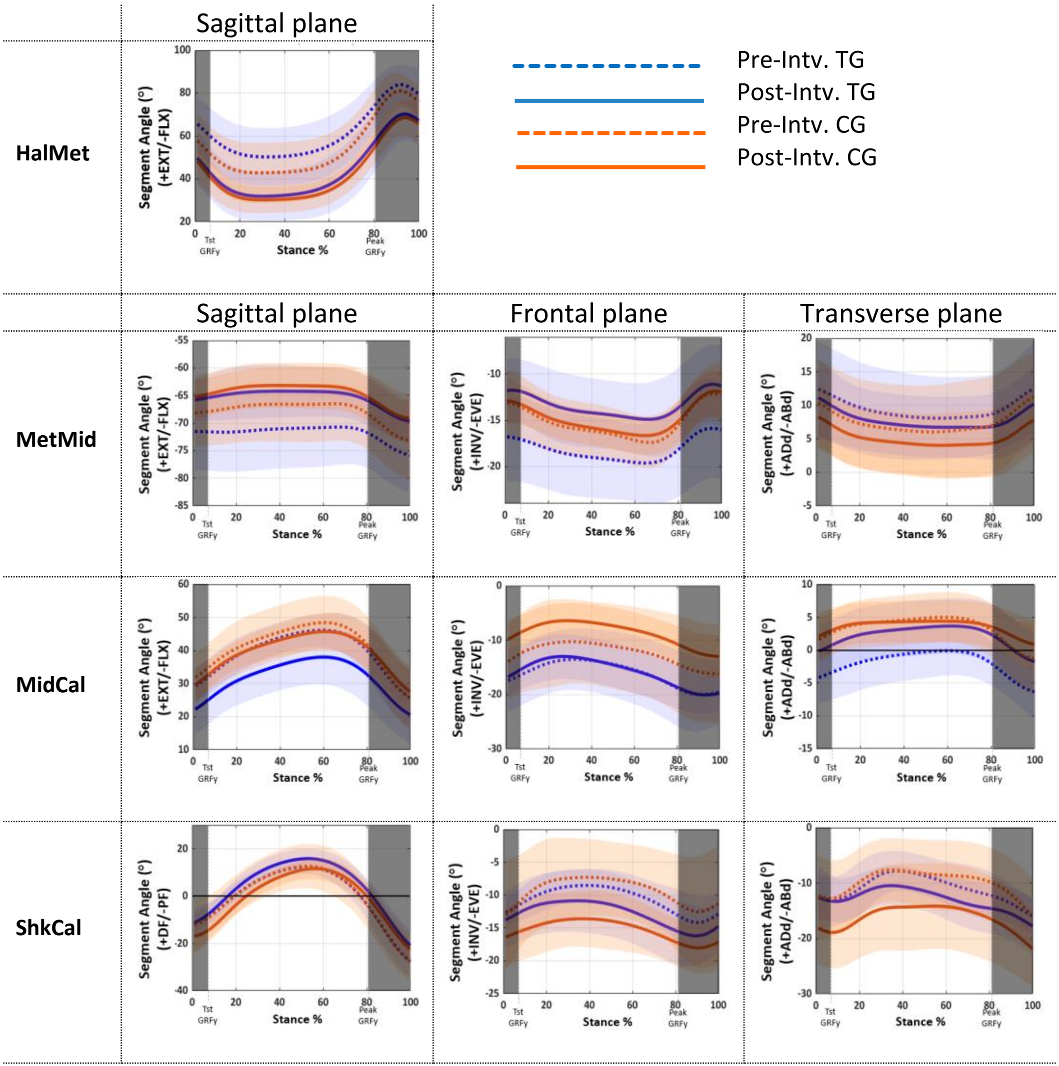
	Control group (n = 10)			**Training group (n = 8)**			**2-way ANOVA**			**Minimum Sample size**
	Pre	**Post**	**%**	**Pre**	**Post**	**%**	**G**	**S**	**G*S**	**(n TG)**
Foot-to-Floor (°)	-17.39 (3.34)	-15.34 (2.46)	-12%	-16.74 (5.09)	-15.33 (3.36)	-8%	0.55	0.04*	0.46	141
Arches maximum height and length (cm)										
MLA Height (cm)	0.89 (0.00)	0.89 (0.01)	-0.1%	0.89 (0.01)	0.89 (0.01)	0.1%	0.35	0.76	0.31	38
MLA Length (cm)	0.20 (0.01)	0.20 (0.01)	1%	0.20 (0.01)	0.20 (0.01)	2%	0.56	0.01*	0.13	279
MetATA Height (cm)	0.05 (0.01)	0.05 (0.00)	1%	0.05 (0.00)	0.06 (0.01)	14%	0.54	<0.05*	0.09	132
MetATA Length (cm)	0.09 (0.00)	0.09 (0.01)	-5%	0.10 (0.01)	0.09 (0.01)	-9%	0.78	<.00#	0.12	108
Segmental angle maximums (°)										
Min. HalMet (+EXT/-FLX)	36.30 (4.63)	30.07 (6.17)	-17%	48.94 (11.92)	31.78 (11.31)	-35%	0.10	<0.05*	0.24	7
Max. MetMid (+EXT/-FLX)	-64.29 (2.98)	-63.78 (2.57)	-1%	-69.93 (6.78)	-63.84 (4.68)	-9%	0.38	0.01*	0.02*	14
Min. MetMid (+INV/-EVE)	-17.81 (2.34)	-16.91 (2.00)	-5%	-20.08 (4.54)	-15.27 (4.00)	-24%	0.64	0.02*	0.03*	12
Min. MetMid (+ADd/-ABd)	5.81 (7.02)	3.83 (4.84)	-34%	7.66 (6.25)	6.12 (4.84)	-20%	0.78	0.39	0.86	204
Max. MidCal (+EXT/-FLX)	46.73 (6.16)	45.90 (5.49)	-2%	46.80 (5.42)	38.23 (8.21)	-18%	0.25	0.03*	0.04*	10
Min. MidCal (+INV/-EVE)	-12.78 (5.30)	-11.03 (3.06)	-14%	-17.91 (5.40)	-18.68 (3.87)	4%	0.03*	0.45	0.23	569
Min. MidCal (+ADd/-ABd)	2.33 (3.80)	2.66 (3.37)	14%	-3.30 (4.34)	0.60 (4.36)	-118%	0.05	0.03*	0.22	41
Max. CalShk (+DF/-PF)	14.19 (3.33)	12.08 (4.33)	-15%	12.79 (2.85)	16.08 (4.29)	26%	0.20	0.06	0.25	18
Min. CalShk (+INV/-EVE)	-13.66 (3.13)	-18.30 (3.49)	34%	-13.78 (2.85)	-15.63 (4.77)	13%	0.37	0.01*	0.11	67
Min. CalShk (+ADd/-ABd)	-12.74 (6.62)	-19.47 (6.77)	53%	-14.19 (3.50)	-16.24 (4.04)	15%	0.26	<0.05*	0.28	53

*Notes: *p < .05, #p <. 0001; MLA = medial longitudinal arch, MetATA = metatarsal anterior transverse arch; HalMet = Hallux-Metatarsal; MetMid = Metatarsal-Midfoot; MidCal = Midfoot-Calcaneus; CalShk = Calcaneus-Shank; +EXT = Extension; -FLX = Flexion; +INV = Inversion; -EVE = Eversion; +ADd = Adduction; -ABd = Abduction; n TG = number of athletes needed in training group; G = Group; S = Session; G*S = Group*Session interaction*

TABLE 8. 3-4: The pre-intervention to post-intervention comparison of the R^2 values within groups. The minimum sample size per group needed to determine significance is displayed (Noordzij et al., 2010).

	Control group (n = 10)	**Training group (n = 8)**	**p**	**Sample size / group**
Foot-to-Floor	0.99 (0.01)	0.98 (0.02)	0.66	35
Arch height and length				
MLA Height	0.38 (0.14)	0.59 (0.16)	<0.05*	8
MLA Length	0.95 (0.03)	0.85 (0.25)	0.72	31
MetATA Height	0.71 (0.21)	0.89 (0.27)	0.08	28
MetATA Length	0.92 (0.05)	0.62 (0.16)	0.18	2
Segmental angles				
HalMet (EXT/FLX)	0.82 (0.24)	0.69 (0.32)	0.23	73
MetMid (EXT/FLX)	0.80 (0.18)	0.62 (0.30)	0.10	28
MetMid (INV/EVE)	0.80 (0.16)	0.68 (0.25)	0.43	46
MetMid (ADd/ABd)	0.49 (0.28)	0.44 (0.28)	0.50	492
MidCal (EXT/FLX)	0.85 (0.10)	0.88 (0.11)	0.18	192
MidCal (INV/EVE)	0.65 (0.24)	0.81 (0.15)	0.21	23
MidCal (ADd/ABd)	0.72 (0.20)	0.35 (0.15)	<0.05*	4
CalShk (DF/PF)	0.73 (0.19)	0.84 (0.13)	0.07	31
CalShk (INV/EVE)	0.63 (0.27)	0.63 (0.24)	0.68	70 086
CalShk (ADd/ABd)	0.67 (0.15)	0.64 (0.19)	0.65	504

*Notes: *p < .05, #p <. 0001; MLA = medial longitudinal arch, MetATA = metatarsal anterior transverse arch; HalMet = Hallux-Metatarsal; MetMid = Metatarsal-Midfoot; MidCal = Midfoot-Calcaneus; CalShk = Calcaneus-Shank; +EXT = Extension; -FLX = Flexion; +INV = Inversion; -EVE = Eversion; +ADd = Adduction; -ABd = Abduction; n TG = number of athletes needed in training group; G = Group; S = Session; G*S = Group*Session interaction*



Notes: GRFy = Anterior-posterior ground reaction force; TstGRFy = location of the transient braking GRFy; PeakGRFy = location of the peak propulsion GRFy; TG = Training group; CG = Control group; HalMet = Hallux-Metatarsal; MetMid = Metatarsal-Midfoot; MidCal = Midfoot-Calcaneus; CalShk = Calcaneus-Shank

FIGURE 8. 3-3: The mean (bold) and standard deviation (shaded areas) of the segment joint angle curves for each group at pre- and post-intervention. The maximums for each variable were identified between TstGRFy and PeakGRFy (area not blacked out).

8.3.4.4 LAS INJURY RISK FACTORS

8.3.4.4.1 EFFECT OF INTERVENTION

A weak trend towards a significantly smaller increase from pre- to post-intervention in the maximum ankle inversion angle for TG compared to CG was observed (CG = + 10 %, TG = + 4 %; $p = 0.11$). Similarly, TG displayed weak trend towards a significantly different adaptation (larger decrease) in the ankle eversion moment arm length compared to the change observed for the CG post- compared to pre-intervention (CG = - 6 %, TG = - 35 %; $p = 0.13$) (Table 8.3-5).

8.3.4.4.2 GROUP DIFFERENCES BEFORE INTERVENTION

TG and CG displayed no significant pre-intervention variations in any variable associated with ACLR injury risk (Table 8.3-5).

8.3.4.4.3 SAMPLE SIZE

Maximum frontal and transverse plane ankle angles, and maximum ankle inversion velocity was significantly larger post- compared to pre-intervention for both groups (maximum frontal plane ankle angles: $p < 0.05$; maximum transverse plane ankle angles: $p < 0.001$; maximum inversion velocity: $p < 0.05$;). The power calculation showed that to determine the effect of the intervention N = 140 is needed in the TG to determine a significant adaptation in maximum frontal plane ankle angles, N = 32 for maximum transverse plane ankle angles, and N = 47 for maximum ankle inversion velocity (Table 8.3-5).

8.3.4.5 ACL INJURY RISK FACTORS

8.3.4.5.1 EFFECT OF INTERVENTION

TG displayed a trend towards a significant adaptation pre- to post-intervention in the maximum frontal plane knee angle compared to the adaptation observed for the CG in the same period (CG = + 195 %, TG = - 33 %; $p = 0.20$) (Table 8.3-5).

8.3.4.5.2 *GROUP DIFFERENCES BEFORE INTERVENTION*

Maximum knee frontal plane angles had a trend towards significant different pre-intervention angles between groups with CG presenting with a varus angle and TG with a valgus angle (CG = 1.22 ± 3.77°, CG = - 4.21 ± 4.75°; p = 0.07). Pre-intervention maximum frontal plane moment arm lengths also displayed a trend towards significant difference between groups (CG = - 2.73 ± 1.62 cm, TG = - 2.90 ± 2.15 cm; p = 0.07) (Table 8.3-5).

8.3.4.5.3 *SAMPLE SIZE*

Both groups had a significantly smaller maximum knee flexion angle post-intervention compared to pre-intervention, with a minimum of N = 21 participants needed in TG to determine significance (CG = - 22 %, CG = - 19 %; p < 0.001). The maximum frontal plane knee moment arm length was also significantly different for both groups post- compared to pre-intervention (CG = + 71 %, CG = + 24 %; p < 0.05). To find a significant change in the maximum frontal plane knee moment arm length N = 114 participants is needed in TG (Table 8.3-5).

TABLE 8. 3-5: The group mean (SD) pre-intervention to post-intervention comparison of selected ACLR and LAS injury risk variables between CG and TG. The result of the power calculation indicating the minimum TG sample size to determine significance pre- to post-intervention is reported. (As calculated using the equation in (Noordzij et al., 2010)).

	Control group (n = 10)			Training group (n = 8)			2-way ANOVA			Minimum Sample size
	Pre	Post	%	Pre	Post	%	G	S	G*S	(n TG)
Ankle injury risk variables maximums										
+INV/-EVE angle (°)	31.80 (2.86)	34.98 (3.56)	10%	32.82 (4.02)	34.12 (3.75)	4%	0.85	0.02*	0.11	140
+ADd/-ABd angle (°)	-1.03 (6.01)	-5.31 (6.99)	418%	-4.25 (4.21)	-6.94 (3.44)	63%	0.96	<.00#	0.45	32
+INV/-EVE velocity (°/s)	130.50 (38.56)	161.31 (59.76)	24%	137.75 (48.45)	164.68 (44.49)	20%	0.94	0.02*	0.38	47
+ADd/-ABd velocity (°/s)	272.74 (155.25)	208.95 (95.12)	-23%	215.67 (119.56)	202.59 (80.06)	-6%	0.75	0.38	0.26	914
+INV moment arm (cm)	1.74 (0.53)	1.86 (0.52)	7%	1.46 (0.44)	1.60 (0.43)	10%	0.30	0.32	0.86	152
-EVE moment arm (cm)	0.79 (0.72)	0.75 (0.53)	-6%	0.69 (0.58)	0.45 (0.52)	-35%	0.62	0.18	0.13	82
Knee injury risk variables maximums										
+VAR/-VAL angle (°)	1.22 (3.77)	-1.15 (4.66)	195%	-4.21 (4.75)	-2.80 (3.38)	-33%	0.07	0.81	0.20	131
-EXT/+FLX angle (°)	33.86 (4.70)	26.54 (5.56)	-22%	34.99 (9.24)	28.23 (6.50)	-19%	0.65	<.00#	0.96	21
+IR/-ER velocity (°/s)	456.00 (204.06)	389.35 (103.37)	-15%	407.16 (255.36)	318.38 (152.94)	-22%	0.44	0.33	0.86	83
+VAR/-VAL moment arm (cm)	-2.73 (1.62)	-4.67 (1.46)	71%	-2.90 (2.15)	-3.60 (1.63)	24%	0.07	<0.05*	0.50	114

*Notes:*p < .05, #p <. 001; +EXT = Extension; -FLX = Flexion; +INV = Inversion; -EVE = Eversion; +ADd = Adduction; -ABd = Abduction; +VAR = Varus; -VAL = Valgus; +IR = Internal Rotation; -ER = External Rotation; n TG = number of athletes needed in training group; G = Group; S = Session; G*S = Group*Session interaction*

8.4. DISCUSSION

The aim of this pilot study is to provide proof of concept and guidance for future, large scale randomised controlled intervention studies investigating the influence training the muscles acting on the foot has on segmental movement of the foot, ACLR and LAS injury risk factors, as well as selected performance variables associated with unanticipated change of direction tasks. The proof-of-concept study aims to identify the variables most likely to influence the response to the intervention and provides sample size guidelines for future randomised controlled trials.

8.4.1 PROOF-OF-CONCEPT - BIOMECHANICAL ADAPTATIONS OBSERVED

The principle aim of the pilot study was to provide proof of concept that training the foot musculature has a significant influence on the biomechanical variables associated with segmental foot movement and/or ACL and LAS injury risk factors. Indeed, post-intervention TG displayed a trend towards smaller maximum knee valgus angles, smaller increases in the maximum ankle inversion angle and larger decreases in the maximum ankle eversion moment arm, compared to CG. The trend towards the smaller maximum knee valgus angle observed in TG may be the outcome of the significantly greater medial longitudinal arch height for TG, compared to CG. Post-intervention differences in R^2 values between groups in the sagittal plane metatarsal-midfoot (TG had smaller sagittal plane metatarsal-midfoot angles) and midfoot-calcaneus (TG had smaller transverse plane midfoot-calcaneus, larger sagittal plane midfoot-calcaneus and frontal plane midfoot-calcaneus angles), as well as the sagittal plane calcaneus-shank R^2 values (TG had larger sagittal plane calcaneus-shank angles) were also observed. These changes in the movement of the foot segments may be an indication that TG were able to resist dynamic pronation more effectively than CG throughout stance under the larger kinetic measures observed. The increased speed during post-intervention testing has likely influenced the higher peak loading rates as well as greater braking and vertical GRF of both groups post-intervention compared to pre-intervention (Frederick & Hagy, 1986). TG was thus able to resist dynamic foot deformation more effectively than CG under the higher loads experienced by both groups post-intervention. Resistance to decreased MLA height and pronation in TG may have had an influence on the smaller knee valgus angle observed in TG, as excessive dynamic rearfoot pronation has previously been linked to increase knee valgus angles (Holland, 2016; Kagaya et al., 2013; McLean et al., 2004). Thus, it may be worthwhile to investigate whether training the muscles acting on the foot are able to influence maximum knee valgus angle by increasing the resistance to MLA deformation.

Similarly, group differences in foot segmental movement outcomes may have influenced the group differences to LAS injury risk as mentioned above. Post-intervention, TG displayed a trend towards larger MetATA height, smaller MetATA length and lower sagittal plane hallux-metatarsal R^2 values compared to CG. The adaptations to the sagittal plane hallux-metatarsal, and height and length R^2 value of the MetATA can be interpreted as an increase in forefoot stiffness (Duerinck et al., 2014; Venkadesan et al., 2020). Furthermore, increased hallux and metatarsal flexion (as reported in chapter seven) has also been linked to increased foot stiffness (Farris et al., 2020). A stiffer foot with a higher MLA allows for a larger moment arm for the fibularis group to affect a larger internal eversion moment around the ankle joint, decreasing LAS risk (Glasoe, Yack, & Saltzman, 1999).Indeed, previous research has found that a lateral shift in the centre of pressure, increased the moment arms around the ankle joint resulting in a LAS (Fong, Hong, et al., 2009). Thus, the stiffer forefoot may have acted to shift the centre of pressure more medially providing a stable base which may explain the smaller inversion angles, as well as smaller eversion moment arms observed in TG post-intervention. The smaller inversion angle and smaller eversion moment arm will put the resultant GRF vector closer to the ankle joint centre in this way possibly decreasing LAS injury risk.

8.4.2 Group differences before intervention

The intervention protocol seemed to affect some adaptation in both dynamic movement of the foot segments and ACL and LAS injury risk factors, validating future randomised controlled trials. However, there were some pre-intervention discrepancies between groups. To limit the number of factors that could influence the athletes' response to the initial intervention the athletes were matched for sport and BMI. BMI is thought to play a role in foot function (Butterworth, Landorf, Gilleard, Urquhart, & Menz, 2014) as well as ACLR and LAS injury risk (Gould, Hooper, & Strauss, 2016; McCriskin et al., 2015). The pre-intervention group interaction results provide further guidance as to factors that could influence group differences before intervention. The pilot study revealed significantly shorter pre-intervention stance times and a trend towards larger peak medial-lateral GRF for TG compared to CG. From previous research, it is unclear whether shorter stance times and larger medio-lateral GRF are an indication of superior foot function (Spiteri et al., 2015) and thus whether pre-intervention GRF and stance times would influence intervention results. However, in previous studies, athletes with stronger lower bodies and superior neuromuscular control, which includes ankle biomechanics and therefore conceivably foot function, were associated with larger peak GRF and shorter stance times (Fox, 2018; McLean et al., 2004). GRF and stance times should thus be used to guide matching groups before randomisation.

In the current study, significant differences, or trends towards significant differences, in pre-intervention foot segment angles between groups were observed. TG had a trend towards larger maximum hallux-metatarsal extension angle, which is related to the windlass mechanism and medial longitudinal arch function (Huson, 1991; Farris et al., 2020). TG also had significantly larger maximum midfoot-calcaneus eversion and a trend towards larger midfoot-calcaneus abduction angles. Midfoot-calcaneus movement is related to subtalar joint pronation and has been linked to intrinsic and extrinsic foot muscle function (Jastifer & Gustafson, 2014; Kelly et al., 2014; Neptune, Wright, & Van Den Bogert, 1999). Thus, pre-intervention differences in midfoot-calcaneus function could influence the adaptation to a foot muscle specific intervention. Furthermore, there were also significant pre-intervention differences in the maximum knee valgus angles and the length of the maximum knee valgus moment arm which will affect intervention results. It will thus be prudent to use values from pre-intervention stance time, peak medial-lateral GRF, minimum sagittal plane hallux-metatarsal, frontal- and transverse plane midfoot-calcaneus angles as well as maximum knee valgus angle and knee valgus moment arm length values to guide matched-randomization of the groups (Table 8.4-1).

TABLE 8. 4-1: Variables to consider that could influence the effect of the intervention.

	CG (n =10)	TG (n =8)	
	Pre-Intv.	**Pre-Intv.**	**p-value**
Sport	Na	Na	Na
BMI	23.03 (3.33)	23.73 (1.98)	1.00
Stance time	0.23 (0.02)	0.20 (0.02)	0.03*
Peak medial-lateral GRF	0.75 (0.12)	0.85 (0.09)	0.10
Min HalMet (+EXT/-FLX)	36.30 (4.63)	48.94 (11.92)	0.10
Min. MidCal (+INV/-EVE)	-12.78 (5.30)	-17.91 (5.40)	0.03*
Min. MidCal (+ADd/-ABd)	2.33 (3.80)	-3.30 (4.34)	0.05
Knee +VAR/-VAL angle (°)	1.22 (3.77)	-4.21 (4.75)	0.07
Knee +VAR/-VAL moment arm (cm)	-2.73 (1.62)	-2.90 (2.15)	0.07

*Notes: *p < .05, #p <. 0001; BMI = body Mass Index; GRF = Ground reaction Force; HalMet = Hallux-Metatarsal; MidCal = Midfoot-Calcaneus; +EXT = Extension; -FLX = Flexion; +INV = Inversion; -EVE = Eversion; +ADd = Adduction; -ABd = Abduction; +VAR = Varus; -VAL = Valgus; Na = Not applicable*

8.4.3 Methodology recommendations

The outcomes of the pilot study were influenced by shortcomings in the methodology that should be improved upon in future intervention studies. The main weaknesses involved discrepancy in approach speed during data capture, compliance to the intervention and low participation numbers. In this study the approach speed was significantly higher for both groups post-intervention compared to pre-intervention. The inconsistency in approach speed also denotes that anticipation between athletes were dissimilar. The signal to change direction was given 0.7 seconds after the approach run was initiated. Discrepancy in approach speed thus results in athletes with a slow approach speed having more time to respond to the stimulus and faster runners having less time to respond to the stimulus. Previous research has shown that knee moments were significantly lower in anticipated than unanticipated change of direction tasks (Besier, Lloyd, Ackland, & Cochrane, 2001). Thus, inconsistency between approach speeds could have conceivably obscured the effect of the intervention in this study and should be better monitored in future studies. Ideally all independent variables should be similar for both groups at both pre- and post-intervention evaluations as to not confound intervention effects.

Poor compliance rates possibly further influenced the effect of the intervention. The compliance to supervised sessions was similar to other research (Mickle et al., 2016). However, compliance to unsupervised sessions in this study was less than supervised sessions and inferior to what was reported in other home-based foot strengthening intervention studies (Mickle et al., 2016; Mulligan & Cook, 2013). Poor compliance could possibly obscure the effectiveness of the intervention. Future studies may have higher compliance rates when the intervention programme is linked to or integrated into existing warm-up and/or conditioning programmes.

The effect of the intervention on neuromuscular adaptation was not tested in this study and is a potential gap in the current research. Future studies should aim to determine the adaptation to muscle activity, muscle strength and/or physiological changes to muscle. Monitoring muscle activity during the execution of exercises for the duration of the intervention would also ensure the exercise is activating the intended muscle(s). The aforementioned methods were not included in the current study as fine wire electromyography in deep intrinsic foot muscles during high-intensity change of direction tasks is still problematic with current technology. Furthermore, specialist equipment and individually monitored sessions was logistically unattainable.

8.4.4 SAMPLE SIZE RECOMMENDATIONS

Another key aim of the study was to provide recommendations regarding the minimum sample size needed to determine effectiveness of the intervention. To determine adaptation to dynamic foot function only those variables displaying significantly different or a trend towards significantly different R^2 values was considered. The goodness-of-fit measure compares both the temporal and magnitude of the change between pre- and post-intervention in a time series. The R^2 values thus providing information about the effectiveness of the intervention over the whole stance phase opposed to a snapshot that is provided with a maximum or minimum value. In contrast, the maximum values of the variables associated with ACLR and LAS risk was used as the maximum or extreme ranges of motion is relevant in injury risk. Thus, variables associated with ACLR and LAS risk that presented with a significant or a trend towards a significant G*S interaction in the 2-way ANOVA was identified.

As noted in the biomechanical adaptation sections of the discussion, the R^2 values of MLA height and transverse plane midfoot-calcaneus displayed were significantly different between groups. The variables MetATA height and length, sagittal plane hallux-metatarsal, metatarsal-midfoot, midfoot-calcaneus and calcaneus-shank angles, and frontal midfoot-calcaneus had R^2 values trending towards significant differences between groups. For ACL and LAS injury risk variables there was a trend towards significant differences between groups pre- compared to post-intervention outcomes in frontal plane ankle angles and ankle moment arm length. The frontal plane knee angle also displayed a trend towards significant differences between groups pre- compared to post-intervention. Biomechanical research studies recommend a minimum number of 30 athletes per group when exploring the significant influence of an intervention between groups (Mullineaux & Wheat, 2017). Thus, MLA height and MetATA height and length, sagittal plane metatarsal-midfoot and calcaneus-shank angles, frontal- and transverse plane midfoot-calcaneus angles are recommended to include in future studies of similar design. The number of athletes needed to determine significance for ACLR, and LAS injury variables are not attainable and thus not practical to pursue in similar future studies (Table 8.4-2).

TABLE 8. 4-2: Variables influenced by the intervention with minimum athlete numbers per group needed to determine significance.

Foot function (Time series)			**Injury risk factor (maximum)**		
	R^2 p-value	**n/ group**		**G*S p-value**	**nTG**
Arches			**Ankle injury risk outcomes**		
MLA height	<0.05*	8	+INV/-EVE angle	0.11	140
MetATA Height (cm)	0.08	28	-EVE moment arm	0.13	82
MetATA Length (cm)	0.18	2	**Knee injury risk outcomes**		
Segmental angles			+VAR/-VAL angle	0.20	131
HalMet (+EXT/-FLX)	0.23	73			
MetMid (+EXT/-FLX)	0.10	28			
MidCal (+EXT/-FLX)	0.18	192			
MidCal (+INV/-EVE)	0.21	23			
MidCal (+ADd/-ABd)	<0.05*	4			
CalShk (+DF/-PF)	0.07	31			

8.5. CONCLUSION

Training the foot musculature increased the resistance to deformation of both the MLA and the MetATA, increasing mid- and forefoot stiffness. Foot function has previously been linked to ACLR and LAS risk factors and may therefore be responsible for the decreases observed in some variables (maximum ankle inversion angle, maximum ankle eversion moment arm length, and maximum knee valgus angle) that are associated with aforementioned risk. The power analysis showed that a realistic number of athletes is needed to find significant differences in foot arch and foot segment angle variables, making future foot muscle specific randomised controlled intervention studies feasible. In contrast, a large number of athletes are necessary to find significant differences in injury risk variables.

Compliance to this randomised controlled proof of concept pilot study was poor in relation to other, randomised intervention studies of a similar design (Mickle et al., 2016; Mulligan & Cook, 2013). The poor compliance rates may have influenced the results reported here. The effect that training the muscles of the foot has on ACLR and LAS injury risk factors are thus inconclusive on the account of poor compliance rates. However, future foot muscle specific intervention studies are warranted. Future studies may be more successful in establishing the effectiveness of foot muscle specific training on ACLR and LAS risk factors if a minimum number of 30 athletes per group is recruited, the groups are effectively randomised pre-intervention, approach speed is accurately controlled, and the intervention programme is linked to current warm-up and/or strength and conditioning programmes to increase compliance.

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CHAPTER 9

DISCUSSION

9.1. GENERAL DISCUSSION

High-intensity, unanticipated decelerations, characteristic of court sports, increases the risk of sustaining a ACLR and/or LAS. A body of literature exist suggesting that in closed chain activities, such as running, the foot plays an important role in lower limb biomechanics and thus potentially injury prevention in sports (Baltich, Emery, Stefanyshyn, & Nigg, 2014; Nigg, Baltich, Federolf, Manz, & Nigg, 2017). Indeed, impaired dynamic foot function is linked to an increase in the risk of sustaining an injury in dynamic sport activities (Fong, Hong, et al., 2009; Holland, 2016; Kagaya, Fujii, & Nishizono, 2015; Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005; Willems, Witvrouw, Delbaere, Philippaerts, et al., 2005). However, no current injury prevention programme aims to change the function of the foot directly. The body of this work intended to determine whether a foot muscle specific intervention programme is feasible and able to supplement current ACLR and LAS prophylactic programmes.

Functional anatomy of the foot and lower limb was explored in **Chapter 2**. Passive and active structures of the foot allow the individual segments to perform unique functions and/or acting as a unit to absorb ground reaction forces (GRF) and/or transfer rotational forces to and from the shank. **Chapter 4** investigated the current literature on known injury risk factors associated with ACLR and LAS. The review of the literature revealed that internal tibial rotation and large knee valgus angles, both of which increase ACLR risk, are linked to excessive subtalar joint pronation and rearfoot eversion (Beckett, Massie, Bowers, & Stoll, 1992; Gamada, 2014; Shultz et al., 2015; Stuelcken, Mellifont, Gorman, & Sayers, 2015). The movement of the centre of pressure towards the lateral side of the forefoot and a larger hallux extension range of motion is linked to increased LAS risk (Kristianslund, Bahr, & Krosshaug, 2011; Willems et al., 2005). Specifically, when the forefoot is unable to create a stiff and stable platform to maintain a medial centre of pressure combined with a large resultant GRF, the external inversion-adduction moment arm of the talocrural joint lengthens, increasing LAS risk (Ashton-Miller, Ottaviani, Hutchinson, & Wojtys, 1996; Chu et al., 2010; Fong, Chan, Mok, Yung, & Chan, 2009; Kristianslund et al., 2011).

Previous research has shown that a distal-proximal coupling relationship exists between the movement of the calcaneus and tibial rotation in straight-line running and walking tasks or low intensity change of direction tasks (Dubbeldam, Nester, Nene, Hermens, & Buurke, 2013; Nester, Jarvis, Jones, Bowden, & Liu, 2014; Nigg et al., 2017; Pohl, Messenger, & Buckley, 2007). Muscle activity and coupling relationships change as mode and intensity of locomotion changes (Pohl et al., 2007; Weir, Jewell, Emmerik, & Hamill, 2017). It is thus unclear if the same coupling relationships exists in high-intensity unanticipated changes of direction. Theoretically, in a distal-proximal coupling relationship of the lower limb, stiffening of the medial longitudinal arch (MLA) and limiting calcaneus eversion-abduction should also be able to mitigate the transfer or rotational movement to the knee via the shank, possibly decreasing ACL strain and decreasing ACLR risk. **Chapter 6** thus described and quantified the coupling relationship between the tri-planar movement of the calcaneus and the transverse plane rotations of the shank. The evaluation of the coordination patterns showed that a distal-proximal coupling relationship exist between calcaneus inversion/eversion and adduction/abduction and shank internal/external rotation. The shank's movement towards internal rotation followed calcaneus movement towards adduction and inversion in quick succession indicating distal-proximal coupling. Furthermore, the coupling relationship was observed within the loading phase, the period in which ACLR is likely to occur. Thus, it seems worthwhile to investigate whether manipulating calcaneus by strengthening the muscles acting on the foot may modulate movement within the loading phase to potentially decrease ACLR risk.

Chapter 7 described and quantified the coupling relationship between sagittal plane hallux and frontal plane calcaneus movement, and the coupling relationship between tri-planar metatarsal and frontal plane calcaneus movement. Similarly, the coupling relationship between sagittal plane hallux and transverse plane calcaneus, and triplanar metatarsal movement and transverse plane calcaneus movement, respectively, was described and quantified. LAS risk is initiated during the loading phase and increases during stance as angle velocities increase (Fong, Ha, Mok, Chan, & Chan, 2012; Kristianslund et al., 2011). The coordination evaluation revealed that during the loading phase, metatarsal extension acceleration was coupled with calcaneus eversion acceleration and metatarsal flexion acceleration with calcaneus inversion acceleration. During the same phase frontal plane metatarsal eversion deceleration was coupled with calcaneus eversion deceleration.

Throughout the stance phase, sagittal plane metatarsal and frontal plane calcaneus coordination was mostly anti-phase with metatarsal flexion accelerations coupled with calcaneus inversion accelerations. During the propulsion phase, where calcaneus inversion-adduction velocity reached their respective maximums, hallux flexion, metatarsal flexion, and metatarsal abduction accelerations were observed. During propulsion, the forefoot was thus conceivably responsible for providing forefoot stiffness and stability by flexion accelerations which where possibly generated by the intrinsic and extrinsic toe and forefoot flexors. It thus seems worthwhile to also investigate the effect strengthening the foot muscles may have on LAS risk.

The findings from investigating the calcaneus-shank and forefoot-rearfoot coupling relationships justified a pilot and proof-of-concept study to provide guidelines for future studies. Randomised controlled trials are considered the gold standard of intervention studies (Akobeng, 2005). However, large scale biomechanical randomised controlled trials are time consuming and expensive (Mullineaux & Wheat, 2017). Pilot studies are often undertaken to establish sample size and recruitment potential, to test prospective adaptations and thus the feasibility of a larger scale intervention (Thabane et al., 2010). **Chapter 8** described the pilot study that was conducted to provide proof the concept of a 16-week progressive foot muscle specific intervention programme on the risk factors associated with ACLR and LAS injury. The aim of the proof-of-concept study was to explore the effect of training the foot musculature on: 1) performance and kinetic parameters associated with change of direction tasks as well as ACLR and LAS injury risks; 2) the effect on the medial longitudinal arch (MLA) and the metatarsal anterior transverse arch (MetATA); 3) the change in segmental foot movement; and 4) risk factors associated with ACLR and LAS during unanticipated changes of direction. In addition, the proof-of-concept study aimed to 1) identify the biomechanical variables most likely to show adaptation as a result of the intervention; 2) identify the variables most likely to be different between athletes and potentially influence the intervention effect; and 3) provide sample size guidelines for future randomised controlled trials.

9.2. Practical applications

9.2.1 Calcaneus-shank coupling and its role in ACL injury risk

The results revealed that during the loading phase, where ACLR are most likely to occur, a distal-proximal coordination pattern exists in both the frontal plane calcaneus and transverse plane shank, and transverse plane calcaneus and transverse plane shank coordination patterns. Previous research has shown that neuromuscular control of the calcaneus via the stimulation of intrinsic muscles, abductor hallucis, flexor digitorum brevis, and quadratus plantae is possible (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014). Furthermore, frontal plane movement of the calcaneus is influenced by the concentric contraction of the extrinsic muscles, fibularis (longus and brevis) (Ashton-Miller et al., 1996) and eccentric function of the tibialis posterior (Fraser, Feger, & Hertel, 2016). The observed distal-proximal coupling between the calcaneus and shank, functional anatomy, and previous findings regarding muscle function of the foot muscles, justifies investigating whether foot muscle specific training can reduce ALCR risk.

9.2.2 Foot segment coupling

The forefoot was the main point of contact throughout the stance phase, typical for change of direction tasks in court sports (Losito, 2017). As such forefoot stiffness is important to provide stiffness and stability to the foot (Kessler, Lichtwark, Welte, Rainbow, & Kelly, 2020; Venkadesan et al., 2020). During the braking phase, calcaneus movement towards inversion (eversion deceleration) was coupled with metatarsal movement towards flexion (decelerating metatarsal extension) and inversion acceleration. The coupling effect between the fore foot and rear foot may be explained by the function of the plantar flexors and intrinsic foot muscles (Farris, Birch, & Kelly, 2020; Venkadesan et al., 2020). It was previously accepted that foot stiffness is created via the windlass mechanism, where the hallux and toes extend creating longitudinal stiffness as a result of the function of the plantar aponeurosis (Huson, 1991). This mechanism suggests that a strong coupling relationship should exist between hallux extension and calcaneus inversion-abduction. We did not observe a strong coupling between sagittal plane hallux and frontal plane calcaneus movement suggesting that soft tissue and neuromuscular activity was responsible for the calcaneal movement, rather than the windlass mechanism. Furthermore, previous research showed that when the forefoot is stable the calcaneus is allowed to move freely to align with the shank preserving lateral ankle stability (Fraser, Feger, & Hertel, 2016; Graf & Stefanyshyn, 2012).

During the propulsion phase, we observed calcaneus inversion/eversion coupled with metatarsal flexion/extension respectively, which is similar to other studies (Dubbeldam et al., 2013; Pohl & Buckley, 2008; Pohl et al., 2007). Sagittal plane metatarsal velocities remained larger than frontal plane calcaneus velocities, as in the braking phase, reiterating the importance of creating stiffness in the metatarsal segment providing stiffness and stability to the foot. Maximum calcaneus inversion and adduction acceleration was observed during the propulsion phase. In response, hallux and metatarsal flexion acceleration was observed from the onset of the propulsion phase (≈60 % of stance). When the centre of pressure moves to the forefoot, as expected during propulsion, toe flexors and other muscles responsible for forefoot flexion oppose the extension of the toes and metatarsals that is caused by the GRF (Huson, 1991; Reeser, Susman, & Stern Jr, 1983). The metatarsal and hallux segments thus were accelerated towards the floor, possibly under neuromuscular control of both intrinsic and extrinsic forefoot and hallux flexors, enabling the forefoot to maintain ground contact and provide stiffness and stability during propulsion (Farris et al., 2020). It is thus plausible that training the intrinsic and extrinsic muscles of the foot can provide stiffness and stability to the foot and is worthwhile investigating. A stiffer and possibly more stable foot seems to decrease calcaneus inversion-adduction velocity. Furthermore, when the forefoot is stable the calcaneus is free to align with the shank, possibly decreasing LAS risk.

9.2.3 Proof of concept

The aim of the proof-of-concept study was to firstly explore the effect strengthening the foot musculature has on performance and kinetic parameters. As well as the effect on foot function and the risk factors associated with ACL and LAS during change of direction tasks. Secondly the proof-of-concept study aimed to identify biomechanical variables that are most likely to be influenced by the intervention, identify pre-intervention biomechanical variables that could influence the effect of the intervention, and provide guidance for future sample size selection.

Thirty-five female athletes aged 16 to 25, and competing in court sports (netball, volleyball, racquet sports, and futsal) were recruited from local clubs and high schools. Eleven athletes were either excluded on the basis of not being able to attend the first testing session, having had a serious lower limb injury in the previous six months, or a BMI higher than 30 **(see Chapter 7)**. The remaining 24 athletes were matched for sport and BMI and then randomised to TG or CG. TG underwent a 16-week progressive foot muscle specific intervention, in addition to normal sport specific training, while CG continued normal sport specific training. TG performed the foot muscle specific exercises three times per week. One session per week was supervised by the researcher and two sessions per week were unsupervised. The exercises in the intervention need little equipment and were chosen with the aim to improve foot mobility, increase neuromuscular control and strengthen the muscles acting on the foot. Exercises increased in complexity and intensity over four phases. Ground reaction force data and 3D motion capture was used to collect multisegmental movement of the dominant foot and the lower limb during high intensity change of direction tasks.

Four athletes dropped out of the TG. One athlete sustained a stress fracture to the 5th metatarsal (the injury was unrelated to the intervention), one athlete had to withdraw as a result in change of circumstance, and two athletes were unavailable to attend the post-intervention testing session. Eight from the initial twelve athletes completed the 16-week intervention with a compliance rate of 89 % (range 69 % to 97 %) to the supervised session and 72 % (range 58 % to 84 %) to the unsupervised sessions.

To ensure reliability of the comparison of pre-intervention to post-intervention results, it is important to identify all variables that could influence an athlete’s response to a foot muscle specific intervention. The pilot study found pre-intervention differences between groups that could potentially influence the athlete’s adaptation to a foot muscle specific intervention. We therefore recommend that in addition to matching athletes for sport and BMI, athletes should also be matched for stance time, peak medial-lateral GRF, maximum hallux-metatarsal extension angle, maximum inversion/eversion and adduction/abduction MidCal angles, as well as maximum knee valgus angle and maximum knee valgus moment arm length.

To provide guidance for future studies we identified the variables that were influenced by the intervention. R^2 goodness-of-fit measures were used to evaluate the adaptation of the height and length of the MLA and MetATA as well as the segmental foot angles over the stance phase within groups. An unbalanced ANOVA was used to test pre- to post-intervention adaptations of the maximums of the ACL and LAS injury risk factor outcome variables between groups. The sample sizes needed to find significance for each of the identified variables were then calculated to determine feasibility of future studies of similar design (Noordzij et al., 2010). The significant differences, or trend towards significant, adaptations between TG and CG for the variables; MLA height, sagittal plane MetMid, triplanar MidCal and sagittal plane CalShk suggests that TG presented with a foot that is more resistant to deformation post-intervention. Excessive dynamic rearfoot pronation is thought to increase knee valgus angles and ACL injury risk (Holland, 2016; Kagaya, Fuji, & Nishizono, 2013; McLean, Lipfert, & Van Den Bogert, 2004). Thus, the increased resistance to deformation in the feet of TG athletes may have had an influence on the smaller knee valgus angle observed for TG and potentially decreased ACL injury risk. Similarly, the MetATA for the TG presented with a trend towards a stiffer forefoot as a result of the larger MetATA height, lower MetATA length and lower sagittal plane HalMet R^2 values. The stiffer forefoot in TG could have been responsible for the trend towards a smaller ankle inversion angle and the smaller eversion moment arm for TG compared to CG, shifting the resultant GRF of the TG closer to the ankle joint centre. In this way, possibly decreasing LAS injury risk.

The sample size calculation for MLA height and MetATA height and length, sagittal plane MetMid and CalShk angles, frontal and transverse plane MidCal angles all fall within the recommended sample size (30 athletes per group) (Mullineaux & Wheat, 2017). However, the sample size needed in TG to determine the influence on ACL and LAS risk factors fall outside the minimum recommended sample size of 30. It thus seems that an intervention study of similar design is worth perusing if the outcome is limited to investigating changes in foot function. A much larger number of athletes per group is needed to investigate the effect of a foot muscle specific intervention on outcome variables associated with ACL and LAS injury risk factors.

9.3. THESIS LIMITATIONS AND FUTURE DIRECTIONS

9.3.1 LIMITATIONS

Limitations mainly surrounded methodology of data collection and adherence to the intervention.

Unanticipated change of direction stimulus was controlled via a pre-programmed timing gate system which provided both an audio and visual stimulus the athlete had to respond to. However, an earlier study found increased peak medial GRF, hip flexion angle, hip abduction angle and knee valgus angle when athletes had to evade opponents (McLean et al., 2004). The current study may therefore not have evoked appropriate biomechanical outcomes. However, adding an opponent inside the motion capture area would have obstructed the camera's field of view.

The approach speed was monitored to fall within 10 % of 4.5 $m.s^{-1}$. As athlete's have different abilities, some may have found changing direction at this speed taxing and other less so. Thus, athletes who found the task easy may change their biomechanics when faced with a more challenging task, obscuring our findings here. Furthermore, the stimulus was set to 0.7 sec after the approach run was started. This may be inappropriate as athletes have different abilities and some athletes may have had more time from stimulus to the force plate where they were instructed to change direction resulting in the task being anticipated. Additionally, the approach speed was higher at post-intervention compared to pre-intervention for both groups. Although, the approach speed was not significantly different between groups, most variables showed changes from pre-intervention to post-intervention. The significantly faster speed at post-intervention could have influenced the pre-intervention to post-intervention results.

Injury history was limited to an injury six months prior to the recruitment date. Athletes may have returned to sport after injury, but it is believed that athletes with a history of LAS develop chronic ankle instability. Chronic ankle instability influences the neuromechanics of the entire lower limb (Hertel, 2002) and may have influenced the biomechanics of the lower limb in this project. Recruiting experienced female court sport athletes with no history of injury would be difficult, as the population pool is small and history of injury high (Doherty et al., 2014; Langeveld, Holtzhausen, & Coetzee, 2012).

Compliance rates were low for unsupervised sessions. Higher compliance rates could have resulted in different adaptations to TG. Future studies should aim to improve compliance rates by incorporating the intervention into existing pre-training warm-up programmes rather as additional sessions to normal training (Herman, Barton, Malliaras, & Morrissey, 2012).

9.3.2 Considerations for future studies

The findings of this thesis and the insights gained provide valuable recommendations for future RCT studies.

The fact that athletes from different sports were included is a strength of the study, However, the 45° unanticipated change of direction tasks may be more prevalent in sports like netball and footsal, but not in sports like volleyball, tennis and badminton. Some athletes may thus have found the change of direction tasks more challenging than others. Furthermore, an element of motor learning could be present in the athletes that are not familiar with the movement. We did not control for motor learning in athletes that may not be familiar with the movement and this may have influenced pre- to post-intervention results. Future studies may choose a fundamental skill like a hopping task with unanticipated change of direction jump for distance.

Pre-existing anatomical variations and biomechanics may influence the individual reaction to intervention. We also observed large variability in segment velocities and thus coordination patterns. The velocity signals were averaged before coupling angles were calculated. Coupling angles and coordination patterns may thus not be specific to the individual. Future studies may choose a quasi-experimental design which does not require randomization of the participants into a TG and CG, but rather the two groups as similar as possible at pre-intervention (White & Sabarwal, 2014).

Muscle activity and muscle strength was not measured to gauge the effectiveness of the intervention. This study also did not consider the effect of fatigue, which is regarded to play a role in increasing ACL injury risk (Brüggeman, 1996) or the effect of chronic ankle instability which is known to increase the risk for future LAS (McCriskin, Cameron, Orr, & Waterman, 2015). Future studies should consider comparing pre- to post-intervention measures of strength, muscle activity, foot stiffness, and ankle stability and stiffness. It would also be prudent to study the effect of fatigue on in the coupling relationship between segments, foot function or injury risk factors

It is known that skin markers move relative to the bony landmark it represents, influencing the reliability of kinematic data (Alenezi, Herrington, Jones, & Jones, 2016). As technology improves and marker less biomechanical studies are refined, more accurate representation and tracking of foot segment should be considered. Furthermore, the current biomechanical analysis was conducted barefoot as surface markers are more visible in barefoot condition than in shod. It is thus unclear whether these findings can be generalized beyond the barefoot condition. Future studies should also investigate analysis in shoes designed for the respective sports.

Only females between 16 – 25 years of age were recruited. The findings here can thus not be generalized to different age groups or other genders. Future studies should include a larger and more representative group of the sporting population

9.4. CONCLUSIONS

A distal-proximal coupling relationship between frontal and transverse plane movement of the calcaneus and transverse plane rotational movement of the tibia was established. These findings support the hypothesis that training the muscles acting on the foot will influence rearfoot/calcaneus coordination in unanticipated change of direction tasks and conceivably influence ACLR risk factors. Coupling relationships between frontal plane calcaneus and sagittal plane metatarsal movement was observed in the loading and stance phase of unanticipated change of direction tasks. In the propulsion phase, sagittal plane hallux and metatarsal flexion acceleration was observed in response to calcaneus inversion and adduction acceleration. Sagittal plane forefoot velocity was higher than calcaneus velocity in this period, indicating that the forefoot is leading the coupling relationship and possibly providing stiffness and stability in the forefoot, which would provide stiffness to the foot, but also allow the calcaneus to move freely and align with the shank to preserve lateral ankle stability.

The proof-of-concept study found that training muscles acting on the foot may increase the foot's resistance to deformation and increases forefoot stiffness. The improved resistance to deformation of the MLA and stiffness in the MetATA of TG may have played a role in decreasing the maximum knee valgus angle, and subsequently may indicate a decrease in ACLR risk. The observed trend to decreases in maximum ankle inversion angle and maximum ankle eversion moment arm length of TG may further decrease LAS risk.

The proof-of-concept pilot study aimed to identify variables that could influence the effect of the intervention. In addition to matching athletes for sport and BMI it is also advisable to establish individual's stance time, peak medial-lateral GRF, maximum hallux-metatarsal extension angle, maximum midfoot-calcaneus inversion/eversion, and adduction/abduction angles, as well as maximum knee valgus angle and maximum knee valgus moment arm length before randomisation.

A further aim of pilot studies is to provide direction regarding variables that are most likely to show significant adaptation to intervention with an attainable and realistic sample size. A sample size of 30 athletes per group is recommended for biomechanical studies (Mullineaux & Wheat, 2017). The R^2 values of MLA height, MetATA height and length were the only variables identified that showed significant adaptation to the intervention, or displayed a trend towards a significant adaptation, and the power calculation showed a realistic sample size. The variables, maximum/minimum sagittal plane metatarsal-midfoot angle, frontal- and transverse plane midfoot-calcaneus angles and sagittal plane calcaneus-shank angles also had significant adaptations and displayed attainable sample size recommendations for future studies.

Training the muscles of the foot thus seems to change dynamic foot function and a reasonable number of athletes per group is necessary to determine significance, suggesting that future foot muscle specific intervention studies are feasible. However, the low compliance rate may have influenced the effect of the intervention on ACLR and LAS risk factors. Thus, the proof-of-concept study was unable to provide definitive guidance whether training the muscles acting on the foot should be implemented to supplement current ACLR and LAS prophylactic programmes. Future studies should follow the guidelines provided here to establish the effectiveness of implementing a foot muscle specific training programme to supplement current ACLR and LAS prophylactic programmes.

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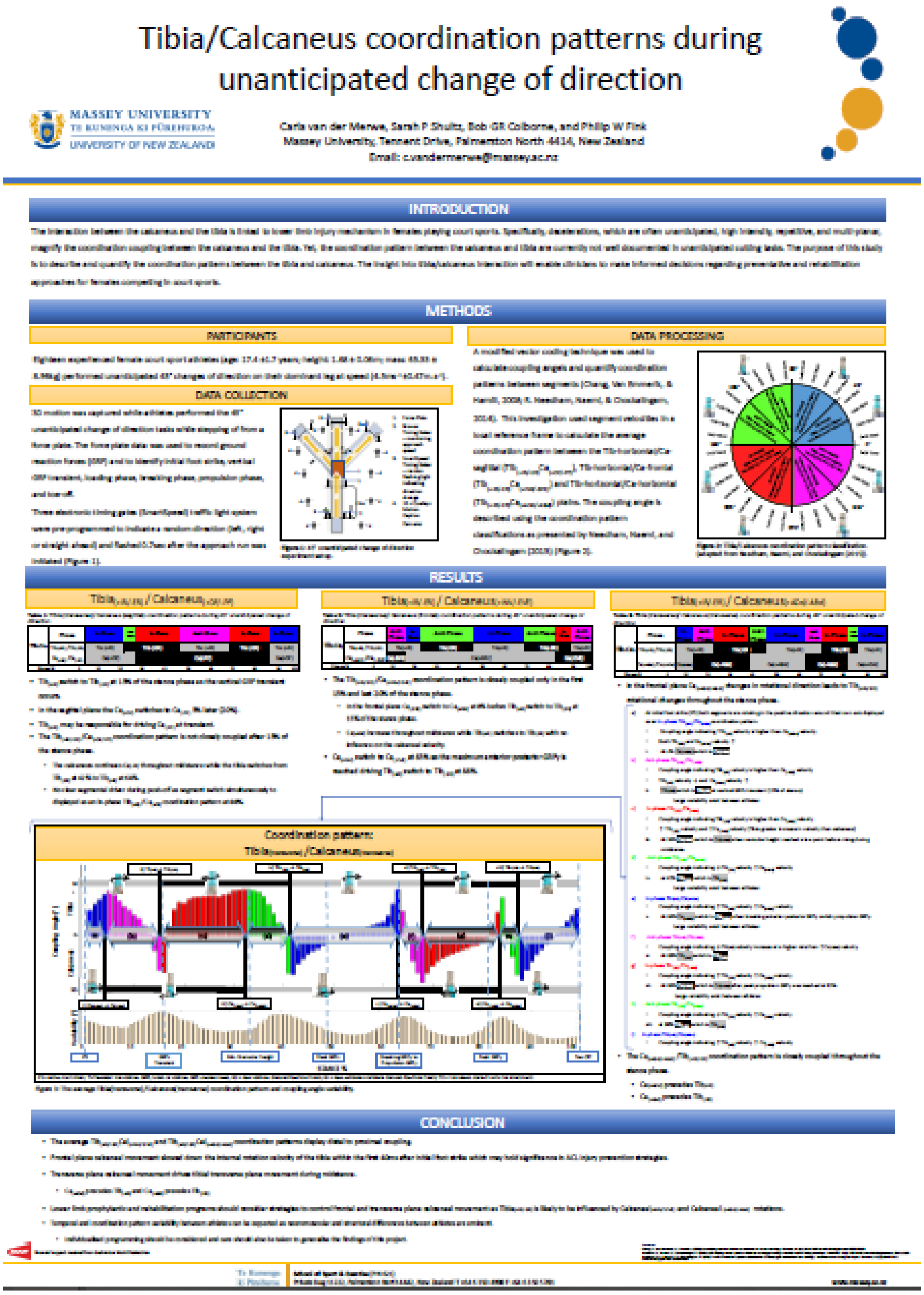
APPENDICES

APPENDIX A: PRESENTATIONS

CONFERENCE PRESENTATIONS

POSTER PRESENTATION: 11TH AUSTRALASIAN BIOMECHANICS CONFERENCE, AUCKLAND, NEW ZEALAND, 3-5 DECEMBER 2018



ORAL PRESENTATION: INTERNATIONAL SOCIETY OF BIOMECHANICS, CALGARY, CANADA, 31 JULY – 4 AUGUST 2019

The effect of a 16-week foot muscle specific intervention program on ACL and LAS injury mechanisms

Carla van der Merwe[1], Sarah P. Shultz[2], G Robert Colborne[3], Philip W Fink[1]
[1]School of Sport, Exercise and Nutrition, Massey University, Palmerston North, New Zealand
[2]Department of Kinesiology, University of Seattle, Seattle, WA, United States of America
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Summary

A 16-week progressive foot muscle specific intervention program decreased some parameters associated with non-contact anterior cruciate ligament (ACL) and lateral ankle sprain (LAS) injury mechanisms. Changes in inter-foot and foot-shank segmental relationships were also observed.

Introduction

Abrupt, unanticipated changes of direction put participants of court sports at an increased risk of sustaining lower limb ligament injuries. Foot mechanics play a key role in lower limb injury mechanisms, however, very few prophylactic programs aim to improve foot function directly. The aim of this project was to investigate the effect that specific foot muscle strengthening had on foot and shank segment kinematics and how these changes in kinematics, if any, relate to mechanisms associated with ACL and LAS injuries.

Methods

Eighteen skilled female court sports athletes (age: 17.4 ±1.7 years; height: 1.68 ± 0.06m; mass: 65.53 ± 8.96kg) performed unanticipated, 45° changes of direction on their dominant leg at running speed (4.5ms^{-1} ±0.47m.s^{-1}).

Kinematic and kinetic data were collected from the dominant leg by a ten camera Qualisys system. A Kistler force platform was used to collect ground reaction forces (GRF). Spherical markers were placed on the lower limb and feet using a multi-segmental foot model [1].

The athletes were matched for sport and BMI and then randomly placed in either the training (N=8) or control (N=10) group. The training group underwent a progressive 16-week foot and lower leg muscle specific strengthening program. The exercises were performed three times per week; one session each week was supervised by the researcher.

Results and Discussion

Data was collected during the braking phase for ACL and from foot strike to peak propulsion force for LAS injury parameters (Figure 1 and Table 1). For ankle injury parameters the training group trended towards a smaller increase in inversion velocity and inversion moment arm length. The training group also had an increase in the eversion moment arm length, all of which suggest a decrease in LAS risk.

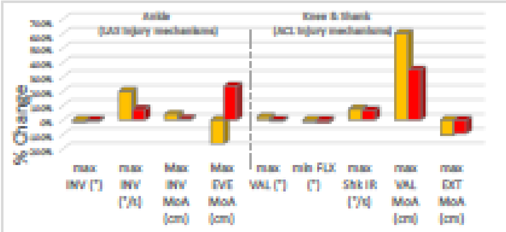

Figure 1: The % change in injury mechanism parameters from pre- to post-intervention in the control and training groups.
Yellow=control group, Red= intervention group

For knee injury mechanisms, the average valgus angle and external valgus GRF moment arm tended to be smaller for the training group compared to the control group pre- to post-intervention, possibly decreasing the risk for ACL injury.

Conclusions

Strengthening the muscles acting on the foot displayed a trend towards a more stable front foot. It is likely that this change reduced the frontal plane ankle GRF-moment arms, decreasing LAS injury risk. Furthermore, increasing calcaneal resistance to eversion-abduction likely caused the decrease in the knee valgus angle and the valgus GRF moment arm, reducing ACL injury risk.

Acknowledgements

Funding for this project was granted by the Badminton World Federation.

References

[1] Leardini A et al. (2007). *Gait Posture*, 25: 453-62

Table 1: Injury parameters pre-and post-intervention in the control and training groups. Percent change from pre-intervention are in brackets

		Max. Ankle Inversion angle (°)	Max. Ankle Inversion velocity (°/s)	Max. Ankle Inversion MoA (cm)	Max. Ankle Eversion MoA (cm)	Max. Knee valgus angle (°)	Min. Knee Flexion angle (°)	Shank Internal rotation velocity (°/s)	Max. Knee Valgus MoA (cm)	Max. Knee Extension MoA (cm)
Control Group	**Pre**	28.266	59.241	0.849	0.115	6.391	32.794	176.230	0.583	0.045
	Post	23.439 (-17%)	177.474(200%)	1.198 (41%)	-0.073 (-164%)	7.735 (21%)	27.571 (-16%)	314.071 (78%)	4.090 (602%)	-0.002 (105%)
Training Group	**Pre**	30.296	54.490	0.631	0.116	7.739	35.478	200.531	0.540	0.040
	Post	26.642 (-12%)	95.711 (76%)	0.678 (8%)	0.391 (236%)	6.956 (-10%)	28.604 (-19%)	337.540 (68%)	2.423 (349%)	0.002 (-94%)

MoA =GRF moment arm

ORAL PRESENTATION: Sport and Exercise Science New Zealand, Palmerston North, New Zealand, 27 – 29 November 2019

The effect of a 16-week foot muscle specific intervention program on non-contact anterior cruciate ligament (ACL) and lateral ankle sprain (LAS) injury risk

Carla van der Merwe[1], Sarah P. Shultz[2], Bob GR Colborne[3], Kim Hébert-Losier[4], Philip W Fink[1]

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Introduction

ACL injury risk increase when excessive subtalar joint pronation (medial longitudinal arch (MLA) height), coupled with internal tibial rotation creates large knee valgus angles under extreme loads. Forefoot (metatarsal anterior transverse arch (MetATA)) instability, linked to larger ankle moment arms increases LAS injury risk. We investigated the effect of a foot-muscle specific intervention on ACL and LAS injury risk factors.

Methods

Eighteen skilled female court sport athletes were matched (sport and BMI) and randomized to the training (TG) or control group (CG). Athletes performed unanticipated, 45° changes of direction barefoot at speed. 3D motion (multi-segmental foot model [1]) and force data was collected from the dominant lower limb. The TG underwent a progressive 16-week foot muscle specific exercise program. R^2 goodness-of-fit and ANOVA analysis tested pre-to-post-intervention adaptations within and between groups respectively.

Results

TG R^2 MLA height values were larger ($p < 0.05$), MetATA height values larger ($p = 0.08$) and length smaller ($p = 0.18$). TG maximum knee valgus angle decreased ($p = 0.20$), ankle inversion angle had a smaller increase ($p = 0.11$) and ankle eversion moment arm length had a larger decrease ($p = 0.13$).

Discussion

The TG had dynamically stiffer arches compared to the CG, possibly decreasing some ACL and LAS injury risk factors.

Conclusions

Training the foot muscles increased arch stiffness, possibly influencing the frontal plane knee and ankle biomechanics, decreasing some ACL and LAS injury risk factors.

Take home message

A 16-week progressive foot muscle specific intervention changed foot function and decreased some ACL and LAS injury risk factors.

Acknowledgments

Funding for this project was granted by the Badminton World Federation.

References

[1] Leardini A et al. (2007). *Gait Posture*, **25**: 453-62

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*Phone:*0210 260 9008*
*Type of file: *Oral*
*Emerging researcher? *Yes*
*Student or Full Time Employment? *Full time employment*
*Abstract discipline (Closest match): *biomechanics*

Appendix B: Publications

Publications (In Press and under Review)

REVIEW OF THE LITERATURE (IN PRESS)

RESEARCH QUARTERLY FOR EXERCISE AND SPORT
https://doi.org/10.1080/02701367.2020.1739605

Routledge
Taylor & Francis Group

SPECIAL TOPIC

Check for updates

Foot Muscle Strengthening and Lower Limb Injury Prevention

Carla van der Merwe [a], Sarah P. Shultz[b], G. Robert Colborne[a], and Philip W. Fink[a]

[a]Massey University; [b]University of Seattle

ABSTRACT

Background and objectives: The active and passive structures of the foot act in unison to not only be compliant enough to assist in ground reaction force attenuation but also resist deformation to provide a stable base of support. A foot that is unable to adjust to the imposed demands during high-intensity sporting activities may alter the moments and forces acting on the joints, increasing the risk of non-contact anterior cruciate ligament ruptures (ACLR) and lateral ankle sprains (LAS). Prophylactic strengthening programs are often used to reduce the risk of these injuries, but at present, very few prophylactic programs include foot-specific strengthening strategies. The aim of this theoretical review is to ascertain the prophylactic role strengthening muscles acting on the foot may have on ACLR and LAS injury risk. **Methods**: Literature relating to risk factors associated with ACLR and LAS injury and the anatomy and biomechanics of normal foot function was searched. In addition, ACLR and LAS injury prevention programs were also sought. A theoretical, narrative approach was followed to synthesize the information gathered from the articles. **Results**: The foot segments are governed by the congruity of the articulations and the activity of the foot muscles. As such, there is a coupling effect between shank, calcaneus, midfoot, and hallux movement which play a role in both ACLR and LAS injury risk. **Conclusions**: Strengthening the muscles acting on the foot may have a significant impact on ACLR and LAS injury risk.

ARTICLE HISTORY
Received 21 February 2019
Accepted 2 March 2020

KEYWORDS
Athletic injury; anterior cruciate ligament; ankle sprain; foot; biomechanics

The anatomy and function of the foot are remarkably complex with both passive and active structures functioning as a single component. The foot segments act as an interactive unit to not only become a stiff configuration to support and propel the body but also become compliant enough to adapt to the contact surface and dissipate ground reaction forces (GRF) (Gwani et al., 2015; Holowka et al., 2017; McKeon et al., 2014; Williams et al., 2014). Yet very little is known about the foot's role in injury mechanics and subsequently, the possible prevention of the injuries that athletes are exposed to during high impact sports. In sports where abrupt, unanticipated changes of direction are common, the athlete is at an increased risk for sustaining non-contact injuries, such as anterior cruciate ligament ruptures (ACLR) and lateral ankle sprains (LAS) (Cassell et al., 2012; Doherty et al., 2014; Fong, Chan et al., 2009; McKay et al., 2001). ACLR and LAS injuries primarily occur during unbalanced single-leg decelerations (B. P. Boden et al., 2009; Fong et al., 2012; Kristianslund et al., 2011; McCriskin, 2015; Olsen et al., 2004; Stuelcken et al., 2015). During movements where lower body stability is compromised (Powers, 2010; Stacoff et al., 1996) and GRFs are large (B. P. Boden et al., 2009; B. P. S. Boden et al., 2010; Hewett et al., 2005; Raina & Nuhmani, 2014; Serpell et al., 2012), external moments around the knee or ankle can exceed the injury threshold of the structures stabilizing the joint, causing injury to the soft tissue of joints (Fong, Chan et al., 2009; Fong et al., 2012; Kristianslund et al., 2011; Powers, 2010).

Prophylactic programs have been successful in minimizing the incidence of ACLR and LAS injury; they commonly include physical (i.e. balance, plyometric, strength, coordination) training, education about injury mechanisms, and/or technique training and feedback (Kaminski et al., 2013; Kerkhoffs et al., 2012; McCriskin, 2015; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Paszkewicz et al., 2012; Schiftan et al., 2015; Shultz et al., 2015; Yoo et al., 2010). However, these programs often fail to report on the mechanisms involved in the reduction of injury incidences (Eils et al., 2010; Hewett & Johnson, 2010; Kaminski et al., 2013; Kynsburg et al., 2010; McKay et al., 2001; McKeon & Mattacola, 2008; Noyes & Barber-Westin, 2014; Paszkewicz et al., 2012; Vriend et al., 2016; Yoo et al., 2010). Jump landings, stopping, and changing direction are all closed kinetic chain

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DRC 16

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STATEMENT OF CONTRIBUTION
DOCTORATE WITH PUBLICATIONS/MANUSCRIPTS

We, the candidate and the candidate's Primary Supervisor, certify that all co-authors have consented to their work being included in the thesis and they have accepted the candidate's contribution as indicated below in the *Statement of Originality*.

Name of candidate:	Carla van der Merwe
Name/title of Primary Supervisor:	Philip W Fink
Name of Research Output and full reference:	
Foot muscle strengthening and lower limb injury prevention	
In which Chapter is the Manuscript /Published work:	Chapter 4
Please indicate:	
• The percentage of the manuscript/Published Work that was contributed by the candidate:	90%
and	
• Describe the contribution that the candidate has made to the Manuscript/Published Work:	
Search of the literature and drafting the manuscript.	
For manuscripts intended for publication please indicate target journal:	
Research Quarterly for Exercise and Sport	
Candidate's Signature:	Carla van der Merwe Digitally signed by Carla van der Merwe Date: 2020.06.05 14:42:04 +12'00'
Date:	05 June 2020
Primary Supervisor's Signature:	Philip Fink Digitally signed by Philip Fink Date: 2020.07.01 11:35:23 +12'00'
Date:	1 July 2020

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GRS Version 4– January 2019

CHAPTER 5 (UNDER REVIEW)

Using a modified vector coding technique to describe the calcaneus-shank coupling relationship during unanticipated changes of direction: theoretical implications for prophylactic ACL strategies.

Carla van der Merwe[1], Sarah P. Shultz[2], G Robert Colborne[3], Kim Hébert-Losier[4], Philip W Fink[1]

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Conflict of Interest Disclosure: None.

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Running Title: *Lower limb coupling in unanticipated cutting task*

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STATEMENT OF CONTRIBUTION
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We, the candidate and the candidate's Primary Supervisor, certify that all co-authors have consented to their work being included in the thesis and they have accepted the candidate's contribution as indicated below in the *Statement of Originality*.

Name of candidate:	Carla van der Merwe
Name/title of Primary Supervisor:	Philip W Fink
Name of Research Output and full reference:	
Using a modified vector coding technique to describe the calcaneus-shank coupling relationship during unanticipated changes of direction: Theoretical implications for prophylactic ACL strategies	
In which Chapter is the Manuscript /Published work:	Chapter 5
Please indicate:	
• The percentage of the manuscript/Published Work that was contributed by the candidate:	85%
and	
• Describe the contribution that the candidate has made to the Manuscript/Published Work:	
Data collection and processing, statistical analysis and drafting the manuscript.	
For manuscripts intended for publication please indicate target journal:	
Physiotherapy Practice and Research	
Candidate's Signature:	Carla van der Merwe Digitally signed by Carla van der Merwe Date: 2020.06.05 14:42:04 +12'00'
Date:	05 June 2020
Primary Supervisor's Signature:	Philip Fink Digitally signed by Philip Fink Date: 2020.07.01 11:35:50 +12'00'
Date:	1 July 2020

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CHAPTER 6 (UNDER REVIEW)

ORIGINAL ARTICLE

The coordination patterns of the foot segments in relation to lateral ankle sprain injury mechanism during unanticipated changes of direction.

Carla van der Merwe[a], Sarah P. Shultz[b], G Robert Colborne[c], Kim Hébert-Losier[d], Philip W Fink[a]

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Declarations of interest: none

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STATEMENT OF CONTRIBUTION
DOCTORATE WITH PUBLICATIONS/MANUSCRIPTS

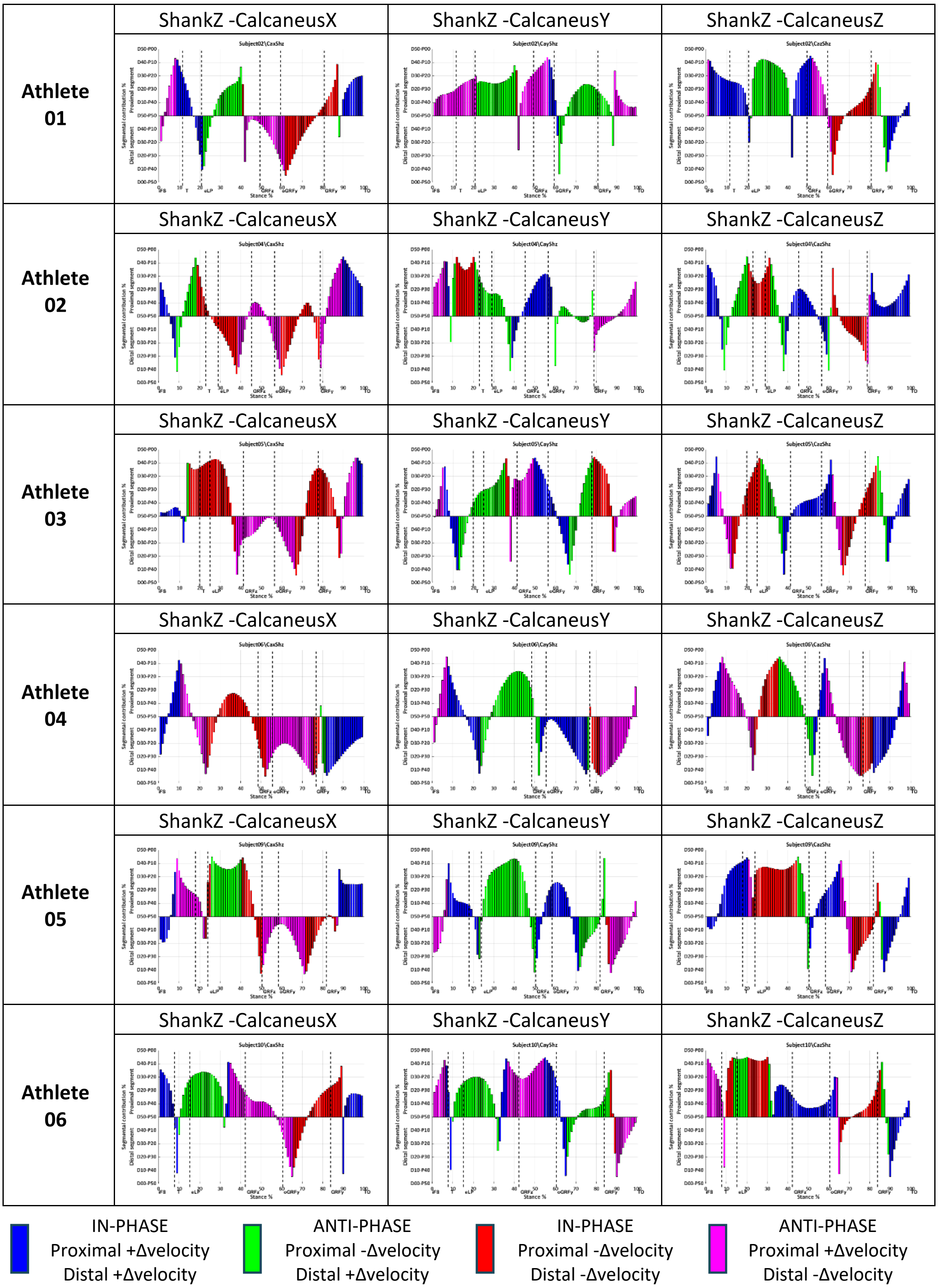
We, the candidate and the candidate's Primary Supervisor, certify that all co-authors have consented to their work being included in the thesis and they have accepted the candidate's contribution as indicated below in the *Statement of Originality*.

Name of candidate:	Carla van der Merwe
Name/title of Primary Supervisor:	Philip W Fink
Name of Research Output and full reference:	
DESCRIBING THE COUPLING RELATIONSHIP BETWEEN FOOT SEGMENTS DURING UNANTICIPATED CHANGES OF DIRECTION: THEORETICAL IMPLICATIONS FOR PROPHYLACTIC LATERAL ANKLE SPRAIN STRATEGIES	
In which Chapter is the Manuscript /Published work:	Chapter 6
Please indicate:	
• The percentage of the manuscript/Published Work that was contributed by the candidate:	85%
and	
• Describe the contribution that the candidate has made to the Manuscript/Published Work:	
Data collection and processing, statistical analysis and drafting the manuscript.	
For manuscripts intended for publication please indicate target journal:	
The Foot	
Candidate's Signature:	Carla van der Merwe Digitally signed by Carla van der Merwe Date: 2020.06.05 14:42:04 +12'00'
Date:	05 June 2020
Primary Supervisor's Signature:	Philip Fink Digitally signed by Philip Fink Date: 2020.07.01 11:36:15 +12'00'
Date:	1 July 2020

(This form should appear at the end of each thesis chapter/section/appendix submitted as a manuscript/ publication or collected as an appendix at the end of the thesis)

APPENDIX C: INDIVIDUAL COORDINATION PATTERNS

INDIVIDUAL COORDINATION PATTERNS: SHANK-CALCANEUS



Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

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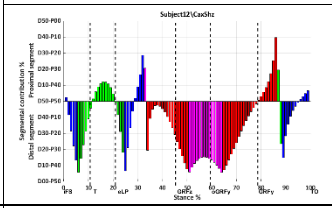
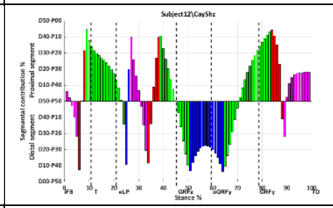
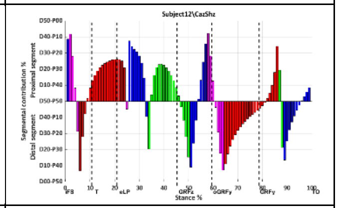
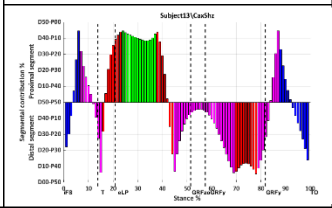
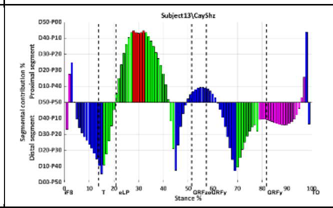
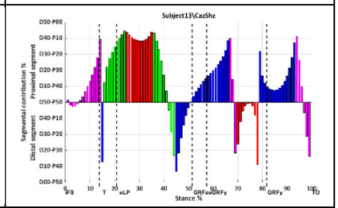
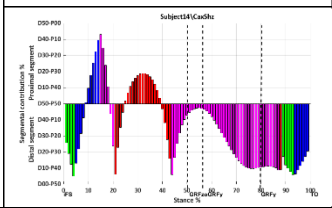
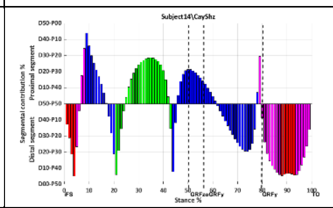
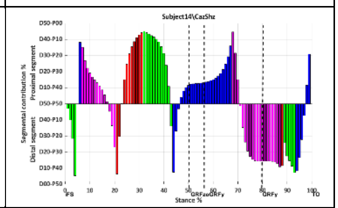
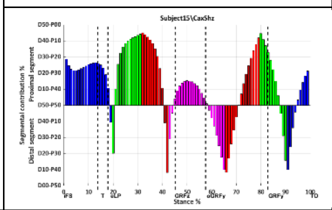
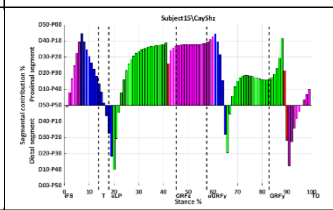
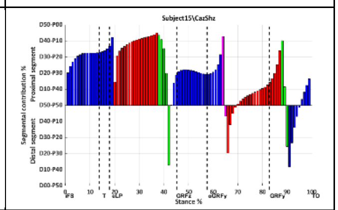
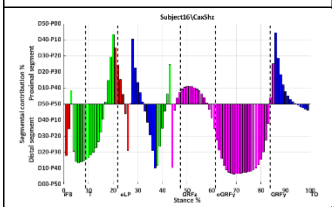
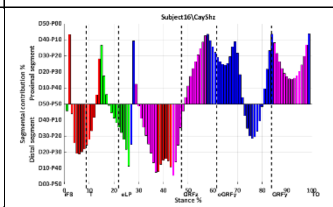
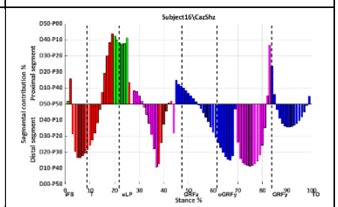
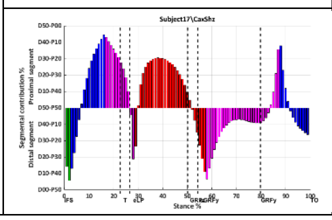
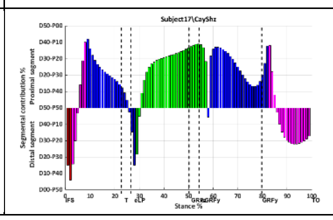
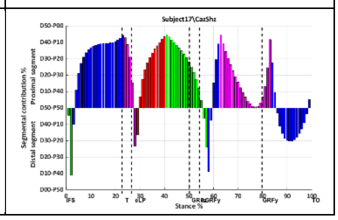
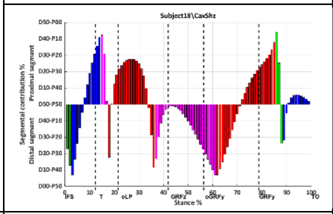
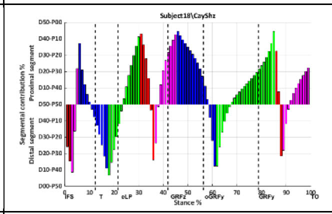
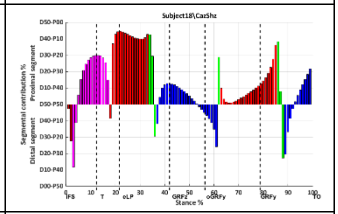
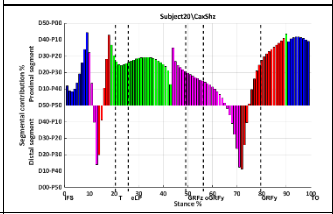
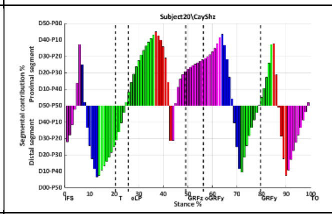
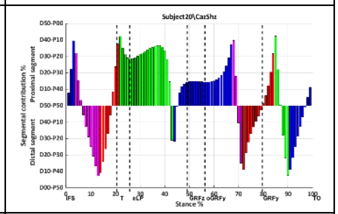
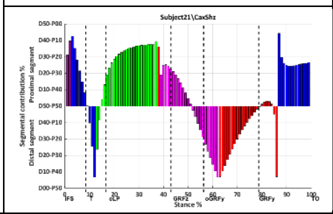
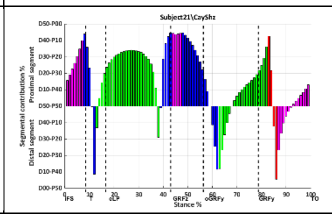
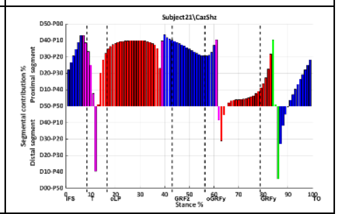
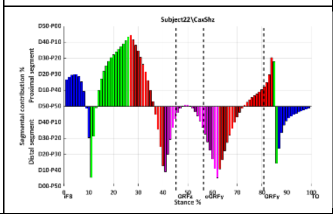
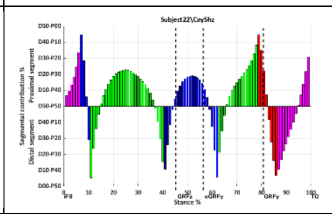
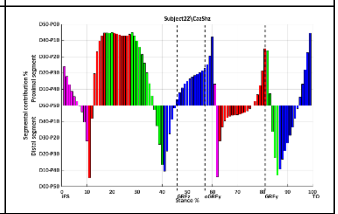
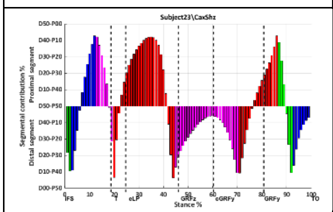
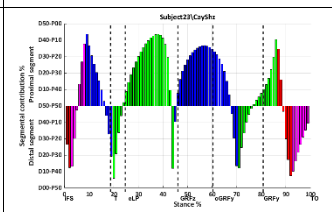
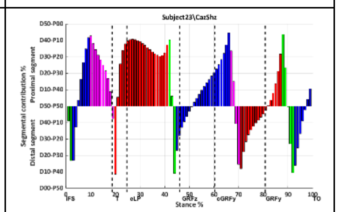
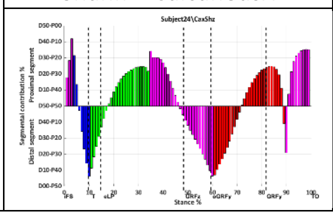
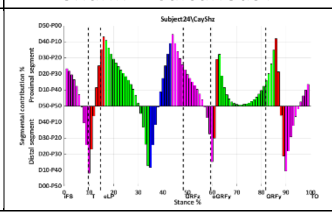
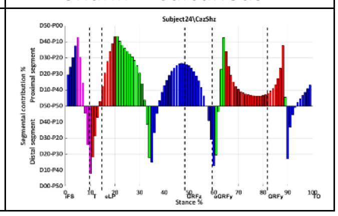
Athlete 07			
Athlete 08			
Athlete 09			
Athlete 10			
Athlete 11			
Athlete 12			

IN-PHASE Proximal +Δvelocity Distal +Δvelocity	**ANTI-PHASE** Proximal -Δvelocity Distal +Δvelocity	**IN-PHASE** Proximal -Δvelocity Distal -Δvelocity	**ANTI-PHASE** Proximal +Δvelocity Distal -Δvelocity

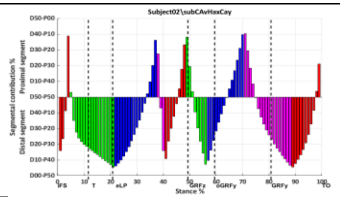
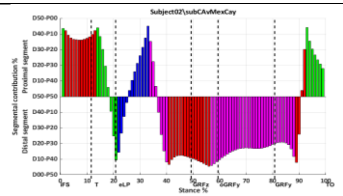
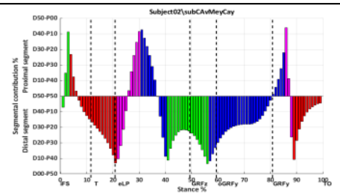
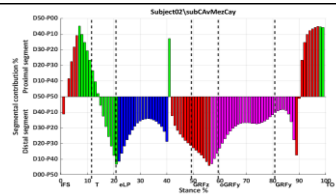
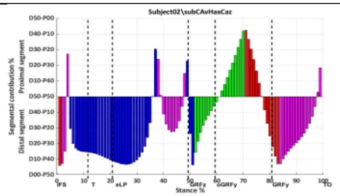
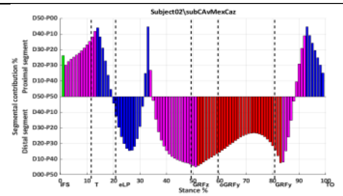
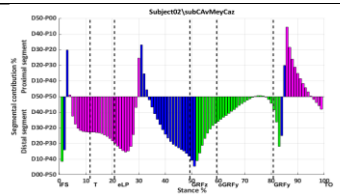
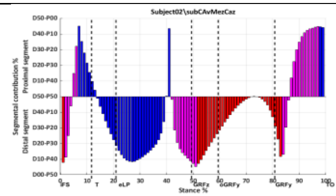
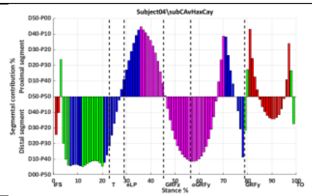
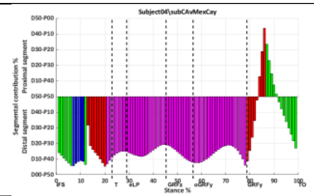
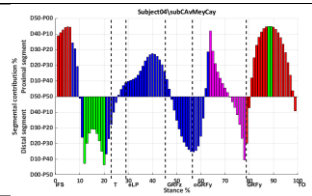
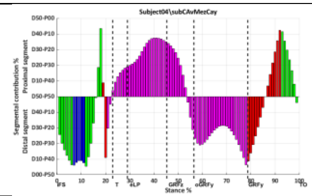
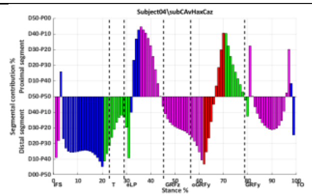
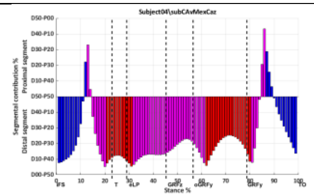
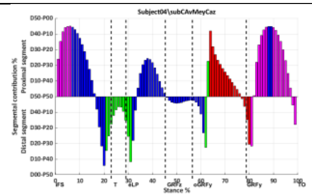
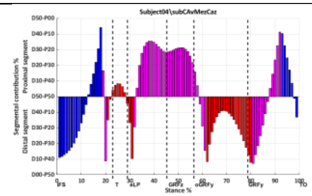
Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

	ShankZ -CalcaneusX	ShankZ -CalcaneusY	ShankZ -CalcaneusZ
Athlete 13			
	ShankZ -CalcaneusX	ShankZ -CalcaneusY	ShankZ -CalcaneusZ
Athlete 14			
	ShankZ -CalcaneusX	ShankZ -CalcaneusY	ShankZ -CalcaneusZ
Athlete 15			
	ShankZ -CalcaneusX	ShankZ -CalcaneusY	ShankZ -CalcaneusZ
Athlete 16			
	ShankZ -CalcaneusX	ShankZ -CalcaneusY	ShankZ -CalcaneusZ
Athlete 17			
	ShankZ -CalcaneusX	ShankZ -CalcaneusY	ShankZ -CalcaneusZ
Athlete 18			

IN-PHASE Proximal +Δvelocity Distal +Δvelocity	**ANTI-PHASE** Proximal -Δvelocity Distal +Δvelocity	**IN-PHASE** Proximal -Δvelocity Distal -Δvelocity	**ANTI-PHASE** Proximal +Δvelocity Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

INDIVIDUAL COORDINATION PATTERNS: FOOT SEGMENTS

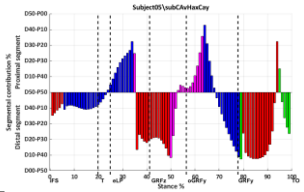
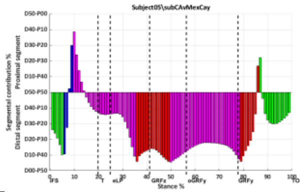
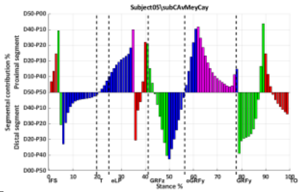
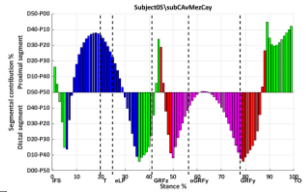
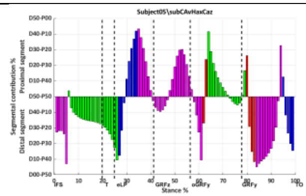
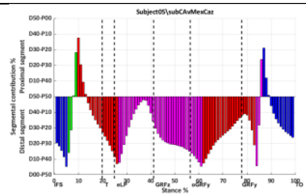
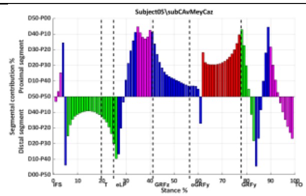
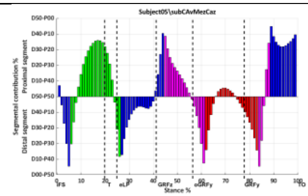
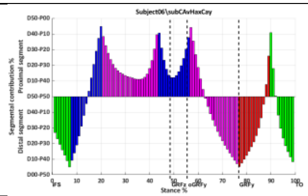
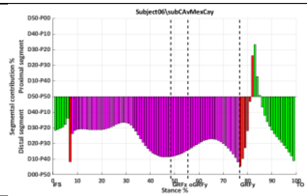
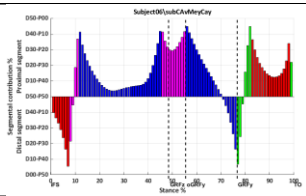
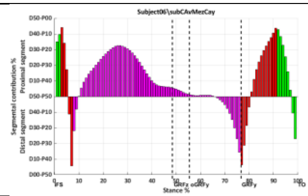
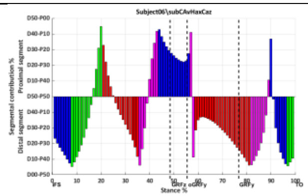
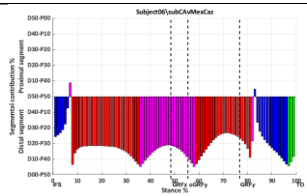
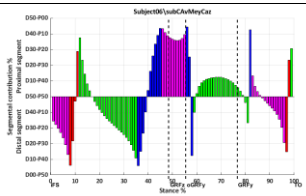
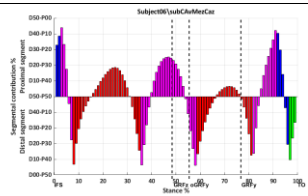
Athlete 01	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
Athlete 02	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				

IN-PHASE
Proximal segment +Δvelocity
Distal segment +Δvelocity

ANTI-PHASE
Proximal segment -Δvelocity
Distal segment +Δvelocity

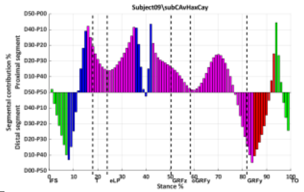
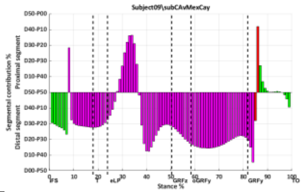
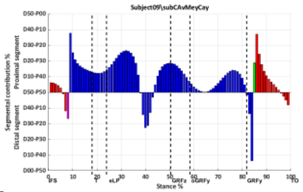
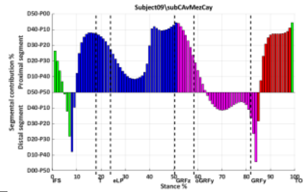
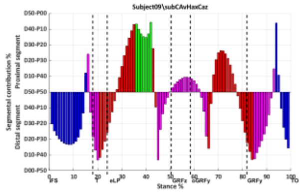
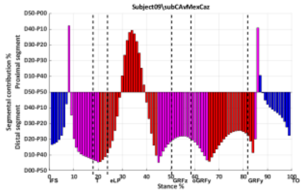
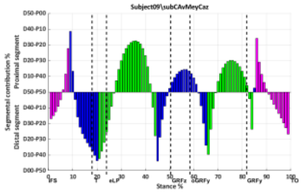
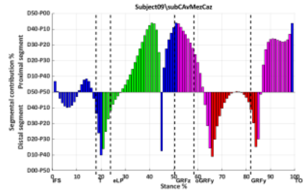
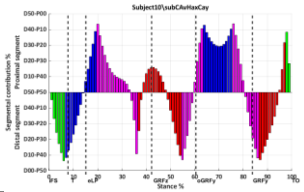
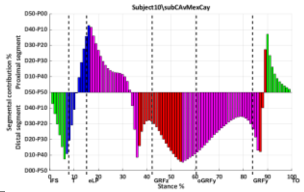
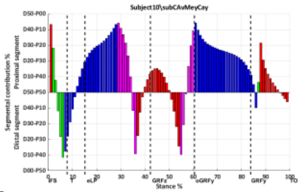
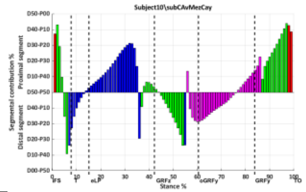
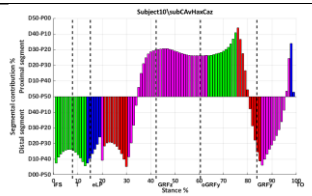
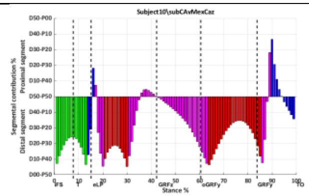
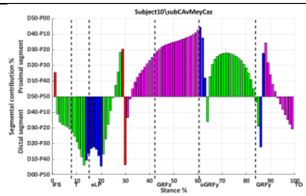
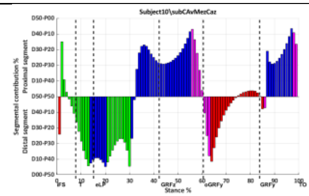
IN-PHASE
Proximal segment -Δvelocity
Distal segment -Δvelocity

ANTI-PHASE
Proximal segment +Δvelocity
Distal -Δvelocity

Athlete 03	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
Athlete 04	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				

IN-PHASE Proximal segment +Δvelocity Distal segment +Δvelocity	**ANTI-PHASE** Proximal segment -Δvelocity Distal segment +Δvelocity	**IN-PHASE** Proximal segment -Δvelocity Distal segment -Δvelocity	**ANTI-PHASE** Proximal segment +Δvelocity Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

Athlete 05	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
Athlete 06	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				

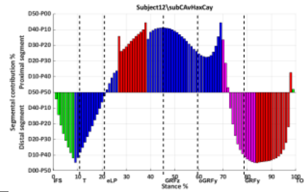
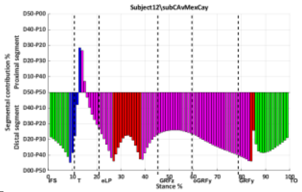
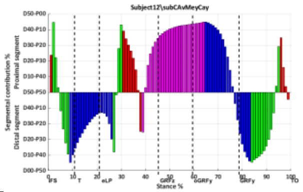
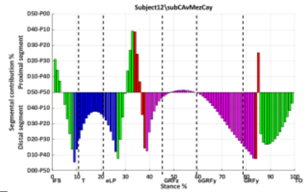
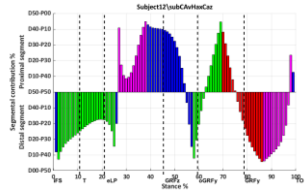
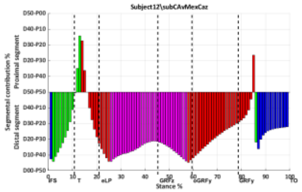
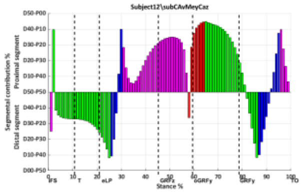
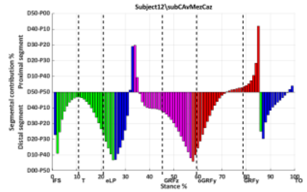
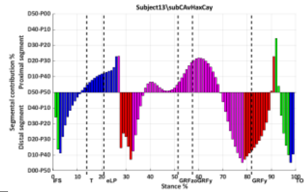
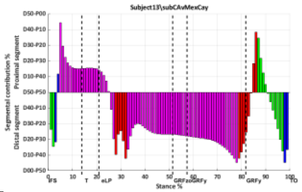
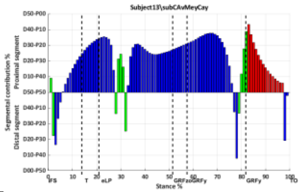
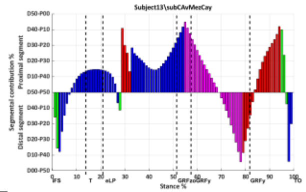
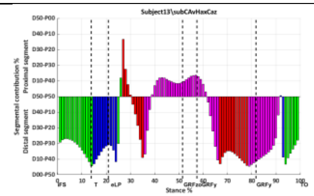
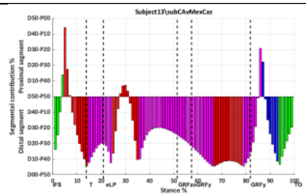
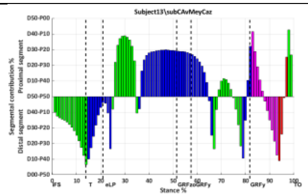
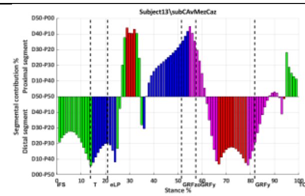
IN-PHASE
Proximal segment +Δvelocity
Distal segment +Δvelocity

ANTI-PHASE
Proximal segment -Δvelocity
Distal segment +Δvelocity

IN-PHASE
Proximal segment -Δvelocity
Distal segment -Δvelocity

ANTI-PHASE
Proximal segment +Δvelocity
Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

Athlete 07	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
Athlete 08	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				

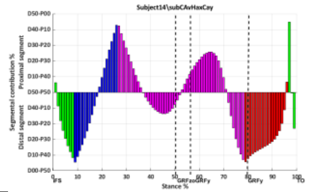
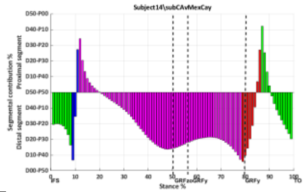
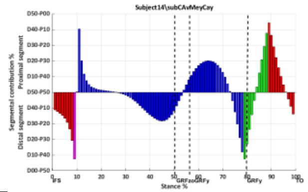
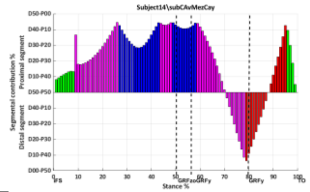
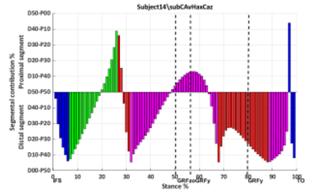
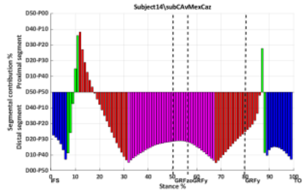
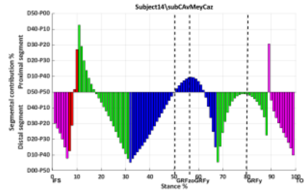
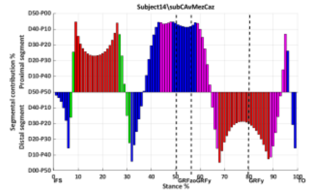
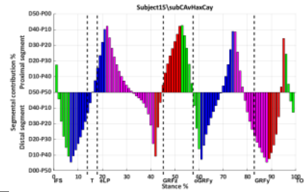
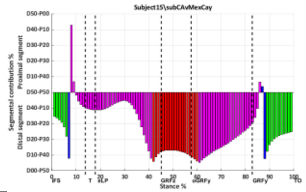
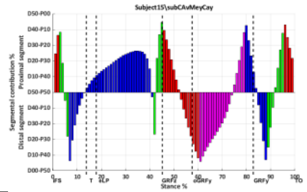
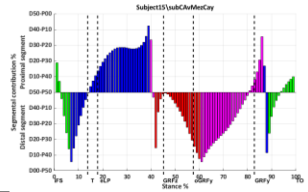
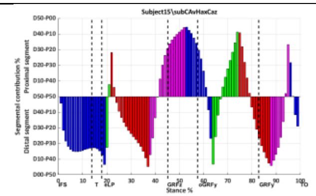
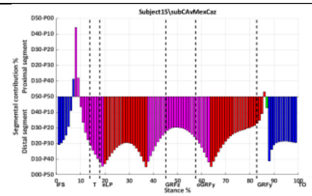
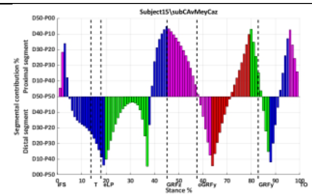
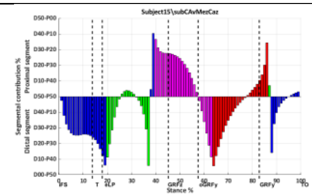
IN-PHASE
Proximal segment +Δvelocity
Distal segment +Δvelocity

ANTI-PHASE
Proximal segment -Δvelocity
Distal segment +Δvelocity

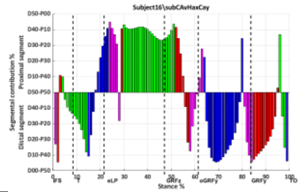
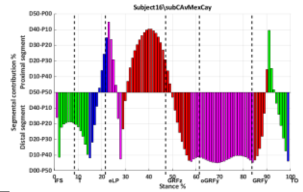
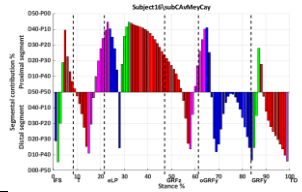
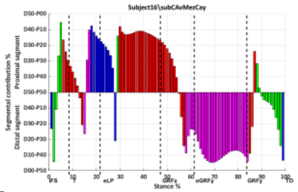
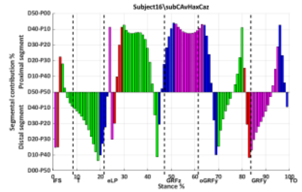
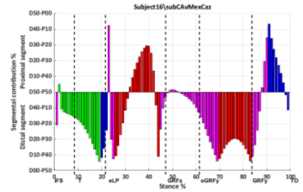
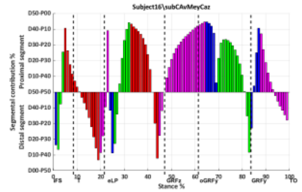
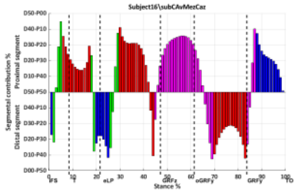
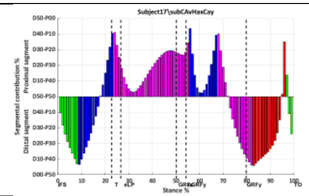
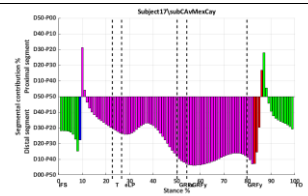
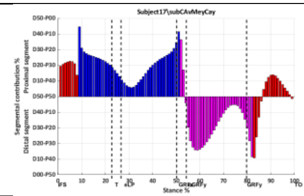
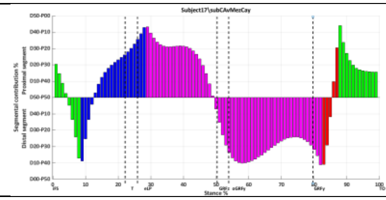
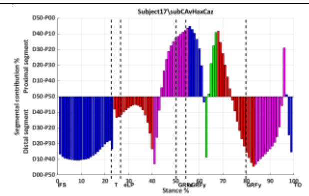
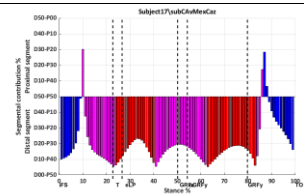
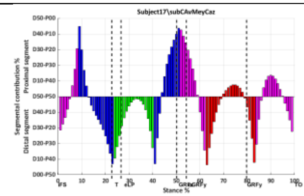
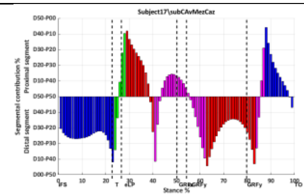
IN-PHASE
Proximal segment -Δvelocity
Distal segment -Δvelocity

ANTI-PHASE
Proximal segment +Δvelocity
Distal -Δvelocity

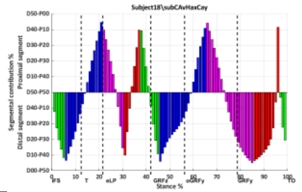
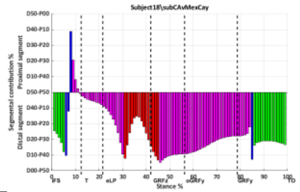
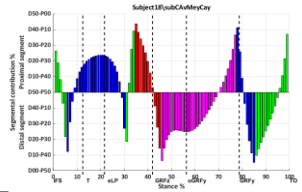
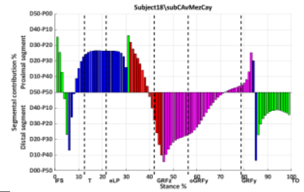
Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force.

Athlete 09	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
Athlete 10	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
	IN-PHASE Proximal segment +Δvelocity Distal segment +Δvelocity	**ANTI-PHASE** Proximal segment -Δvelocity Distal segment +Δvelocity	**IN-PHASE** Proximal segment -Δvelocity Distal segment -Δvelocity	**ANTI-PHASE** Proximal segment +Δvelocity Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

Athlete 11	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
Athlete 12	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
	IN-PHASE Proximal segment +Δvelocity Distal segment +Δvelocity	**ANTI-PHASE** Proximal segment -Δvelocity Distal segment +Δvelocity	**IN-PHASE** Proximal segment -Δvelocity Distal segment -Δvelocity	**ANTI-PHASE** Proximal segment +Δvelocity Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

Athlete 13	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
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	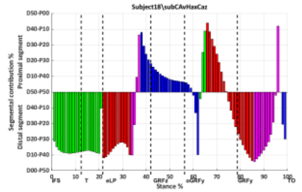	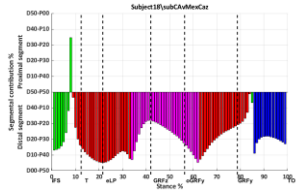	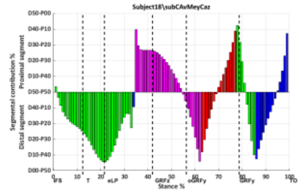	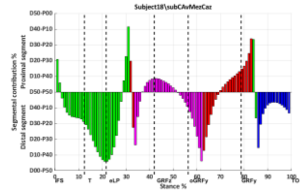
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	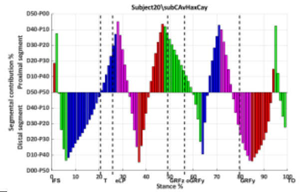	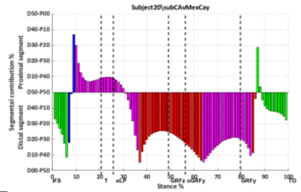	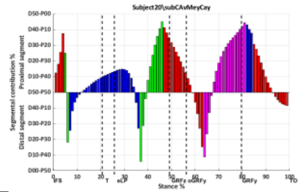	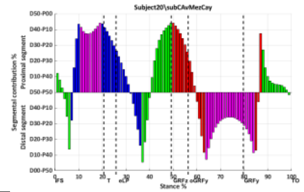
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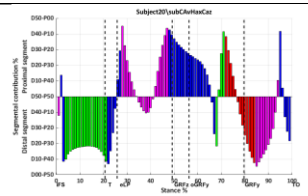
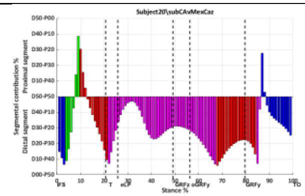
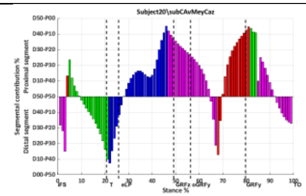
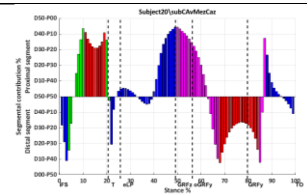
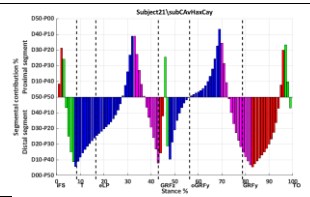
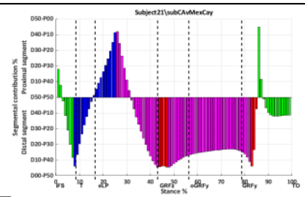
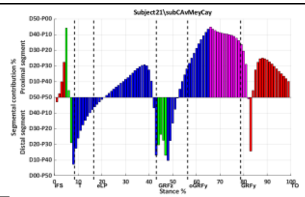
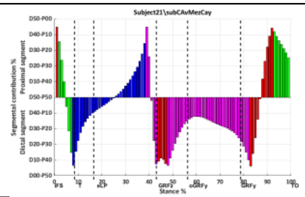
				
	IN-PHASE Proximal segment +Δvelocity Distal segment +Δvelocity	**ANTI-PHASE** Proximal segment -Δvelocity Distal segment +Δvelocity	**IN-PHASE** Proximal segment -Δvelocity Distal segment -Δvelocity	**ANTI-PHASE** Proximal segment +Δvelocity Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

Athlete 15	CalcaneusY -HalluxX	CalcaneusY -MetatarsalX	CalcaneusY -MetatarsalY	CalcaneusY -MetatarsalZ
				
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	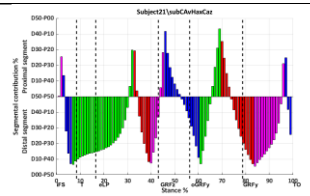	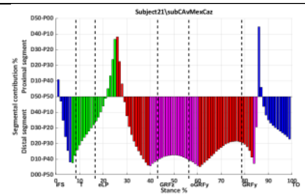	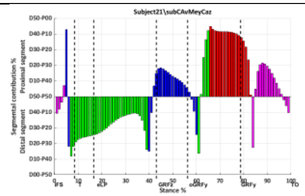	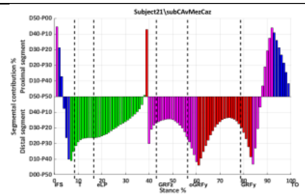
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	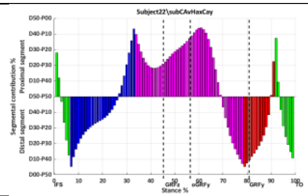	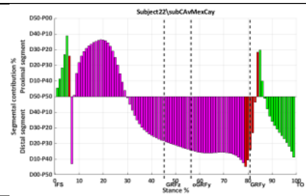	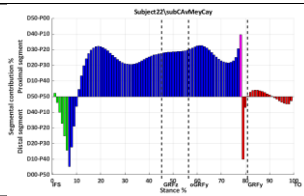	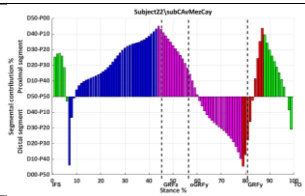
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	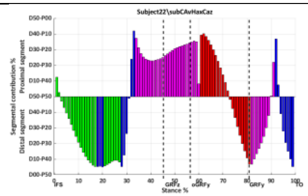	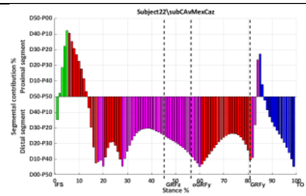	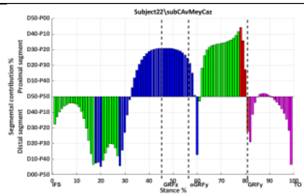	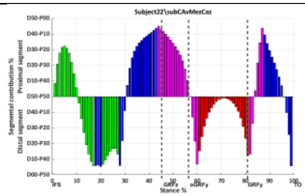

IN-PHASE
Proximal segment +Δvelocity
Distal segment +Δvelocity

ANTI-PHASE
Proximal segment -Δvelocity
Distal segment +Δvelocity

IN-PHASE
Proximal segment -Δvelocity
Distal segment -Δvelocity

ANTI-PHASE
Proximal segment +Δvelocity
Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

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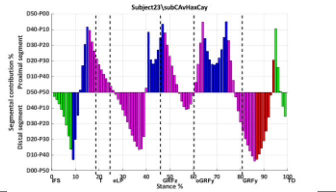
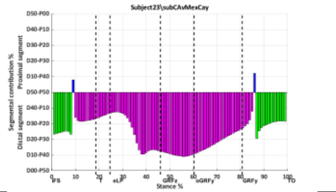
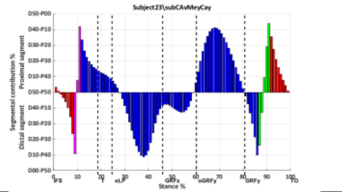
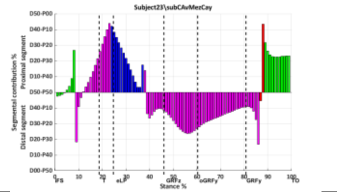
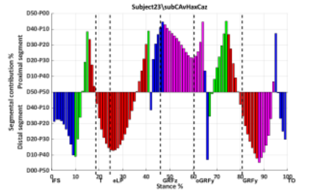
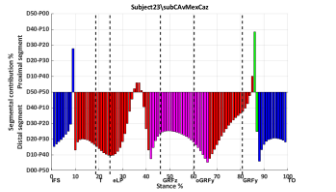
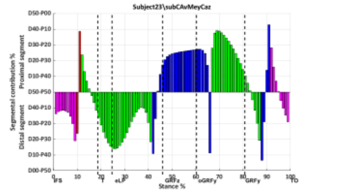
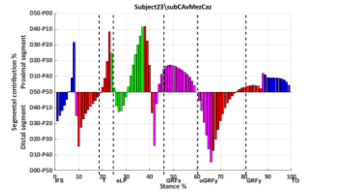
				
	CalcaneusZ- HalluxX	CalcaneusZ- MetatarsalX	CalcaneusZ- MetatarsalY	CalcaneusZ- MetatarsalZ
				
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	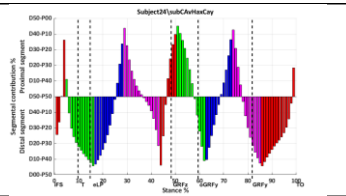	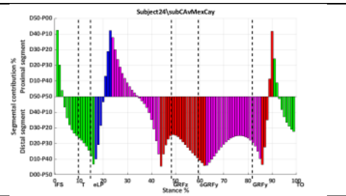	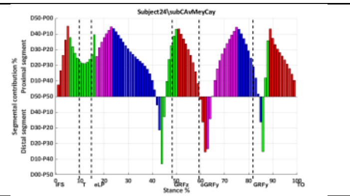	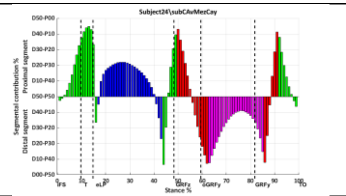
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	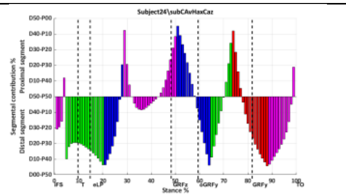	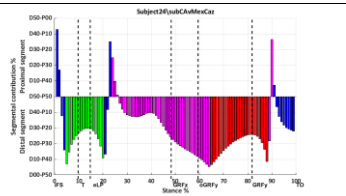	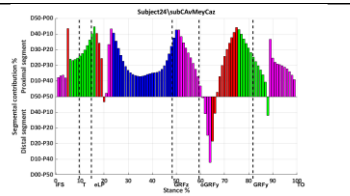	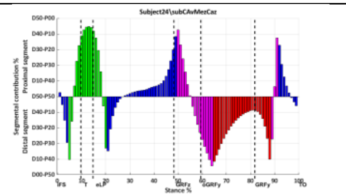

IN-PHASE
Proximal segment +Δvelocity
Distal segment +Δvelocity

ANTI-PHASE
Proximal segment -Δvelocity
Distal segment +Δvelocity

IN-PHASE
Proximal segment -Δvelocity
Distal segment -Δvelocity

ANTI-PHASE
Proximal segment +Δvelocity
Distal -Δvelocity

Notes. IFS = initial foot strike; T = transient; eLP = end of loading phase; GRFz = peak vertical ground reaction force; oGRFy = transition point from braking to propulsion phase in the anterior-posterior ground reaction force; peak anterior-posterior ground reaction force

APPENDIX D: BADMINTON REPORT

Funding for this project was received form Badminton World Federation

The role of the foot and a progressive foot muscle strengthening intervention programme on common ligament injury mechanism in females participating in court sports.

Carla van der Merwe[a], Sarah P. Shultz[b], Bob G. Robert Colborne[c], and Philip W. Fink[a]

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INTRODUCTION

Court games are characterised by rapid and repetitive short bursts of multiplanar accelerations and decelerations (Chandler, Pinder, Curran, & Gabbett, 2014; Drinkwater, 2008). Abrupt, unanticipated changes of direction put participants of court sports at an increased risk for sustaining non-contact anterior cruciate ligament ruptures (ACLR) and lateral ankle sprains (LAS) (Cassell, 2012; Doherty et al., 2014; Fong, Chan, Mok, Yung, & Chan, 2009a; McKay, Goldie, Payne, & Oakes, 2001).

In sports, 'high risk' movements increasing ACLR risk, is associated with a large knee valgus angle occurring in the presence of a relatively large explosive ground reaction force (GRF) (Powers, 2010). During a LAS an increased inversion ankle angle in the presence of an extended moment arm causes strain to the ligaments of the lateral ankle (Fong, Hong, et al., 2009b; Gehring, Wissler, Mornieux, & Gollhofer, 2013b). In such conditions the resultant GRF may not act through the ankle and/or knee joint centres of the stance leg, increasing the risk for lower leg ligament injuries (Beynnon, Renstrom, Alosa, Baumhauer, & Vacek, 2001; Boden, Torg, Knowles, & Hewett, 2009). Female court sport athletes are at an increased risk for sustaining a ACLR and LAS due to differences in hormonal, anatomical, neuromuscular and biomechanical characteristics compared to males competing in the same sport (Gould, Hooper, & Strauss, 2016; Shultz et al., 2015).

Injury prevention programmes play an increasingly important role in athlete preparation (Joseph & Finch, 2017; Meeuwisse, Tyreman, Hagel, & Emery, 2007). ACLR and LAS injury mechanism research site foot mechanics as part of the injury mechanism, but very few intervention programmes aim to change foot function directly (Gamada, 2014; Hewett et al., 2005; Shultz et al., 2015; Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005b). Several researchers suggest that one potential way to reduce the frequency and severity of these injuries would be to alter the foot function through strengthening the intrinsic and extrinsic muscles of the foot (Boden, Sheenan, Torg, & Hewett, 2010; Nigg, Baltich, Federolf, Manz, & Nigg, 2017). The aim of this project is therefore, to investigate the effect strengthening specific muscles acting on the foot will have on the injury risk factors associated with ACLR and lateral ankle sprain injuries.

METHODS

PARTICIPANTS

Eighteen female court sport athletes (age: 17.4 ± 1.49 years; height: 1.68 ± 0.04 m; mass: 64.53 ± 9.02 kg) gave written consent to take part in this study. All participants were experienced court sport athletes (netball, volleyball, and badminton), had sufficient skill to perform vertical jumps and sidestep cuts. All participants were injury and pain free the preceding six months and had no diagnosed musculoskeletal or neurological condition affecting movement or foot function. Ethical approval was sought and received from Massey University Human Ethics Committee.

INTERVENTION

The participants were randomly allocated to the training group (TG) or the control group (CG). The TG underwent an additional 16-week progressive foot muscle-strengthening programme. Foot muscle strengthening exercises were performed three times per week. The researcher supervised one exercise session per week. The exercises chosen aimed to strengthen both the intrinsic and extrinsic muscles acting on the foot.

DATA COLLECTION

MARKER PLACEMENT

Spherical retro-reflective markers were used to define joint axis and track the motion of the segments of the lower extremities. Markers (diameter 10 mm) were placed bilaterally on the anterior superior iliac spine, posterior superior iliac spine, greater trochanter, head of the fibula, tibial tuberosity, and medial and lateral femoral epicondyles. Rigid sets of 4 markers were also placed bilaterally on the lateral side of the thigh and shank. For data capturing in the barefoot (BFT) condition markers (diameter 8 mm) were placed on landmark sites according to the multi-segmental foot model as defined by Leardini et al. (2007a). In the shod (SHD) condition markers where placed according to model used by DiCesare et al. (2015). The landmark locations where captured during the static trial for each participant when the participant was in the anatomical position. The global coordination system was used defining the X-axis (sagittal) as orientated medial to lateral, Y-axis (frontal) orientated posterior to anterior and the Z-axis (horizontal) orientated vertically distal to proximal. Kinematic and kinetic data were concurrently collected from the dominant leg by a ten camera (six Oqus 3-series and four Oqus 7-series) Qualisys system (Qualisys, Gothenburg, Sweden) sampling at 200Hz and a force platform (Kistler, 569x DAQ, Winterhur, Switzerland) sampling at 1600Hz.

Experiment protocol

SHD and BFT condition was randomised. Participants underwent a familiarisation period prior to any data collected by running the length of the marked runways. During the familiarisation, the participants were guided to achieve the desired approach speed (4.5 ms^{-1} 0.47 m.s^{-1}) for the cutting movements in each footwear condition. The jumps were performed after the familiarisation and before the cutting task.

Counter movement jump

Participants were required to perform counter movement vertical jumps (hands on hips) (CMJ) taking off from and landing on both feet on the force platform. The force platform was used to record GRF at take-off and landing during the CMJ. The landing phase was recorded from > 20 N vertical GRF. Lower limb kinematics were recorded at maximum vertical GRF when landing from the CMJ.

Cut

The participants performed 45° (from direction of progression) unanticipated cutting tasks on their dominant leg. Participants were required to perform five successful trials. A trial was deemed successful if the approach run was within 10 % of the desired 4.5ms^{-1} approach speed, the force plate was struck with the entire dominant foot and the cut was performed towards the indicated direction. Three electronic timing gates (Smart Speed, Fusion Sport, Queensland, Australia) were placed 5 m away and on the opposite side of the force plate. One set of electronic speed gates was placed straight ahead; another timing gate were placed on a 45° angle to the left and another to the right from the approach direction (Figure 1). The SmartSpeed traffic light system was pre-programmed to indicate a random direction (left, right or straight ahead) and flashed 0.7 sec after the approach run was initiated. The approach speed was monitored by an additional photocell timing system (Brower Timing System, USA).

Data processing

Visual 3D^{TM} (v5.0, C-Motion, USA) and MATLAB (R2017a, Mathworks, MA USA) were used to calculate height, knee, and ankle frontal plane angles at maximum GRF impact at the time of landing from the CMJ. Kinetic and kinematic data were processed to calculate approach speed, deceleration, stance time, frontal- and sagittal plane knee moment arms and angles, frontal plane ankle moment arms and angles during the unanticipated cutting task. Kinematic data for the cutting task was recorded at transient impact force (if present) and peak vertical GRF. The vertical ground reaction force at 20 N threshold identified the stance phase. All data were time scaled and normalized to 100 % of the stance phase. Kinetic and kinematic data were processed in Visual 3D™ (v5.0, C-Motion, USA). Marker trajectories were interpolated for missing signals and smoothed using a sixth-order low-pass Butterworth filter with a 10Hz cut-off frequency. Ground reaction forces were filtered using a sixth-order low-pass Butterworth filter with a cut-off frequency of 12 Hz.

OUTCOMES MEASURES

PERFORMANCE AND KINETIC VARIABLES

Performance outcomes included the counter movement jump height. For the change of direction task approach speed and stance time will be considered (McLean, Lipfert, & Van Den Bogert, 2004; Spiteri et al., 2015).

KINEMATIC VARIABLES

LAS injury risk factor variables were the maximums of the ankle inversion angle (Panagiotakis et al., 2017) and maximum lengths of the inversion moment arms (Fong, Chan, Mok, Yung, & Chan, 2009c) at transient and peak GRF. ACLR injury risk factor variables were the maximum of the knee valgus angle and the length of the valgus moment arm (Stuelcken, Mellifont, Gorman, & Sayers, 2015) at transient and peak GRF.

RESULTS

PERFORMANCE VARIABLES

Athlete's performed significantly better shod compared to barefoot in jump height ($F(1, 9) = 16.8$, $p < .05$) and approach speed of the 45°cutting task ($F(1, 9) = 11.31$, $p < .05$). In contrast, the stance time during the 45°cutting task was significantly shorter in the barefoot condition compared to shod ($F(1, 9) = 20.4$, $p < .05$).

The TG's acceleration performance through the 45°cutting task improved. The TG had a smaller decrease in acceleration in both barefoot (CG decrease 59 %, TG decrease 11 %) and shod conditions (CG decrease 44 %, TG decreased 17 %) compared to the CG.

KINEMATIC VARIABLES

JUMP

Kinematics for the dominant lower limb was recorded at the highest vertical ground reaction force during the landing phase of the jump. The knee valgus angle displayed a significantly larger increase from pre-intervention to post-intervention in the shod condition than the increase observed in the barefoot condition, $F(1, 9) = 9.06$, $p < .05$.

The ankle inversion angle was significantly larger ($F(1, 9) = 135$, $p < .0001$) in the barefoot condition compared to the shod condition (which displayed an average eversion angle). The average ankle inversion angle was also significantly larger ($F(1, 9) = 6.56$, $p < .05$) post-intervention compared to pre-intervention. There was a significant difference ($F(1, 9 = 14.43$, $p < .05$) in the change of the ankle inversion angle between footwear conditions from pre- to post-intervention.

45° CUTTING TASK

ANKLE FRONTAL PLANE

Lower limb kinematics was recorded during the transient and peak vertical GRF incidences during the 45° cutting movement. The external ankle inversion moment arm was significantly larger in the shod condition compared to barefoot condition at transient ($F(1, 9) = 167.7$, $p < .001$) and at peak ($F(1, 9) = 98.57$, $p < .0001$)). The ankle inversion moment arm at the transient was also significantly different ($F(1, 9) = 8.95$ $p < .05$,)) post- compared to pre-intervention between footwear conditions, the shod condition having larger decreases compared to the barefoot condition.

For peak inversion moment arms, the TG had a significantly large decrease (128 %) post-intervention while barefoot, $F(1, 7) = 8.83$, $p < .05$. In contrast to the external inversion ankle moment arm, the ankle inversion angle was significantly larger in the barefoot condition than the shod condition at both points measured (Transient: $F(1, 7) = 63.06$, $p < .0001$, Peak: $F(1, 7) = 129.79$, $p < .001$). The ankle inversion angle was also significantly larger post-intervention for both groups at Transient, $F(1, 9) = 16.91$, $p < .05$.

KNEE FRONTAL PLANE

The barefoot conditions presented with a significantly larger external knee varus moment arm than shod condition $F(1, 9) = 38.78$, $p < .05$ at transient. The average external knee frontal plane moment arm at transient was also significantly larger post-intervention compared to pre-intervention, ($F(1, 9) = 9.84$, $p < .05$,). Both groups presented with significantly smaller knee valgus angles while barefoot than shod at both transient ($F(1, 9) = 13.81$, $p < .05$) and peak ($F(1, 9) = 22.87$, $p < .05$) time points.

TABLE 1: Changes to performance from hands-on-hips vertical jump and 45° cutting movement.

		BAREFOOT						SHOD						Significant interactions				
		Control Group (n=10)			**Training Group (n=8)**			**Control Group (n=10)**			**Training Group (n=8)**							
		PreI	**PostI**	**% ↑↓**	***PreI***	**PostI**	**↑↓%**	***PreI***	**PostI**	**↑↓%**	***PreI***	**PostI**	**↑↓%**	**C**	**S**	**SG**	**SC**	**SGC**
Counter movement jump																		
Jump Height (cm)	Mean	*20.29*	21.44	↑ 5%	*23.13*	23.38	↑ 1%	*21.24*	23.18	↑ 8%	*25.98*	25.05	↓ 4%	*				
	SD	5.14	5.42		6.76	2.66		5.65	4.83		5.99	4.65						
45°cutting movement																		
Approach Speed (m/s)	Mean	*3.73*	4.36	↑ 17%	*3.77*	4.29	↑ 14%	*3.88*	4.39	↑ 13%	*3.90*	4.37	↑ 12%	*	#			
	SD	0.33	0.28		0.24	0.37		0.30	0.34		0.34	0.40						
Stance Time (sec)	Mean	*0.23*	0.21	↓ 10%	*0.20*	0.19	↓ 9%	*0.24*	0.23	↓ 7%	*0.22*	0.21	↓ 6%	*	*			
	SD	0.02	0.02		0.02	0.01		0.02	0.02		0.02	0.02						
Acceleration (m/s^2)	Mean	*1.96*	0.81	↓ 59%	*2.65*	2.35	↓ 11%	*1.35*	0.76	↓ 44%	*2.60*	2.17	↓ 17%		*			
	SD	1.37	1.83		1.01	2.24		1.33	1.89		1.78	2.39						

**p < .05, #p < .0001; ↑↓=increase/decrease in performance variable; Significant interactions: C = condition, S = session, SvG = session vs group, SvC = session vs condition, SvGvC = session vs group vs condition.*

TABLE 2: Changes to kinematic variables from vertical counter movement jump.

		BAREFOOT						SHOD						**Significant interactions**				
		Control Group (n=10)			**Training Group (n=8)**			**Control Group (n=10)**			**Training Group (n=8)**							
		PreI	**PostI**	**% ↑↓**	**PreI**	**PostI**	**↑↓%**	**PreI**	**PostI**	**↑↓%**	**PreI**	**PostI**	**↑↓%**	**C**	**S**	**SG**	**SC**	**SGC**
Jump Kinetics (dominant leg)																		
Knee Valgus Angle (°)	Mean	-2.70	-2.67	↓ 1%	-1.81	-2.98	↑ 65%	-2.30	-5.14	↑ 123%	-2.42	-5.00	↑ 107%				*	
	SD	3.89	4.95		4.66	4.18		4.74	4.61		5.29	6.76						
Ankle Inversion Angle (°)	Mean	4.65	8.16	↑ 76%	3.48	8.93	↑ 157%	-2.20	-1.75	↑ 20%	-1.73	-0.50	↑ 71%	#	*		*	
	SD	4.20	5.50		5.87	4.26		4.08	4.43		5.29	6.76						

**p < .05, #p < .0001; ↑↓=increase/decrease in variable value; Significant interactions: C = condition, S = session, G = group, SvG = session vs group, SvC = session vs condition, GvC = group vs condition, SvGvC = session vs group vs condition.*

TABLE 3: Changes to ankle frontal plane kinetics during 45° cutting movement.

		BAREFOOT						SHOD						Significant interactions				
		Control Group (n=10)			Training Group (n=8)			Control Group (n=10)			Training Group (n=8)							
		PreI	**PostI**	**% ↑↓**	**PreI**	**PostI**	**↑↓%**	**PreI**	**PostI**	**↑↓%**	**PreI**	**PostI**	**↑↓%**	**C**	**S**	**SG**	**SC**	**SGC**
Ankle external inversion moment arm (cm) (+ = inversion/ - = external)																		
Transient	Mean	0.68	0.89	↑ 29%	0.58	0.37	↓ 37%	1.97	1.51	↓ 23%	1.83	1.61	↓ 12%	#			*	
	SD	0.68	0.74		0.34	0.62		0.72	0.64		0.69	0.86						
Peak	Mean	0.32	0.57	↑ 78%	0.25	-0.07	↓ 128%	1.43	1.23	↓ 14%	0.97	1.03	↑ 6%	#				*
	SD	0.66	0.67		0.52	0.64		0.71	0.63		0.65	0.67						
Ankle inversion angle (◦) (+ = inversion/ - = eversion)																		
Transient	Mean	19.42	25.85	↑ 33%	19.86	25.09	↑ 26%	17.67	18.22	↑ 3%	16.07	18.44	↑ 15%	*	*			
	SD	.68	0.74		0.34	0.62		4.47	3.86		3.08	4.39						
Peak	Mean	14.99	19.17	↑ 28%	17.99	21.35	↑ 19%	10.38	11.38	↑ 10%	9.78	13.42	↑ 37%	*				
	SD	4.32	6.50		6.80	4.46		5.34	4.73		5.89	2.88						

**p < .05, #p < .0001; ↑↓=increase/decrease in variable value; Significant interactions: C = condition, S = session, SG = session vs group, SC = session vs condition, SGC = session vs group vs condition.*

TABLE 4: Changes to knee frontal plane knee kinematics during 45° cutting movement.

		BAREFOOT						SHOD						Significant interactions				
		Control Group (n=10)			**Training Group (n=8)**			**Control Group (n=10)**			**Training Group (n=8)**							
		PreI	**PostI**	**% ↑↓**	**PreI**	**PostI**	**↑↓%**	**PreI**	**PostI**	**↑↓%**	**PreI**	**PostI**	**↑↓%**	**C**	**S**	**SG**	**SC**	**SGC**
Knee external valgus moment arm (cm). (+ = varus moment arm/ - = valgus moment arm)																		
Transient	Mean	1.21	-1.95	↑ 261%	0.81	-0.94	↑ 217%	1.85	0.95	↑ 48%	2.12	1.71	↑ 19%	*	*			
	SD	2.16	2.87		2.37	3.70		1.67	2.46		1.79	2.65						
Peak	Mean	4.49	3.60	↑ 20%	3.50	3.25	↑ 7%	4.47	4.23	↑ 5%	3.80	3.07	↑ 19%					
	SD	1.13	2.26		1.69	1.84		1.83	1.88		1.98	1.95						
Knee valgus angle (°). (+ = varus angle/ - = valgus angle)																		
Transient	Mean	-3.30	-1.98	↓ 40%	-3.84	-2.96	↓ 23%	-3.68	-5.85	↑ 59%	-6.98	-6.80	↓ 3%	*				
	SD	3.77	4.08		4.21	4.81		4.34	4.07		3.86	5.16						
Peak	Mean	-3.64	-6.22	↑ 71%	-5.79	-5.44	↓ 6%	-7.19	-9.54	↑ 33%	-7.40	-9.34	↑ 26%	*				
	SD	3.94	4.41		3.39	4.50		5.04	3.44		5.58	4.59						

**p* < .05, #*p* < .0001; ↑↓=increase/decrease in variable value; Significant interactions: C = condition, S = session, SG = session vs group, SC = session vs condition, SGC = session vs group vs condition.*

Discussion

A summary of the significant changes in performance and risk factors as a result of intervention and of each footwear condition is displayed in Table 5.

Performance

In this study general sport specific training and conditioning had a larger effect on approach velocity on the 45° cutting task than intervention training and footwear condition as both groups displayed better post-intervention velocities in both footwear conditions. Footwear condition did seem to influence the CMJ height, and approach velocity of the cutting tasks. Both jump height and approach velocity were significantly improved when shod compared to barefoot. In contrast, the stance time and acceleration through the stance time of the 45° cutting task was improved when barefoot. Thus, no clear recommendation with regards to footwear condition to enhance athletic performance of females during court sport activities can be made.

It is interesting to note that although both groups had improved approach speed the acceleration through the stance phase of the cutting task was worse for both groups at the post-intervention test. However, the athletes that underwent the strength training intervention had a smaller loss in acceleration in both footwear conditions possibly indicating the benefit of strengthening the muscles acting on the foot.

Kinematic outcome variables

The kinematic outcome variables recorded during the CMJ as they relate to ACL and LAS injuries were mostly reduced when performed barefoot. When landing barefoot from a jump the knee valgus angle of the shank was significantly smaller than when landing shod possibly reducing ACL injury risk when barefoot. Ankle eversion (a possible indication of rearfoot eversion) is associated with larger knee valgus angles (Holland, 2016; Kagaya, Fuji, & Nishizono, 2013; McLean et al., 2004). In contrast, larger inversion angles were observed during barefoot landing, possibly increasing the risk for LAS injury during CMJ landing. It can therefore be inferred that the risk to ACL injuries may be reduced when performing CMJ barefoot or in a shoe that mimics the barefoot condition. Although, there might be an increased risk for LAS injury when performing CMJ barefoot compared to shod.

During the 45° barefoot cutting task the external ankle inversion moment arm was shorter at both transient and peak time points. The resultant GRF thus passed closer to the ankle joint centre compared to the shod tasks, possibly decreasing the risk for LAS injury in the barefoot condition. At transient, an increase in the external inversion moment arm for the barefoot condition was observed post-intervention. However, the length of the moment arm was still smaller compared to the shod condition at this timepoint, maintaining a lower risk for LAS in the barefoot condition. Furthermore, at peak the TG had a significantly larger decrease in the external inversion moment arm for the barefoot condition compared to the barefoot condition of the CG post-intervention. The transient and peak ankle inversion angles was larger barefoot and may therefore increase the risk for LAS. However, the increased moment arm length observed shod holds the greater risk for LAS incidence (Fong, Chan, et al., 2009c; Fong, Hong, et al., 2009b) . Thus, undergoing a foot musculature strength training programme and being barefoot may decrease the risk for LAS injury during unanticipated change of direction tasks.

In the barefoot condition the average transient external knee valgus moment arm was significantly larger than shod. However, the length of the valgus moment arm observed at transient in the barefoot condition was smaller than the valgus moment arm in the shod condition. Thus, putting the resultant GRF closer to the knee joint centre. Similarly, the peak external valgus moment arm barefoot was shorter than the moment arm in the shod condition. The shorter moment in the barefoot condition decreases the risk for ACL injury at both time points. Furthermore, the knee valgus angles were also smaller in the barefoot condition at both time points, also decreasing ACL injury risk. The intervention training also seems to have limited the increase in the external knee frontal plane moment arm length decreasing ACL injury risk further.

TABLE 5: The outcome variables that where influenced by the intervention and/or footwear condition.

Barefoot	Shod	Intervention	‘Normal’ training
Counter Movement Jump			
Smaller knee valgus angle reduces risk of ACL injury	Greater jump height		
	Smaller ankle inversion angle, reduces LAS risk		
45°cutting movement			
Shorter stance time increases performance	Faster approach speed increases performance		
Smaller decrease in acceleration through stance increases performance			
Smaller ankle frontal plane moment arm – resultant GRF closer to ankle joint centre at Transient and Peak - decrease risk for LAS	Smaller ankle inversion angle – decrease risk for LAS	Larger decrease in ankle external eversion moment arm length (barefoot) – decrease LAS risk	
Knee frontal plane moment arm closer to knee joint centre – decreases risk for ACL injury			
Smaller Knee valgus angle – decreases risk for ACL injury			

CONCLUSION

ACL and LAS injury risk was reduced when CMJ and 45° cutting tasks was performed barefoot. The risk to ACL and LAS injury seems to be further reduced for the athletes who performed the strengthening exercises. Strengthening the muscles acting on the foot and performing court sport activities barefoot or in a barefoot type shoe may therefore be recommended to females. No clear recommendation can be made about footwear regards to performance enhancement as results were varied. Equally, strengthening the muscles acting on the foot had no clear statistically significant regards to performance enhancement.

It should be made clear that although this experiment may have found a reduced risk for ACL and LAS in barefoot or barefoot type shoe wear for females participating in court sport, the tasks were conducted in a controlled laboratory. Research is needed in a 'real world' setting to be able to obtain conclusive evidence regarding the ability to reduce injury risk without causing other injury. Prudent care should thus be taken when introducing barefoot/ barefoot like shoe wear into court sport activities. Should an athlete decide to take advantage of the benefits associated with adopting barefoot/ barefoot like shoe play and/or training, a gradual footwear change complimented by strength training for the muscles acting on the foot is strongly recommended.

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APPENDIX E: INFORMATION AND QUESTIONNAIRES

COACH'S INFORMATION AND PERMISSION

MASSEY UNIVERSITY
COLLEGE OF HEALTH
TE KURA HAUORA TANGATA

Foot strengthening and injury prevention

Coach's Information & Permission

The team(s) you are coaching and/or managing are invited to take part in a university research project that will investigate the effect that a foot strengthening intervention protocol will have on the injurious movements patterns associated with common injuries in females that participate in court sports like netball, badminton, basketball, volleyball and tennis. The research is being conducted by the School of Sport and Exercise at Massey University. The contact details are below:

Principal Investigator: PhD candidate Mrs Carla van der Merwe Email: c.vandermerwe@massey.ac.nz Phone: 06 951 6817 ext. 83817	Primary supervisor: Dr Philip Fink Email: P.Fink@massey.ac.na Phone: 06 951 7330 ext. 84530

Why is this research important?

Injuries resulting from sports participation can lead to hospitalisation and incur substantial costs. The loss of practice and game time as a result of an injury can influence the athlete's performance, and in addition, rehabilitation can be lengthy and complicated. Injured athletes often end up with long-term dysfunction, which results in impaired performance, the decline in sports activity and performance or altogether termination of sports participation. It is, therefore, important to employ strategies to eliminate or, at least, decrease the occurrence and severity of these injuries.

Most conditioning programs do include a component of injury prevention. The current New Zealand SportSmart programme is an example of such a programme geared towards preventing ankle and knee injuries. However, although foot and ankle mechanics are cited in both ankle and knee injury mechanisms, very few injury prevention programs aim to improve the strength and function of the muscles controlling the foot. Researchers suggest that a strong foot has an important role to play in safe ankle, knee and hip movements. This study aims to investigate the influence that strengthening the foot muscles will have on ACL and lateral ankle sprain injury mechanisms.

Page 1

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MASSEY UNIVERSITY
COLLEGE OF HEALTH
TE KURA HAUORA TANGATA

Who are we looking for?

We aim to evaluate females aged 16 to 25 years or old who are currently (and primarily) playing for a team competing in any court sports (netball, badminton, basketball, volleyball, etc.). She will be eligible to take part in this study if she doesn't have a current lower extremity injury nor had had an injury or surgery within the previous 6 months that required her to be absent from participating in a sporting activity for more than 4 weeks. In addition, she should not have been diagnosed to have any foot deformity/abnormality by a clinician.

What is going to happen?

If the athlete decides to take part in this study, she will be asked to attend testing occasions may be asked to take part in an intervention programme that will last 12 weeks. She will be placed in 1 of 2 groups. The 1st group will be the intervention group and this group will perform a set of foot specific exercises in addition to all other normal training. The 2nd group will act as the control group. The 2nd group will not perform any additional, foot specific exercises.

Both groups will be tested before the start of the season, and then again after an intervention period of 12 weeks. The intervention group then will be tested again before the start of the next playing season. Testing will take place at Massey University's LATU facility (see Map) where different types of data will be collected (see Table 1).

The participants placed in the intervention study will be required to do 3x20min sessions consisting of exercises aimed at strengthening the muscles acting on the foot. Most of the session will be home based. Regular supervised session will be arranged to ensure correct execution of exercises. Both groups will receive a pair of court sports shoes, which should be worn during regular court sport training sessions and matches. Athletes will be required to keep a dairy of the number of sessions completed per week (intervention group only) and the amount of time spend wearing the provided shoes. Information about her injury history from the previous year, until the end of the study, will also be collected.

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TE KURA HAUORA TANGATA

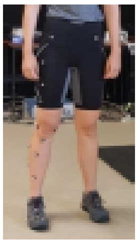
Testing sessions

Anthropometrics

Height and Weight will be measured.

3D Motion Capture

We will attach reflective markers (small silver balls) to her feet, legs and hips. We will use these markers to record the movements of her limbs in 3D while stepping onto a force plate when she executes jumping and agility movements. She will perform these movements twice. Once while barefoot and also in shoes that will be provided. She will be familiarised with the different jumping and agility movements that she will perform before any data is recorded.

(a) 

(b) 

Figure 1: Example of reflective markers placed while wearing shoes (a) and barefoot (b)

Jumps

She will perform a vertical jump for height. She will step with both feet on the force plate, jump up as high as she can and land back on the force plate with both feet. The force platform is embedded in the floor and will record ground reaction forces. We will record 5 attempts. This whole sequence will be repeated while wearing sports shoes and while being barefoot.

Change of direction

During the change of direction tasks, she will be asked to run 5m towards the force plate and react to a set of flashing lights. She will be required to keep her approach speed similar to what she would normally do in a game situation. The lights will indicate whether

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she should turn 45° to the left, 45° to the right or continue with a forward straight run. This whole sequence will be repeated while wearing shoes and while being barefoot.

Testing will take approximately 2 hours.

Clothing requirements

She will need to wear short tights or compression shorts without any reflective strips during the testing sessions. The reflective markers need to be attached to the skin or tight-fitting clothes in order to capture true limb movements with as little as possible anomalies. Regular clothes that are not skin-tight or have reflective strips may hide or interfere with the markers and render the data captured ineffectual and should not be worn. The markers will be attached with double sided tape and will be held down with additional sports strapping tape. These adhesives do not cause damage to the skin or clothes and are widely used in biomechanical and sporting environments.

Shoes will be provided for all testing sessions and treated with disinfected shoe spray to ensure high standards of sanitation.

What are the benefits of taking part in this study?

She will be able to receive a report *on request* with her results after the 2nd testing session to indicate her progress throughout the season. This data will not necessarily indicate whether she will sustain an injury, as injuries occur due to multiple factors. It will, however, indicate whether she have improved her jumping, speed, agility and deceleration abilities during the season. It will also indicate the changes in the foot, ankle, knee and hip movements associated with ACL and lateral ankle sprain injuries during the playing season. She would have contributed to the sport scientific community by helping us to understand if and how the body adapts to the training of the foot muscles and whether we can help prevent future traumatic events and keep New Zealand's females safe and on the sports field.

What will happen to this information?

All the information that the researchers collect will be kept on a computer. Her details and results will remain confidential, and her name will not be used at any time during the study. Only the researcher and supervisor will have access to the data. We may use the data that we collect in publications or during presentations, but no one will be able to tell which data is hers. If she requests her results, her report will only be sent to herself.

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Participant's rights.

She is under no obligation to accept this invitation. If she decides to participate, she has the right to:

- decline to answer any particular question;
- withdraw from the study at any time *(if she choose to withdraw she cannot withdraw her data from the analysis after the data collection has been completed)*;
- ask any questions about the study at any time during participation;
- provide information on the understanding that her name will not be used, unless she gives permission to the researcher;
- be given access to a summary of the project findings when it is concluded
- request a report of her individual results

What is the next step?

If she has any questions, she can ask any member of the research team at any time. If she has read and understood everything and she is happy to take part, please contact the principal investigator. She will then be asked to complete 'The Health and Injury History Recruitment Questionnaire'. Details in the health-screening questionnaire will help determine if she is eligible to participate

Project contacts

If she has any further questions or concerns about the project, either now or in the future, please contact either Mrs Carla van der Merwe or the primary supervisor (details on page 1).

MUHEC application

This project has been reviewed and approved by the Massey University Human Ethics Committee: Southern A, Application 16/77. If she has any concerns about the conduct of this research, please contact Mr Jeremy Hubbard, Chair, Massey University Human Ethics Committee: Southern A, telephone 04 801 5799 x 63487, email humanethicsouthA@massey.ac.nz.

Compensation for Injury

If any physical injury results from her participation during the testing sessions of this study, she should visit a treatment provider to make a claim to ACC as soon as possible. ACC cover and entitlements are not automatic and her claim will be assessed by ACC in accordance with the Accident Compensation Act 2001. If her claim is accepted, ACC must inform her of her entitlements and must help her access those entitlements. Entitlements may include, but not be limited to, treatment costs, travel costs for

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rehabilitation, loss of earnings, and/or lump sum for permanent impairment. Compensation for mental trauma may also be included, but only if this is incurred as a result of physical injury.

If her ACC claim is not accepted, she should immediately contact the researcher. The researcher will initiate processes to ensure she receives compensation equivalent to that to which she would have been entitled had ACC accepted her claim.

I have read and understood that the information contained in this document will be communicated to athletes I coach/manage and that athlete participation is voluntary.

Coach/Manager name: ____

Signature: ______________________________ Date: ______________

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Table 1: Time line between intervention and testing

2018						2019
January	February	March	April	May	June	Jan
Pre-season Testing for both groups	*Pre-season Testing for both groups*	**Intervention exercises performed by Intervention group (Group 1) only**			*Post-intervention testing for both groups*	*Follow-up Post-post-intervention testing for Intervention Group only*

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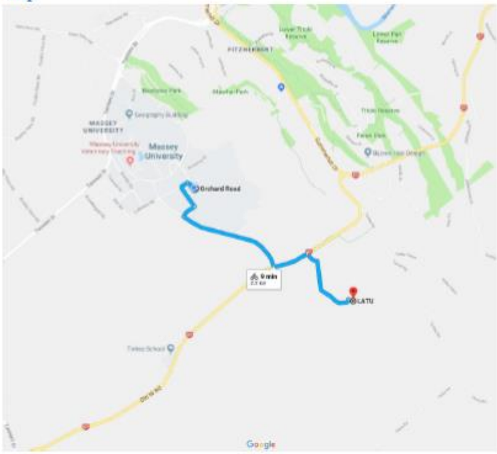
Map



Directions from Massey University campus to LATU (-40.397043, 175.636801)

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PARTICIPANT'S INFORMATION

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Foot strengthening and injury prevention

Participant's Information

You are invited to take part in a university research project that will investigate the effect that a foot strengthening intervention protocol will have on the injurious movements patterns associated with common injuries in females that participate in court sports like netball, badminton, basketball, volleyball and tennis. The research is being conducted by the School of Sport and Exercise at Massey University. The contact details are below:

Principal Investigator: PhD candidate Mrs Carla van der Merwe Email: c.vandermerwe@massey.ac.nz Phone: 06 951 6817 ext. 83817	Primary supervisor: Dr Philip Fink Email: P.Fink@massey.ac.nz Phone: 06 951 7530 ext. 84530

Why is this research important?

Injuries resulting from sports participation can lead to hospitalisation and incur substantial costs. The loss of practice and game time as a result of an injury can influence the athlete's performance, and in addition, rehabilitation can be lengthy and complicated. Injured athletes often end up with long-term dysfunction, which results in impaired performance, the decline in sports activity and performance or altogether termination of sports participation. It is, therefore, important to employ strategies to eliminate or, at least, decrease the occurrence and severity of these injuries.

Most conditioning programs do include a component of injury prevention. The current New Zealand SportSmart programme is an example of such a programme geared towards preventing ankle and knee injuries. However, although foot and ankle mechanics are cited in both ankle and knee injury mechanisms, very few injury prevention programs aim to improve the strength and function of the muscles controlling the foot. Researchers suggest that a strong foot has an important role to play in safe ankle, knee and hip movements. This study aims to investigate the influence that strengthening the foot muscles will have on ACL and lateral ankle sprain injury mechanisms.

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Who are we looking for?

We aim to evaluate females aged 16 to 25 years or old who are currently (and primarily) playing for a team competing in any court sports (netball, badminton, basketball, volleyball, etc.). You will be eligible to take part in this study if you don't have a current lower extremity injury nor had had an injury or surgery within the previous 6 months that required you to be absent from participating in a sporting activity for more than 4 weeks. In addition, you should not have been diagnosed to have any foot deformity/abnormality by a clinician.

What is going to happen?

If you volunteer to take part in this study, you will be asked to attend testing occasions and may be asked to take part in an intervention programme that will last the length of the playing season. You will be placed in 1 of 2 groups. The 1st group will be the intervention group and this group will perform a set of foot specific exercises in addition to all other normal training. The 2nd group will act as the control group. The 2nd group will not perform any additional, foot specific exercises. Both groups will be tested before the start of the season, and then again after an intervention period of 12 weeks. The intervention group then will be tested again before the start of the next playing season. Testing will take place at Massey University's LATU facility (see Map) where different types of data will be collected (see Table 1).

The participants placed in the intervention study will be required to do 3x20min sessions consisting of exercises aimed at strengthening the muscles acting on the foot. Most session will be home based. Regular supervised session will be arranged to ensure correct execution of exercises. Both groups will receive a pair of court sports shoes, which should be worn during regular court sport training sessions and matches. You will be required to keep a dairy of the number of sessions completed per week (intervention group only) and the amount of time spend wearing the provided shoes. Information about your injury history from the previous year, until the end of the study, will also be collected.

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Anthropometrics

Height and Weight will be measured.

3D Motion Capture

We will attach reflective markers (small silver balls) to your feet, legs and hips. We will use these markers to record the movements of your limbs in 3D while stepping onto a force plate when you execute jumping and agility movements. You will perform these movements twice. Once while barefoot and also in shoes that will be provided. You will be familiarised with the different jumping and agility movements that you will perform before any data is recorded.

(a) 

(b) 

Figure 1: Example of reflective markers placed while wearing shoes (a) and barefoot (b)

Jumps

You will perform a vertical jump for height. You will step with both feet on the force plate, jump up as high as you can and land back on the force plate with both feet. The force platform is embedded in the floor and will record ground reaction forces. We will record 5 attempts. This whole sequence will be repeated while wearing sports shoes and while being barefoot.

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Testing will take approximately 2 hours.

Clothing requirements

You will need to wear short tights or compression shorts without any reflective strips during the testing sessions. The reflective markers need to be attached to the skin or tight-fitting clothes in order to capture true limb movements with as little as possible anomalies. Regular clothes that are not skin-tight or have reflective strips may hide or interfere with the markers and render the data captured ineffectual, and should not be worn. The markers will be attached with double sided tape and will be held down with additional sports strapping tape. These adhesives do not cause damage to the skin or clothes and are widely used in biomechanical and sporting environments.

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What are the benefits of taking part in this study?

You will be able to receive a report *on request* with your results after the 2nd testing session to indicate your progress throughout the season. This data will not necessarily indicate whether you will sustain an injury, as injuries occur due to multiple factors. It will, however, indicate whether you have improved your jumping, speed, agility and deceleration abilities during the season. It will also indicate the changes in the foot, ankle and knee movements associated with ACL and lateral ankle sprain injuries during the playing season. You would have contributed to the sport scientific community by helping us to understand if and how the body adapts to the training of the foot muscles and whether we can help prevent future traumatic events and keep New Zealand's females safe and on the sports field.

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Participant's rights.

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- request a report of your individual results

What is the next step?

If you have any questions, you can ask any member of the research team at any time. If you have read and understood everything and is happy to take part, please contact the principal investigator. You will then be asked to complete 'The Health and Injury History Recruitment Questionnaire'. Details in the health-screening questionnaire will help determine if you are eligible to participate.

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treatment costs, travel costs for rehabilitation, loss of earnings, and/or lump sum for permanent impairment. Compensation for mental trauma may also be included, but only if this is incurred as a result of physical injury.

If your ACC claim is not accepted, you should immediately contact the researcher. The researcher will initiate processes to ensure you receive compensation equivalent to that to which you would have been entitled had ACC accepted your claim.

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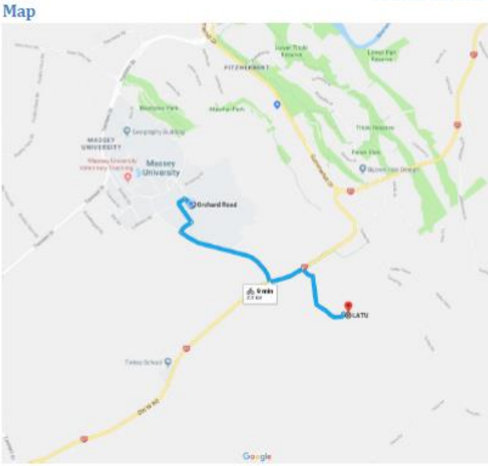
Map



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HEALTH AND INJURY RECRUITMENT QUESTIONNAIRE

MASSEY UNIVERSITY
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Foot strengthening and injury prevention

Health and Injury Recruitment Questionnaire

Participant Detail

Participant Name: ____________________

Date of Birth: | D | D | M | M | Y | Y | Y | Y | Age: __________

Shoe Size: __________ (indicated UK/US/EUR)

Main Sporting activity: ____________________

Secondary Sporting activity: ____________________

Contact Detail:

Email: ____________________

Mobile: ____________________

Contact in case of Emergency: ____________________

Emergency Contact Mobile: ____________________

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General Health Questions

Please complete the following physical activity readiness questionnaire. The form aims to identify any health problems so that we can avoid any risk of illness or injury. The information provided by you on this form will be treated with the strictest confidentiality (Cardinal, Esters, & Cardinal, 1996).

Please read the 8 questions below carefully and answer each one honestly: Check YES or NO	YES	NO
1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?	◯	◯
2. Do you feel pain in your chest when you do physical activity?	◯	◯
3. In the past month, have you had chest pain when you were not doing physical activity?	◯	◯
4. Do you lose your balance because of dizziness or do you ever lose consciousness?	◯	◯
5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?	◯	◯
6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?	◯	◯
7. Is your doctor currently prescribing the wear of foot orthotics for daily use or during activity?	◯	◯
8. Are you pregnant/ been pregnant or given birth in the last 6months?	◯	◯
9. Do you know of any other reason why you should not do physical activity?	◯	◯

If you have answered 'Yes' to any of the General Health Questions, please provide full details here:

____

____

____

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Injury Questionnaire

Please answer the questions below regardless of whether or not you have problems with your back, hips, knees, ankles or feet. Select the alternative that is most appropriate to you, and in the case that you are unsure, try to give an answer as best you can anyway. The term "problems" refers to pain, ache, stiffness, swelling, instability/giving way, locking or other complaints related to one or both limbs(Clarsen, Myklebust, & Bahr, 2013).

Please read the 4 questions below carefully and answer each one honestly:
Check FULL PARTICIPATION WITHOUT PROBLEMS, FULL PARTICIPATION BUT WITH PROBLEMS, REDUCED PARTICIPATION, NOT ABLE TO PARTICIPATE

	FULL PARTICIPATION WITHOUT PROBLEMS	FULL PARTICIPATION BUT WITH PROBLEMS	REDUCED PARTICIPATION	NOT ABLE TO PARTICIPATE
1. Have you ever had any difficulties participating in normal training and competition due to back, hips, knees, ankles or feet problems?	○	○	○	○
2. To what extent have you reduced your training volume due to problems the *past 4 weeks*?	○	○	○	○
3. To what extent have problems affected your performance during the *past 4 weeks*?	○	○	○	○
4. To what extent have you experienced problems related to your sport during the *past 4 weeks*?	○	○	○	○

I have read, understood and completed this questionnaire.

Signature: ______________________ Date: ____________

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References:

Cardinal, B. J., Esters, J., & Cardinal, M. K. (1996). Evaluation of the revised physical activity readiness questionnaire in older adults. *Medicine and Science in Sports and Exercise, 28*(4), 468-472.

Clarsen, B., Myklebust, G., & Bahr, R. (2013). Development and validation of a new method for the registration of overuse injuries in sports injury epidemiology: the Oslo Sports Trauma Research Centre (OSTRC) overuse injury questionnaire. *Br J Sports Med, 47*(8), 495-502. doi:10.1136/bjsports-2012-091524

Hollander, K., van der Zwaard, B. C., de Villiers, J. E., Braumann, K. M., Venter, R., & Zech, A. (2016). The effects of being habitually barefoot on foot mechanics and motor performance in children and adolescents aged 6-18 years: study protocol for a multicenter cross-sectional study (Barefoot LIFE project). *J Foot Ankle Res, 9*(1), 36. doi:10.1186/s13047-016-0166-1

Principal Investigator: PhD candidate
Mrs Carla van der Merwe
Email: c.vandermerwe@massey.ac.nz
Phone: 06 951 6817 ext. 83817

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ANKLE AND KNEE STABILITY QUESTIONNAIRE

MASSEY UNIVERSITY
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Foot strengthening and injury prevention

Ankle and Knee Instability Questionnaire

Participant Detail

Participant Name: ____________________

Functional ankle/knee instability

Please tick the ONE statement in EACH question that BEST describes your ANKLE and KNEE.	ANKLE			KNEE		
	LEFT	RIGHT	NONE	LEFT	RIGHT	NONE
1. Have you ever injured your ankle/ knee?	○	○	○	○	○	○
2. Have you ever seen a doctor for an ankle/ knee injury	○	○	○	○	○	○
a. If yes, How did the doctor grade your MOST SERIOUS injury?	Grade 1 ○ Grade 2 ○ Grade 3 ○	Grade 1 ○ Grade 2 ○ Grade 3 ○	○	Grade 1 ○ Grade 2 ○ Grade 3 ○	Grade 1 ○ Grade 2 ○ Grade 3 ○	○
3. Did you ever use a device (such as crutches) because you could not bear weight an ankle/knee injury?	○	○	○	○	○	○
a. If yes, In the most serious case, how long did you need to use the device?	1-3 days ○ 4-7 days ○ 1-2 weeks ○ 2-3 weeks ○ >3 weeks ○	1-3 days ○ 4-7 days ○ 1-2 weeks ○ 2-3 weeks ○ >3 weeks ○	○	1-3 days ○ 4-7 days ○ 1-2 weeks ○ 2-3 weeks ○ >3 weeks ○	1-3 days ○ 4-7 days ○ 1-2 weeks ○ 2-3 weeks ○ >3 weeks ○	○

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Please tick the ONE statement in EACH question that BEST describes your ANKLE and KNEE.	ANKLE			KNEE		
	LEFT	RIGHT	NONE	LEFT	RIGHT	NONE
4. Have you ever experienced a sensation of your ankle/knee "giving way"?	○	○	○	○	○	○
a. If yes, When was the last time your ankle/knee "gave way"?	<1 month ago ○ 1-6 months ago ○ 6-12 months ago ○ 1-2 years ago ○ >2 years ago ○	<1 month ago ○ 1-6 months ago ○ 6-12 months ago ○ 1-2 years ago ○ >2 years ago ○	○	<1 month ago ○ 1-6 months ago ○ 6-12 months ago ○ 1-2 years ago ○ >2 years ago ○	<1 month ago ○ 1-6 months ago ○ 6-12 months ago ○ 1-2 years ago ○ >2 years ago ○	○
5. Does your ankle/knee ever feel unstable while walking on a flat surface?	○	○	○	○	○	○
6. Does your ankle/knee ever feel unstable while walking on uneven ground?	○	○	○	○	○	○
7. Does your ankle/knee ever feel unstable during recreational or sport activity?	○	○	○	○	○	○
8. Does your ankle/knee ever feel unstable while going up stairs?	○	○	○	○	○	○
9. Does your ankle/knee ever feel unstable going *down* stairs?	○	○	○	○	○	○

Signature: ______________________________ Date: ______________

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Te Kunenga ki Pūrehuroa

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APPENDIX F: ETHICS APPROVAL AND STATEMENT OF CONTRIBUTION

MASSEY UNIVERSITY
TE KUNENGA KI PŪREHUROA
UNIVERSITY OF NEW ZEALAND

Date: 09 March 2017

Dear Carla Van der Merwe

Re: Ethics Notification - **SOA 16/77 - The potential of barefoot and minimalistic shoe wear to decrease risk of common injuries in adolescent females participating in court games**

Thank you for the above application that was considered by the Massey University Human Ethics Committee: **Human Ethics Southern A Committee** at their meeting held on **Friday, 10 March, 2017**.

Approval is for three years. If this project has not been completed within three years from the date of this letter, reapproval must be requested.

If the nature, content, location, procedures or personnel of your approved application change, please advise the Secretary of the Committee.

Yours sincerely

BJ Finch

Dr Brian Finch
Chair, Human Ethics Chairs' Committee and Director (Research Ethics)

Research Ethics Office, Research and Enterprise
Massey University, Private Bag 11 222, Palmerston North, 4442, New Zealand T 06 350 5573; 06 350 5575 F 06 355 7973
E humanethics@massey.ac.nz W http://humanethics.massey.ac.nz

APPENDIX G: INTERVENTION PROGRAMME DESCRIPTIONS

PHASE I: Week 1-4				PHASE II: Week 5-8				PHASE III: Week 9-12				PHASE IV: Week 13- 16			
Exercise	Ss	Rs	*Tempo	Exercise	Ss	Rs	Tempo	Exercise	Ss	Rs	Tempo	Exercise	Ss	Rs	Tempo
WARM-UP AND MOBILITY															
Metatarsal opener	2	5	1-3-1-0	**Metatarsal opener**	2	5	1-3-1-0	**Metatarsal opener**	1	10	1-3-1-0	**Metatarsal opener**	1	10	1-3-1-0
Toe extension / flexion	2	5	1-3-1-0	**Toe extension / flexion**	1	10	1-3-1-0	**Toe extension / flexion**	1	10	1-3-1-0	**Toe extension / flexion**	1	10	1-3-1-0
DISSOCIATION															
Individual toe tap	2	5	1-1-1-1	**Individual toe press**	2	10	1-3-1-1	**Toe wave**	1	10	1-3-1-1	**Toe wave**	1	10	1-3-1-1
Toe splay	2	10	1-1-1-1												
STRENGTH															
Lightest strength flexi-band				**Medium strength flexi-band**				**Strongest flexi-band**							
Big toe press - seated	2	10	1-1-1-3	**L/R = Short foot and R/L = Inversion**	2	10	SF: 1-3-0-0 Inv: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Inversion**	3	10 +2/ wk	SF: 1-3-0-0 Inv: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Inversion**	3	20	SF: 1-3-0-0 Inv: 1-1-3-0
Short foot – seated	2	10	1-1-1-3												
Inversion – seated	2	10	3-0-1-2												
Eversion – seated	2	10	3-0-1-2	**L/R = Short foot and R/L = Eversion**	2	10	SF: 1-3-0-0 Eve: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Eversion**	3	10 +2/ wk	SF: 1-3-0-0 Eve: 1-1-3-0	**L/R = Short foot (+ toe spread) and R/L = Eversion**	3	20	SF: 1-3-0-0 Eve: 1-1-3-0
Dorsiflexion - seated	2	10	3-0-1-2												
Calf raise	2	10	3-0-1-2	**Single leg calf raises**	2	10	3-0-1-3	**Single leg hop**	3	10	1-0-3-0	**Continues single leg hop**	3	20	1-0-3-0

** Tempo: Concentric - Isometric/Stretch hold – Eccentric – Isometric/Stretch hold; Ss = Sets; Rs = Reps; SF = Short foot; Inv = Inversion; Eve = Eversion; wk = week*

Warm-up and Mobility

Metatarsal Opener: PHASES I - IV

Equipment:

- Ball (provided)

Set up:

- In a standing position.
- Shift your body weight to the non-exercised foot.
- Place the ball underneath the exercised foot.

Movement:

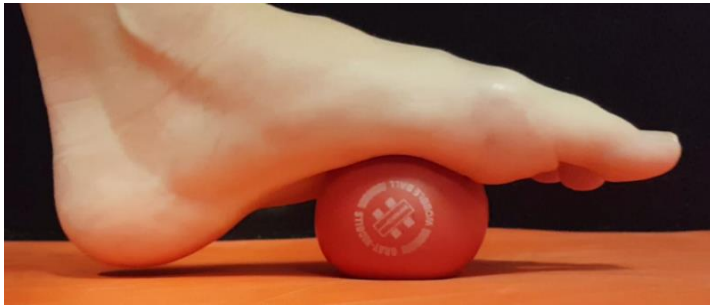
- Start with the ball under the heads of the metatarsals (ball of the foot).
- Roll the ball towards the heel and back towards the metatarsal heads.
- Gently apply pressure to the exercised foot so that the toes start to splay over the ball.
- 3 seconds hold at end range.
- Return to starting position and change the position of the ball to different parts of the foot.

Note:

- Cease to apply pressure when the pressure becomes painful.

REPEAT ON BOTH FEET

PHASE I - IV



TOE FLEXION/EXTENSION: PHASE I – IV

Equipment:

- None

PHASE I

Set up:

- In a seated position.
- Bend the knee of the exercised foot so that the heel and ball of the foot are on the floor.

Movement:

- Grab hold of the big toe with the hand on the same side as the exercised foot.
- Keep the heel the exercised foot in contact with the floor.
- Gently bring the big toe up as far as possible.
- Hold this position for the indicated amount of time.
- Now lift the ball of the foot off the floor and gently push the big toe towards the floor.

PHASE II, III & IV

Set up:

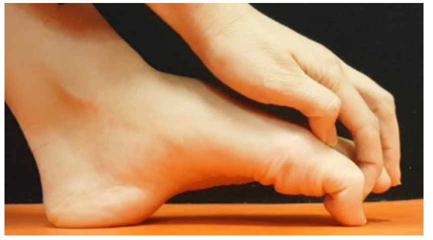
- Put your weight on the none exercised foot.
- Bend the knee of the exercised foot so that the tips of the toes of the exercised foot is on the floor.

Movement:

- Keeping the toes in touch with the floor bring the heel upwards until the toes form a 90-degree angle at the floor.
- Gently apply pressure to the toes in this position - holding the position for 3 seconds.
- Bring the heel further forward and role the heel in front of the toes, rolling over the toes.
- Gently apply pressure on the toes in this position - holding the position for 3 seconds.

Note: The stretch should be uncomfortable, but not painful. REPEAT ON BOTH FEET

PHASE I



PHASE II, III & IV

DISSOCIATION

INDIVIDUAL TOE TAP: PHASE I

Equipment:

- None

Set up:

- In a standing position.
- Toes and heels flat on the floor.

Movement:

- Lift all the toes of the floor simultaneously.
- Tap the big toe down and lift it back up again.
- Now repeat with the little toe followed by the rest of the toes from small toe to the second toe. Lift it back up again and in reverse order.

REPEAT ON BOTH FEET

PHASE I

TOE SPLAY: PHASE I

Equipment: • None Set up: • In a standing position. • Toes and heels flat on the floor. Movement: • Splay the toes as far apart as possible. • Bring the toes back together aiming not to flex the toes. REPEAT ON BOTH FEET	**PHASE I**  

Individual toe press: PHASE II

	PHASE II
Equipment: • None Set up: • In a standing position. • Toes and heels flat on the floor. Movement: • Lift all the toes simultaneously. • Press the big toe down into the floor as hard as possible. • Hold this position for 3 seconds. • Lift the big toes. • Put down the rest of the toes starting at the small toe • When all the toes are on the floor, press all the toes into the floor. • Hold this position for 3 seconds. Note: • Always keep the ball of the foot and the heel in contact with the floor.	  

TOE WAVE: PHASE III & IV

Equipment:

- None

Set up:

- In a standing position.
- Toes and heels flat on the floor.

Movement:

- Lift all the toes off the floor simultaneously.
- Tap the big toe down and lift it back up again.
- Tap down with the little toe followed by the rest of the toes from small toe to the big toe.
- Lift the toes back up starting from the big toe followed by the second toe to the little toe.

Note:

- Always keep the ball of the foot and the heel in contact with the floor.

PHASE III & IV

Strength

Big toe press: PHASE I

Equipment:

- None

Set up:

- In a seated position.
- With the knees bent.
- Plantar surface of the foot flat on the floor.

Movement:

- Lift up all the toes simultaneously.
- Press the big toe down into the floor as hard as possible.
- Hold this position for 3 seconds.

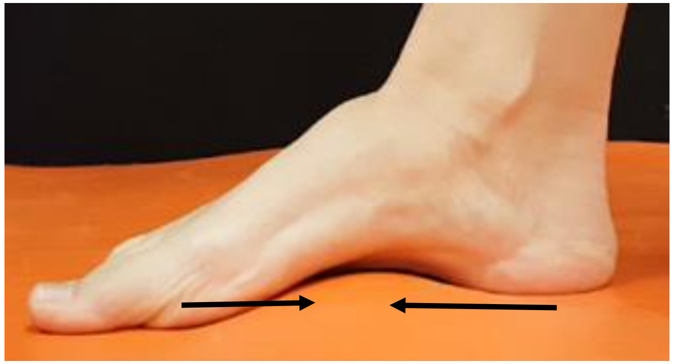
Note:

- Always keep the ball of the foot and the heel in contact with the floor.

REPEAT ON BOTH FEET

PHASE I

SHORT FOOT: PHASE I

Equipment:	PHASE I
• None Set up: • In a seated position. • With the knees bent. • Plantar surface of the foot flat on the floor. Movement: • Spread out the toes as in the toe splay exercise. • Raise the arch (middle part of the foot) by sliding the big toe towards the heel. • Hold this position for 3 seconds. Note: • Always keep the ball of the foot and the heel in contact with the floor. • Do not flex and grab with the toes. REPEAT ON BOTH FEET	 

SHORT FOOT: PHASE II

Equipment:

- None

Set up:

- In a standing position.

Movement:

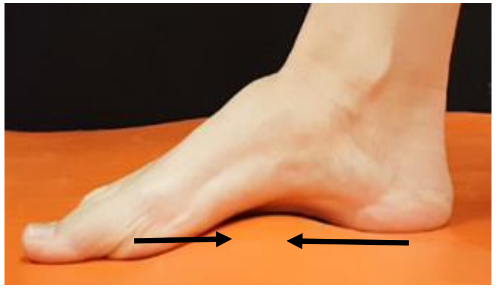
- Spread out the toes as in the toe splay exercise.
- Raise the arch (middle part of the foot) by sliding the big toe towards the heel.
- Hold this position for 3 seconds.

Note:

- Always keep the ball of the foot and the heel in contact with the floor.
- Do not flex and grab with the toes.

REPEAT ON BOTH FEET

PHASE II



Ankle Inversion: PHASE I

PHASE I

Equipment:

- Flexi band

Set up:

- In a seated position.
- Stretch your legs out in front of you.
- Dorsiflex the feet so that the toes point straight up towards the ceiling.
- Place the flexiband just below the ball of the exercised foot.
- Cross the opposite foot across the shank of the exercised foot to add resistance to the band.
- Grab hold of the flexi band towards the little toes.

Movement:

- Start with the exercised foot twisted to the inside (adducting).
- Resist the pull of the band while you allow the band to pull the foot towards the outside (3 sec).
- Without pausing at the end range (0 sec)
- Twist the foot to the inside, back to the starting position.
- Hold this position for 2 seconds.

Note:

- Keep the heel of the exercised foot on the floor.
- Do not rotate the shank.

REPEAT ON BOTH FEET

Ankle Inversion: PHASE II

Equipment:
- Flexi band

Set up:
- In a standing position.
- Heel and metatarsal heads in contact with the floor.
- Place the flexi band around the head of the first metatarsal (big toe).

Movement:
- Start with the exercised foot on the floor.
- Lift the front of the foot of the floor (keep the heel of the exercised foot in contact with the floor).
- Twist the exercised foot to the inside and upwards in such a way that the big toe is higher that the little toe.
- Hold this position for 3 seconds.
- Resist the pull of the band while you allow the band to pull the foot towards the starting position (3 sec).
- Without pausing at the end range (0 sec), twist the foot to the inside, and upwards.

Note:
- Keep the heel of the exercised foot on the floor.
- Do not rotate the shank.

REPEAT ON BOTH FEET

PHASE II

SHORT FOOT AND INVERSION: PHASE III & IV

Equipment:

- Flexi band
- Partner or fixed object.

Set up:

- In a standing position.
- The heel and metatarsal heads of both feet in contact with the floor.
- Place the flexiband around the head of the first metatarsal (big toe) of the exercised foot.

Movement:

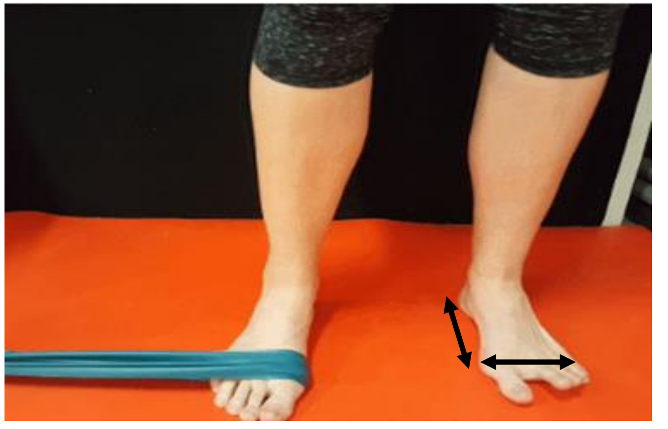
- Spread the toes of the stabilizing foot and lift the arch as in the 'Short Foot' exercise.
- Rotate the exercised foot to the inside and upwards in such a way that the big toe is higher that the little toe.
- Hold this position for 3 seconds.
- Resist the pull of the band while you allow the band to pull the foot towards the starting position (3 sec).

Note:

- Keep the heel of the exercised foot on the floor.
- Do not rotate the shank.

REPEAT ON BOTH FEET

PHASE III & IV



ANKLE EVERSION: PHASE I

Equipment:

- Flexi band

Set up:

- In a seated position
- Stretch your legs out in front of you.
- Dorsiflex the feet so that the toes point straight up.
- Place the flexi band around the ball of the exercised foot.
- Grab hold of the flexi band towards the big toe.
- Place the opposite foot parallel to the exercised foot, one-foot length apart so that this foot apply pressure on the band.

Movement:

- Start with the exercised foot twisted to the outside (abducting).
- Resist the pull of the band while you allow the band to pull the foot towards the inside (3 sec).
- Without pausing at the end range (0 sec) twist the foot to the outside, back to the starting position.
- Hold this position for 2 seconds.

Note:

- Keep the heel of the exercised foot on the floor.
- Do not rotate the shank.

REPEAT ON BOTH FEET

PHASE I

ANKLE EVERSION: PHASE II

Equipment:
- Flexi band
- Partner or fixed object.

Set up:
- In a standing position.
- Heel and metatarsal heads in contact with the floor.
- Place the flexi band around the head of the fifth metatarsal (little toe)

Movement:
- Start with the exercised foot on the floor.
- Lift the front of the foot off the floor (keep the heel of the exercised foot in contact with the floor).
- Twist the exercised foot to the outside and upwards in such a way that the little toe is higher that the big toe.
- Hold this position for 3 seconds.
- Resist the pull of the band while you allow the band to pull the foot towards the starting position (3 sec).
- Without pausing at the end range (0 sec) twist the foot to the inside, and upwards.
- Hold this position for 2 seconds.

Note:
- Keep the heel of the exercised foot on the floor.
- Do not rotate the shank.

REPEAT ON BOTH FEET

PHASE II

SHORT FOOT AND EVERSION: PHASE III & IV

Equipment:

- Flexi band
- Partner of fixed object.

Set up:

- In a standing position.
- The heel and metatarsal heads of both feet in contact with the floor.
- Place the flexiband around the head of the first metatarsal (big toe) of the exercised foot.

Movement:

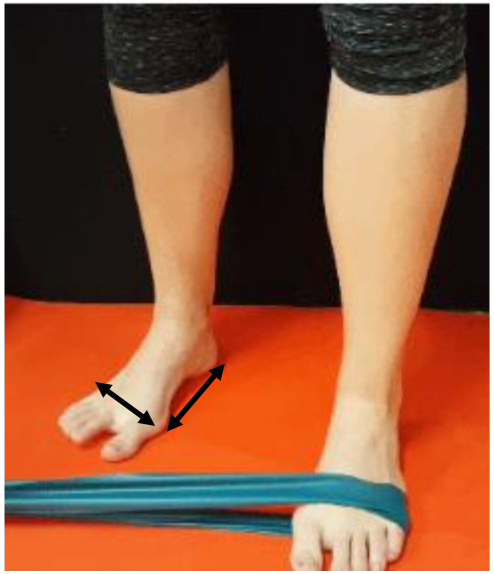
- Place the flexiband around the head of the first metatarsal (big toe) of the exercised foot.
- Spread the toes of the stabilizing foot and lift the arch as in the 'Short Foot' exercise
- Rotate the exercised foot to the outside and upwards in such a way that the little toe is higher that the big toe.
- Hold this position for 3 seconds.
- Resist the pull of the band while you allow the band to pull the foot towards the starting position (3 sec).

Note:

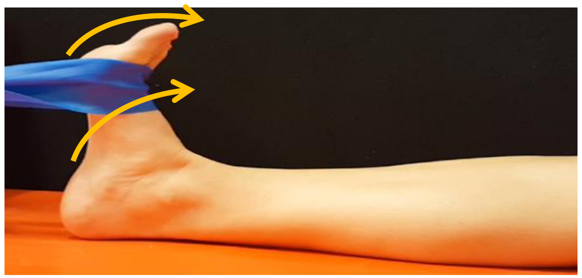
- Keep the heel of the exercised foot on the floor.
- Do not rotate the shank.

REPEAT ON BOTH FEET

PHASE III & IV



ANKLE DORSI FLEXION: PHASE I

Equipment:	PHASE I
Equipment: • Flexi band Set up: • Fix the flexi band to a sturdy object. • In a seated position. • Stretch your legs out in front of you. • Dorsiflex the feet so that the toes point straight up. • Place the flexi band around the front of the foot at the base of the toes of the exercised foot. • The flexi band is directed away from you. • Apply pressure to the flexi band. Movement: • Start with the toes of the exercised foot pointed towards you. • Resist the pull of the band while you allow the band to pull the foot away from you (3 sec). • Without pausing at the end range (0 sec) • Pull the toes towards you, back to the starting position. • Hold this position for 2 seconds. Note: • Keep the heel of the exercised foot on the floor. • Do not bend the knee or hitch up the hip. REPEAT ON BOTH FEET	 

CALF RAISE: PHASE I

Equipment:

- Step
- Ball

Set up:

- Standing upright.
- Balance on the balls of the feet on the edge of the step.
- Squeeze the ball between the heels.

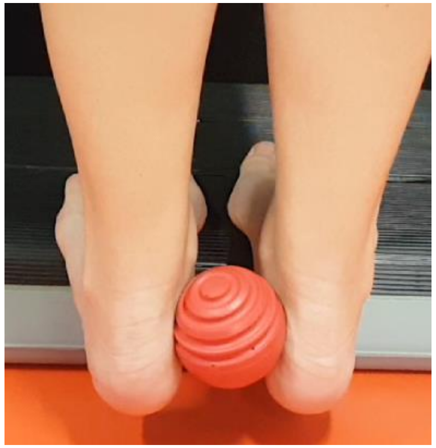
Movement:

- Slowly lower the heels down as far possible without losing contact with the step (3 sec).
- Without pausing at the bottom (0 sec).
- Raise the heels back up as far as possible (1 sec).
- Hold this position (2 sec).
- Control the return to the starting (3 sec).

Note:

- Maintain a full range of motion throughout all repetitions.

PHASE I



SINGLE LEG CALF RAISE: PHASE II

	PHASE II	
Equipment: • Step Set up: • Standing upright. • Balance on the exercised leg. Movement: • Slowly lower the heel down as far as possible without losing contact with the step (3 sec). • Without pausing at the bottom (0 sec). • Raise the heel back up as far as possible (1 sec). • Hold this position (3 sec). • Control the return to the starting in the indicated amount of time. Note: • Maintain a full range of motion throughout all repetitions. REPEAT ON BOTH FEET		

SINGLE LEG JUMP: PHASE III

Equipment:	**PHASE III**	
Equipment: • None. Set up: • Standing upright with hands on the hips. • Balance on the exercised foot. Movement: • Perform a single leg jump. • Focus on using the toes to push you off from the floor. • Control the lowering of the heel towards the floor on landing. • Find your balance and repeat the movement. REPEAT ON BOTH FEET		

Single leg hops: PHASE IV

Equipment:	**PHASE IV**	
Equipment: • None Set up: • Standing upright with hands on the hips. • Balance on the exercised foot. Movement: • Perform consecutive single leg jumps. • Focus on using the toes to push you off from the floor. • Control the lowering of the heel towards the floor on landing. • Find your balance and repeat for as many consecutive jumps (not exceeding 20) as possible without putting the other foot down. REPEAT ON BOTH FEET