The effect of foot orthoses on muscle activity and morphology, foot biomechanics and skin sensitivity

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This thesis is dedicated to the memory of John Potier

(1941-2017) who was there for me at the start of my university journey and was an inspiration

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Abbreviations

ABH	Abductor hallucis
ATINS	Achilles tendon at the insertional site
ATMID	Achilles tendon at the mid-portion
BMI	Body mass index
CAD/CAM	Computer-aided design/computer-aided manufacturing
CI	Confidence interval
CNS	Central nervous system
CSA	Cross sectional area
CV	Coefficient of variation
EMG	Electromyography
EVA	Ethylene-vinyl acetate
FA	Fast adapting
FAI	Fast adapting type I
FAII	Fast adapting type II
FDB	Flexor digitorum brevis
FDL	Flexor digitorum longus
FHL	Flexor hallucis longus
FO	Foot orthoses
FPI	Foot posture index
GRF	Ground reaction force
ICC	Intra class correlation
IPAQ	International Physical Activity Questionnaire
MaxES	Inversion at foot contact
MaxEv	Peak rearfoot eversion
MaxMEv	Peak external eversion moment
MaxMInv	Peak external inversion moment
MG	Medial gastrocnemius
MLA	Medial longitudinal arch
MTP1	1st metatarsal head
MTP5	5th metatarsal head

MTSS	Medial tibial stress syndrome
MU	Motor unit
MUAPs	Motor unit action potentials
MVC	Maximum voluntary contraction
PCSA	Physiological cross-sectional area
PF	Plantar fascia
PFINS	Proximal plantar fascia
PFMID	Mid-portion plantar fascia
PFP	Patellofemoral pain
PL	Peroneus longus
RCT	Randomised control trial
ROM	Range of motion
RR	Risk ratio
SA	Slow adapting
SAI	Slow adapting type I
SAII	Slow adapting type II
SALRE	Subtalar joint axis location rotational equilibrium theory
SPFT	Sagittal plane facilitation theory
STJ	Subtalar joint
SEM	Standard error of measurement
TA	Tibialis anterior
TP	Tibialis posterior
TPTD	Tibialis posterior tendon dysfunction
VM	Vastus medialis
VL	Vastus lateralis

Abstract

Foot orthoses with a medial wedge or medial arch support are commonly used to treat musculoskeletal pathologies by altering external forces applied to the foot, which could consequently alter internal forces generated by muscles. However, little is known about whether systematic changes in foot orthosis geometry result in systematic changes in activation of lower limb muscles. This PhD investigated the effects of foot orthoses on selected lower limb muscles.

A systematic review was conducted to establish if evidence exists that footwear, foot orthoses and taping alter lower limb muscle activity during walking and running. The review identified some evidence that foot orthoses can decrease activity of tibialis posterior in early stance and possibly increase activity of peroneus longus in mid-late stance, while not altering activity of other lower limb muscles. Findings concerning the peroneus longus were limited by previous reports of the poor reliability of EMG recordings from this muscle. A reliability study was thereafter conducted to demonstrate a reliable protocol for recording EMG from the peroneus longus, using ultrasound guidance and small surface sensors to improve results. This technique was used in the subsequent study on the effect of foot orthoses on lower limb muscle activity.

A study of the immediate effect of foot orthoses was undertaken, with the aim of establishing whether medial heel wedging and increased medial arch height affect EMG of shank muscles and foot and ankle moments/motion. Muscle activity was recorded from 23 healthy participants using surface EMG and fine-wire EMG (tibialis posterior) in combination with kinematic and kinetic data during walking in shoes with four different foot orthoses. Tibialis posterior activity decreased in early stance by 17% ($p=0.001$) with a Salfordinsole orthosis with an additional 8 mm increase in medial heel wedging and by 14% ($p= 0.047$) with a Salfordinsole orthosis with both a 6 mm increase in arch height and an 8 mm increase in heel medial wedging. The reduced tibialis posterior activity with medial wedging in combination with reduced external ankle eversion moment provides a possible link between foot orthosis design and biomechanical effect and could be used to inform treatment practice.

Building on the literature review and the results of the immediate effects study, it was hypothesised that altered loading of the foot with long term use of foot orthoses would alter the mechanical work required of internal structures. The purpose of the final study in this PhD was to investigate any effect of using foot orthoses over three months on soft tissue morphology and skin sensitivity. Twenty three healthy participants wore an orthosis that changed peak

pressure in the medial arch and the heel by 8%, while nineteen healthy participants continued to wear their convention footwear. There were no changes in skin sensitivity, or the thickness and cross sectional area of intrinsic foot muscles and connective tissue after three months of orthoses use. This finding provides evidence to challenge the view held by some that foot orthoses make muscles smaller (and weaker).

Chapter 1 INTRODUCTION

1.1 Summary

Foot orthoses (FO) are designed to redistribute load under the foot and thereby alter external joint moments, internal joint moments (generated by muscles and other soft tissues), and change foot motion. Examples include contoured insoles with medial rearfoot wedges and medial arch supports (Murley et al., 2010a) which are the focus of this thesis and are the style of FO for which subsequent references to “FO” refer.

There are a range of common conditions affecting the lower extremity that FO are used to treat or prevent. Some of the most common are: plantar fasciitis, Achilles tendinopathy, shin pain (tibialis posterior tendon dysfunction (TPTD) and medial tibial stress syndrome (MTSS)) and anterior knee pain (such as patellofemoral pain (PFP)). For instance, plantar fasciitis or heel pain (50.3%) were the most commonly reported reasons that practitioners reported recommending a retail pre-fabricated orthotic in a recent survey in the United Kingdom (Nester et al., 2017). However the clinical concepts behind FO prescription are not well supported by biomechanical evidence and the mechanisms with which FO can be an effective clinical treatment are poorly understood (Chevalier and Chockalingam, 2011, Mills et al., 2009, Nester et al., 2003a). Many of the conditions for which FO are used are soft tissue injuries and so it is important to understand how it is that soft tissues might be affected by FO use. The internal load through soft tissue is partly dependent on the activation of muscle, which in turn can be modulated by feedback from the periphery to the central nervous system, including from mechanoreceptors in the skin. As such there could be multiple mechanisms by which FOs exert their effects and the primary mechanism could also differ by pathology. For this reason, a holistic approach underpins the work in this thesis, and the combined effects of FOs on muscle activity and morphology, foot biomechanics and skin sensitivity are investigated.

1.2 Thesis outline

In Chapter 2 the theories on the mechanism of FO effect are reviewed from the classical perspective of foot alignment, kinematics and kinetics and more recent neuromuscular theories. As a particular focus of this thesis is the effect of FO on soft tissues, the latter part of Chapter 2 describes the anatomy of muscles and connective tissue that may be affected by FO use and identifies theories of FO effect that are most relevant to pathology in those tissues.

As revealed in Chapter 2, less is known about the effect of FO on internal forces than external forces at the foot and ankle. Consequently a literature review was undertaken to establish the effect of FO on electromyography (EMG). Chapter 3 is an adapted version of a systematic review, published in Prosthetics & Orthotics International, on the effect of footwear, foot orthoses and taping on lower limb muscle activity during walking and running. Footwear and taping were reviewed along with FO because these external devices can be used to treat similar conditions and as such reviewing the three together could be of clinical relevance. Secondly, from a mechanistic perspective, establishing whether other external devices relevant to foot and ankle biomechanics alter EMG may offer some insight for understanding FO effects.

The main methods used in the experimental studies are detailed in Chapter 4. An overview of electromyography, the specifications of the sensors used and a detailed description of the development of the technique used to record indwelling EMG from the tibialis posterior are provided. The subsequent sections outline motion capture, kinetics and ultrasound theories and techniques.

In the literature reviews in Chapters 2 and 3, poor between day reliability of peroneus longus (PL) EMG recording had been identified. In Chapter 5 a new protocol was developed using ultrasound guidance and small EMG sensors to reliably record from PL during walking. The reliability of recording surface EMG from the abductor hallucis and the medial longitudinal arch angle using markers on wands affixed to the skin through holes in the shoe was also established.

Chapter 6 investigated the effect of FO with medial rearfoot wedging and increases in medial arch height on EMG of the tibialis posterior, tibialis anterior, medial gastrocnemius, peroneus longus and abductor hallucis. Concurrent collection of ankle kinematics and kinetics and medial longitudinal arch angle were recorded to link possible internal and external effects of the FOs.

Chapter 7 investigated the effects over time of FOs on soft tissue morphology, plantar pressure and skin sensitivity. Participants wore FOs that provided a minimum increase of 8% plantar pressure in the medial arch and a minimum decrease of 8% in plantar pressure at the medial heel or continued with their normal footwear if in the control group. The thickness and cross sectional area of structures around the foot and ankle were measured using ultrasound before and after the intervention. Plantar pressure and skin sensitivity to light touch was also measured before and after the intervention.

The discussion in Chapter 8 provides a summary of the thesis and its key findings. The novelty of the work as well as limitations are presented, with directions for future research and implications for clinical practice.

Chapter 2 BACKGROUND

2.1 Introduction

In this chapter firstly the potential mechanisms of foot orthoses (FO) effect are discussed. Since many of the conditions for which FO are used affect soft tissues, there is a focus on effects of FO on the musculoskeletal system. The gaps in the existing literature relating to FO effect are identified and provide the rationale for the subsequent studies in this thesis. In particular there is a need for more research on neuromuscular theories relating to FO. Consequently, this chapter then provides a background of muscle function and EMG and tendon mechanics. Finally this chapter outlines the anatomy and function of specific foot and ankle soft tissues relevant to FO effects.

2.2 Clinical and biomechanical theories of foot orthotic effect

There are a number of clinical and biomechanical theories that have been proposed to explain the possible mechanisms by which FO may exert therapeutic effects. These potential mechanisms relate to foot posture, kinematics, external forces and neuromuscular control and have been the subject of reviews (Harradine and Bevan, 2009, Mills et al., 2009). However no singular theory on the mechanism of FO effect has been accepted and it is possible that the potential for a given mechanism to be effective may be subject-specific, given the large variability in foot function. The mechanism of therapeutic effect may also depend on the condition that the FO is intended to treat, so where appropriate the evidence for theories are discussed with respect to the common conditions for which FO are used, such as patellofemoral pain (PFP).

2.2.1 *Database searches*

A search of the literature was performed in Web of Science, PubMed Central and Google Scholar with no date limits using the following search terms (* and $ are wildcards):

orthot* OR insert OR wedge OR orthos$ OR insole OR skive OR shoe OR foot OR taping OR motion control footwear) AND

kinematic$ OR kinetic$ OR force OR pressure OR motion OR pronation OR calcaneal eversion OR biomechanics OR shock attenuation OR shock absorption OR moment AND

overuse injur* OR lower limb OR leg OR walk* OR run* OR gait OR locomotion or jog*

Rootian theory OR Root OR sagittal plane facilitation theory OR subtalar joint axis location rotational equilibrium theory OR preferred movement path*References were also obtained from the citations of relevant articles identified.

2.2.2 *Rootian theory*

The traditional clinical basis of FO prescription emphasises changing foot alignment to directly change kinematics. This likely stems from the proposed relationship, even if not causal, between certain pathologies and excessive pronation and/or a flat foot posture. The Rootian theory desires feet to be "neutrally aligned", involving mainly the subtalar joint (STJ) alignment in midstance (Root, 1977). Accordingly FO try to correct structural deviations from this neutral alignment in a foot morphology theory (Harradine and Bevan, 2009) or kinematic paradigm (Mills et al., 2009). Foot orthoses can reduce calcaneal eversion by around 2° (Cheung et al., 2011, Mills et al., 2009). However, recent work using medially posted insoles created by a physiotherapist guiding participants' feet towards a neutral alignment in a mold, resulted in reduced peak eversion of the forefoot but not the hindfoot (Kosonen et al., 2017). The Rootian theory is criticised as during asymptomatic gait feet rarely achieve STJ neutral and variability in overall kinematics is large (McPoil and Cornwall, 1994, Nester, 2009). In a study of 100 asymptomatic participants, none of the deformities that form the Rootian theory were associated with specific kinematic patterns in walking and the authors discouraged the use of the model in clinical practice (Jarvis et al., 2017). However, foot orthoses continue to be prescribed based on the theory of achieving a neutral position of the STJ, as positive outcomes can be achieved despite sound theoretical justification and no validated alternative theory has been accepted (Harradine et al., 2018).

2.2.3 *Foot posture theory*

Similar to the Rootian theory aiming to achieve neutral alignment of the STJ in walking, FO can be prescribed with the desire to achieve a neutrally aligned subtalar joint, or neutral foot posture, in standing. There are numerous ways in which *static foot posture* can be assessed and thereafter associated with foot pathology and the need for FO. Posture can be assessed based on foot dimensions, mobility/flexibility, angles, footprint indices and a composite measure of a number of observations (Foot Posture Index, FPI). Feet can be grouped into three broad classifications of foot posture based on how the foot presents when weight bearing (Redmond et al., 2006): 1) pes planus/pronated/flat; 2) normal/neutral and 3) pes cavus/supinated/high arch.

However, static classifications of foot posture have limited resemblance to foot kinematics and as such may have little relevance to muscle and foot function (Buldt et al., 2015). Foot posture is influenced by soft tissue and bone structure, however weight-bearing x-rays of bones that contribute to foot structure explained only ~35% of variance in plantar pressure during walking (Cavanagh et al., 1997). If the contribution of foot structure, which partly determines foot posture, to gait is relatively small, then perhaps aiming to change foot posture/position with FO to achieve a common, desirable walking pattern is misguided.

2.2.3.1 *Quantifying foot posture*

Measures of foot posture often relate to quantifying the medial longitudinal arch (MLA) characteristics, with particular reference to the height of the navicular bone with respect to the floor (Xiong et al., 2010). The MLA is comprised of the calcaneus, talus, navicular, three cuneiforms and 1-3 metatarsals, along with many soft tissue attachments of muscle and connective tissue (Williams and Nester, 2010a). The height of the navicular with respect to the ground, which is sometimes normalised to entire foot length or foot length minus the toes (Cowan et al., 1993, Williams and McClay, 2000), is a simple, easy to perform measure that has demonstrated good validity (Pearson r =0.79), with respect to its radiographic equivalent in older adults (Menz and Munteanu, 2005). However static assessment of navicular drop has been shown to be a poor predictor of dynamic navicular drop (Hoffman et al., 2015). Arch height is also sometimes measured as height of the dorsum at 50% of foot length (Mootanah et al., 2013, Zifchock et al., 2006). Although static arch height is objective, it does not reflect dynamic behaviour of the foot (Razeghi and Batt, 2002). The MLA angle has been tracked dynamically in barefoot walking (Leardini et al., 2007, Rabbito et al., 2011). However, the reliability of tracking the MLA angle using markers attached to the skin through holes in the shoe when walking shod should be established before MLA is used as a variable in studying the effect of FO.

Foot posture is frequently assessed using the foot posture index (FPI) (Redmond, 2005, Redmond et al., 2006) and is commonly used to screen, characterise or subgroup participants in clinical research (Selfe et al., 2016). The FPI is a six-point static assessment tool whereby the observer views the weight bearing foot from multiple angles to obtain a value that will categorise the foot as either supinated, neutral or pronated. The FPI is designed to assess both the forefoot and rearfoot in all three planes using the categories outlined by Redmond (2005, 2006):

1. Talar head palpation

2. Supra and infra lateral malleolar curvature
3. Calcaneal frontal plane position
4. Prominence in the region of the talonavicular joint
5. Congruence of the medial longitudinal arch
6. Abduction/adduction of the forefoot on the rearfoot

Each category is given a value ranging from -2 (highly supinated) to +2 (highly pronated) in increments of 1 (Redmond, 2005, Redmond et al., 2006). A foot is classified as pronated/flat if the total score is ≥ +6 (Maharaj et al., 2018) and a foot is classified as supinated if the total score is ≤ -6. A large scale (n= 1,648) study found a relationship between FPI and age and pathology, but not sex or body mass index BMI (Redmond et al., 2008). However BMI has been shown to influence FPI, as Butterworth et al. (2015) found their obese participants (median (range) 6 (-8 to 12)) had a significantly greater FPI than their non-obese participants (median (range) 2 (-6 to 9)). Consequently FPI may be confounded by body mass similar to arch index measures (Wearing et al., 2004). Nevertheless, FPI has been shown to have the strongest association with kinematic variables in barefoot walking compared to other static foot posture measures and mobility measures (Buldt et al., 2015). Owing to this superior relationship with kinematics over other measures, its speed and ease of use and not requiring specific equipment, the FPI was chosen as the measure of foot posture in the studies in this thesis.

2.2.4 *Decreased internal rotation theory and foot posture: Patellofemoral pain/anterior knee pain*

Foot orthoses that reduce rearfoot eversion may exert their benefit in cases of PFP by decreasing tibial internal rotation and altering knee mechanics as transverse plane motion of the tibia is coupled to frontal plane motion of the foot (Chuter and de Jonge, 2012, Ferber et al., 2005, Fischer et al., 2018, Resende et al., 2015, Rodrigues et al., 2015, Tiberio, 1987, Williams and Nester, 2010c).Tibial internal rotation has been found to decrease with posted non-moulded FO by (1.33° (0.12 to 2.53)) in non-injured cohorts (Mills et al., 2009). Medial wedging in the rearfoot and rearfoot and forefoot together have been shown to reduce peak ankle eversion angle and knee internal rotation moment during step descent (Bonifacio et al., 2018). However no systematic main effect for magnitude of medial rearfoot posting on internal tibial rotation was found in walking by Telfer et al. (2013b), despite a dose-response in other kinematic and kinetic variables at the knee and ankle. Assessing foot posture can be useful in the assessment of PFP as weak muscle strength and a pronated foot was one of the three subgroup identified in PFP patients (Selfe et al., 2016). Expert consensus recommends prefabricated FO for the short term relief of pain in PFP and combined interventions, such as FO and exercise, for short and medium term relief of pain (Collins et al., 2018). However there is a lack of evidence to support the long term (>12 months) efficacy of FO in managing pain in PFP or the use of custom FO (Collins et al., 2018).Identifying subgroups within PFP patients using clinical tests such as the FPI may help tailor interventions and improved the clinical efficacy of treatments like FO (Selfe et al., 2016).

2.2.5 *The sagittal plane facilitation theory (SPFT)*

Another kinematic theory, the sagittal plane facilitation theory (SPFT) (Dananberg, 1986), prioritises anterior displacement of the centre of mass during walking and suggests that interventions such as FO should enable adequate ankle and hallux dorsiflexion. The biomechanical effects of FO prescribed using the SPFT have not been tested and it is unlikely this design could conceivably benefit the range of lower limb pathologies for which anti-pronation FO are prescribed (Sweeney, 2016). However, experimentally induced pronation was found to reduce the anterior displacement of the centre of pressure (COP) in walking (Resende et al., 2019), so perhaps reducing pronation may facilitate anterior displacement of the COP. Nevertheless there are multiple kinematic patterns in asymptomatic feet (Nester et al., 2014), thus the kinematics of patients may not be an explanation for pathology, but part of natural variability in foot function and attempting to “correct” their kinematics with FO is not empirically justified.

2.2.6 *Joint moments theory*

Ankle moments in gait can be altered by FO (McMillan and Payne, 2008, Nester et al., 2003b, Telfer et al., 2013b). Joint moments (from force and moment arm) generate joint kinematics. Ultimately forces cause injury, not motion/position, as forces lead to structures experiencing greater than tolerable load (Payne, 2007). Telfer and colleagues (2013b) reported peak and mean ankle eversion moment reductions of 1.1±1.1% and 2.3±2.1% per 2° of medial posting respectively. Decreased external eversion moment (or increased external inversion), would mean decreased internal inversion moment required from internal structures to counteract the external force. A reduction in loading in these structures could be beneficial for the treatment of injuries such as TP tendinopathy. Reduced internal ankle inversion moment with FO has been found in both walking and running (Hsu et al., 2014, Mundermann et al., 2003, Sweeney, 2016, Williams et al., 2003).

A change in ankle moment could arise from a change in the magnitude of force applied to the foot (quantified as the mean or peak plantar pressure) or by a change in the moment arm, by a change in the point of application of the force (centre of pressure) (Jones et al., 2015). It is thought that medial wedges increase loading on the medial aspect of the calcaneus (Bonanno et al., 2012, Van Gheluwe and Dananberg, 2004). Intrinsic medial wedges have been shown to increase medial rearfoot peak plantar pressure by 15-29% (Sweeney, 2016). However, surprisingly a contrasting effect has been shown with extrinsic medial wedges, reducing medial rearfoot peak plantar pressure (Sweeney, 2016). Intrinsic medial heel wedges, which include the Kirby skive, are inside the heel cup and extrinsic medial wedges are under the heel cup. The difference in effect between intrinsic and extrinsic wedges may be due to the heel cup of the extrinsic wedges reducing the amount of compression of the soft tissue under the heel pad, reducing force transfer, more so than the intrinsic wedge (Sweeney, 2016). Additionally, compared to intrinsic wedges, external wedges have a greater distance between the base of the heel cup to the top of the arch, which means the arch is more likely to be loaded and the heel offloaded sooner, offsetting the effect of the wedge itself. Both intrinsic and extrinsic wedges have been shown to shift the COP medially during loading response and midstance (Sweeney, 2016, Van Gheluwe and Dananberg, 2004). A medial shift in the point of application of force during the contact phase of stance would decrease the external eversion moment arm relative to the axis of rotation in the frontal plane, increasing inversion. Increased medial load on the calcaneus will alter rearfoot alignment and can be conceptualised as if the talus and calcaneus were two gear wheels (**Figure 2-1**).

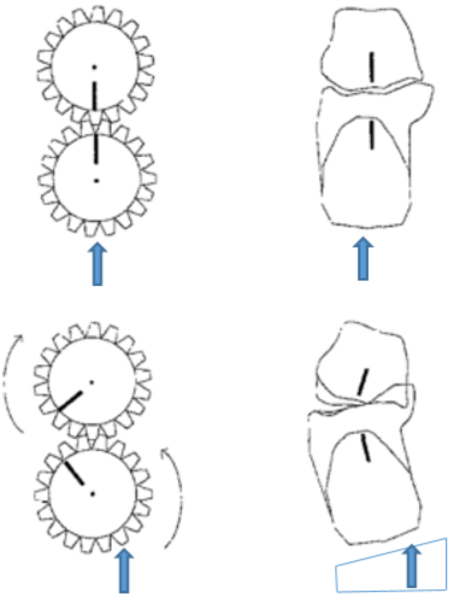


Figure 2-1. The behaviour of the subtalar joint can be likened to two gear wheels. With increased loading on the medial aspect of the calcaneus from a wedge, calcaneal inversion and talar eversion increases. This is like ant-clockwise rotation of the lower gearwheel and clockwise rotation of the upper gearwheel. Original image from (Nester, 1998), adapted in (Sweeney, 2016)

An arch support in FO allows the arch to be loaded when it might not otherwise have been (Williams and Nester, 2010c). An increase in arch height with FO increases peak pressure and total contact area under the medial arch (Sweeney, 2016, Tang et al., 2015). This redistribution of load could affect ankle joint moment by reducing the magnitude of force applied to the rearfoot. There appears to be a ceiling effect for the increase in pressure in the medial midfoot with increase in arch height, in which increases above 3-4 mm do not result in further increases in pressure at the medial arch area above a standard arch support, possibly due to avoiding the pain of compressing soft-tissue (Sweeney, 2016). Increased arch height also shifted the COP more medially during loading response and midstance in the latter work. Greater external support of the medial longitudinal arch may be beneficial in reducing the loading in structures that maintain the arch and thus be clinically beneficial, in the treatment of plantar fasciitis for example.

2.2.7 *Subtalar joint axis location rotational equilibrium theory (a.k.a. tissue stress theory)*

The subtalar joint axis location rotational equilibrium theory (SALRE) is a major theory of FO application. In SALRE position and orientation of the STJ axis itself is important, because it affects the length of the moment arm of the ground reaction force (GRF) and so the magnitude of external moment (Kirby, 2001). According to Kirby, during relaxed calcaneal stance, the normal STJ axis runs from the posterior-lateral aspect of the calcaneus through the first metatarsal head anteriorly (Kirby, 2001). A medial shift in the axis would increase the eversion moment by increasing the eversion moment arm and decreasing the inversion moment arm, causing a net increase in pronation and a pronated foot posture. Supposed axis location is palpable and informs prescription as those with a more medially deviated STJ axis may not benefit from external wedges to increase supination moment (Kirby, 2001). Kirby advocated a medial heel skive, which partially removes the medial heel, creating a varus wedging effect, theoretically increasing the orthotic reaction on the medial side and shifting the centre of external force medially, increasing the supination moment (Kirby, 1992). Foot orthoses could be effective under this model by reducing stress through tissues opposing external moments, which is why this concept is sometimes referred to as the “tissue stress theory” (Harradine and Bevan, 2009). The location of the STJ axis could also theoretically alter internal moment arms of muscles, which would also influence the net joint moment, however evidencing this is limited by difficulty in measuring internal moment arms in the frontal plane. A bone pin study demonstrated incredibly complex foot motion at multiple joints (Lundgren et al., 2008). Motion at the STJ was variable and a lesser magnitude than motion at the talonavicular and calcaneo-cuboid joints. Therefore the SALRE is arguably too simplistic in neglecting moments about the ankle and other joints in the foot.

2.2.8 *Shock attenuation theory*

Mills et al. (2009) discussed shock attenuation in their meta-analysis as another way that FO could reduce external forces, which would affect the corresponding forces experienced by internal structures. Reducing impact forces was proposed as a means of reducing overuse injuries (Mills et al., 2009). Shock-absorbing inserts, without additional material in the medial arch or heel, are sometimes advocated in groups like the military that experience high impact forces with the view to reduce the risk of injuries like medial tibial stress syndrome (MTSS) (Bonanno et al., 2017, Yates and White, 2004). However a meta-analysis found that while there is evidence that FO can be effective in preventing overall injuries (risk ratio (RR) 0.72, 95% CI 0.55 to 0.94) and stress fractures (RR 0.59, 95% CI 0.45 to 0.76), shock-absorbing insoles were not found to be effective in preventing injury (RR ~1.0) (Bonanno et al., 2017). Prefabricated and custom FO designed to reduce shock were not effective in reducing shock magnitude or acceleration rate in treadmill running (Lucas-Cuevas et al., 2017). The shock attenuation paradigm based on shock absorbing material, cannot explain the dose-response effect of anti-pronation FO on ankle moments when material is consistent (Telfer et al., 2013b). Moreover whether small reductions in impact loading are meaningful to injury reduction has been questioned (Nigg et al., 1999).

2.2.8.1 *Medial shin pain (Medial tibial stress syndrome)*

Proposed mechanisms for the therapeutic potential of FO in treating MTSS include changing rearfoot alignment, decreasing impact forces through increased external shock absorption and reducing muscle activity and subsequent traction on the tibia. A pronated foot posture is a possible risk factor for MTSS (Neal et al., 2014, Newman et al., 2013, Winkelmann et al., 2016), and the coupling between pronation at the subtalar joint and tibial internal rotation is well established (Chuter and de Jonge, 2012, Ferber et al., 2005, Fischer et al., 2018, Tiberio, 1987, Resende et al., 2015, Rodrigues et al., 2015, Williams and Nester, 2010c). However as static foot posture only partially explains variance in dynamic foot function (Cavanagh et al., 1997), a pronated foot posture in standing does not necessarily dictate pronation during stance. Larger ranges of internal/external shank rotation and inversion/eversion motion was found in patients with MTSS compared to controls in a study of male soccer players, however there were only eight participants per group (Akiyama et al., 2015).

Foot orthoses may also be beneficial in treating MTSS by increasing shock attenuation due to the possible dampening properties of the FO material. Reducing shock could be beneficial in reducing the force through the tibia, allowing a more balanced rate of bone resorption and bone formation. However this theory is not supported by a review finding that FO are effective for preventing shin pain by 73%, whereas shock-absorbing insoles were not effective in preventing injuries (Bonanno et al., 2017). Alternatively if FO reduced muscle activity and the internal force applied to the tibia via muscle attachments, then this could reduce stress in the fascia and periosteum of the tibia, which may be the site of MTSS (Akuzawa et al., 2016, Ohya et al., 2017).

2.2.9 *Neuromuscular theories*

The “preferred movement path”, “comfort filter” and “muscle tuning” are relatively new paradigms for understanding the mechanism of foot and ankle injury that could be used to explain the mechanism of FO. These were proposed by Nigg and colleagues (2001, 2015, 2017a, 2017b) and were developed particularly in the context of running footwear as well as FO. Mills et al. (2009) also described a “neuromotor control paradigm” in which FO effects are driven by input from the foot sole that modulates muscle activity. Altering muscle activity following a change in load detected by mechanoreceptors in the skin of the foot sole is plausible given the strong synaptic coupling demonstrated between cutaneous afferents from the foot sole and motoneurons supplying leg muscles (Fallon et al., 2005). Additional neuromuscular theories relate to changes in multiple aspects of muscles activity, internal moments and long-term adaptations. Greater exploration of neuromuscular factors is necessary to gain an in depth understanding of the effects of FO on human movement and five theories are discussed below.

2.2.9.1 *The preferred movement path and comfort filter*

In the preferred movement path the athlete has an optimal combination of kinematic and kinetic patterns in running. The preferred movement path is described as the one of "least resistance" (Nigg et al., 2017a, Nigg, 2001, Nigg et al., 2017b) and sometimes as the pattern that involves the least muscle activity and thus metabolic demand (Nigg et al., 2017b). However it has also been stated that with a perturbation, muscle activity can increase and be less than optimal, in order to keep within this preferred pathway (Nigg et al., 2017a). Using a so called "comfort-filter" an individual would select shoes or FO that are the most comfortable for them and allow them to remain in their preferred movement pathway, with least muscular effort (Nigg et al., 2015). As greater comfort in footwear has been related to reduced injury risk, Nigg et al. (2015) suggested that selecting the most comfortable shoe "automatically reduces the injury risk". Exactly how the most comfortable shoes should be the best at preventing injury is not clear. Perhaps the choice of footwear through sensory feedback of perceived comfort facilitates tissue loading and the movement of joints within safe limits. Remaining in a preferred movement pathway could be through comfort and voluntary action, such as footwear and/or FO choice. Alternatively pain signals and cutaneous input could subconsciously lead to altered muscle activation.

Comfort is naturally very subjective and we are not able to accurately predict why one pair of shoes/FO will be comfortable for one person and not another. Nigg et al. (2015) base their argument for the comfort filter paradigm on assessed comfort using a visual analogue scale (VAS) of eight variables after use on a 500 m outdoor course (Mündermann et al., 2001) and categorised ratings of static and dynamic comfort of 13 variables in total (Luo et al., 2009). The measures were deemed reliable owing to consistent ratings from one shoe condition, which was repeated within the testing sessions, however between day reliability and validity of the measures were not guaranteed. No gold standard comfort assessment of footwear and FO exists. The development of a new assessment tool (RUN-CAT) for assessing comfort in running footwear includes a refined list of variables: heel cushioning, shoe stability, forefoot cushioning and forefoot flexibility (Arnold et al., 2018). The assessment was designed with the intention to have discriminative ability between footwear and have clinically meaningful thresholds for change, established through evaluation of both within and between day reliability. The minimal clinically important difference was determined as 9.3-9.9 mm on a 100 mm VAS. Whether immediate assessment of the cushioning of footwear is consistent after long term use is debatable. Further research on the comfort paradigm and its potential mechanisms would benefit from the use of a valid and reliable assessment of comfort, which could include the RUN-CAT.

The preferred movement pathway paradigm for explaining FO effect has appeal in including a neuromuscular component and not treating the body as a purely mechanical system, however the paradigm has yet to be substantiated with much biomechanical evidence. Assessing whether a perturbation facilitates the preferred movement pathway is challenging, as by definition the kinematics and kinetics should remain the same between conditions and so assessment should be made indirectly, like through muscle activation or oxygen consumption (Nigg et al., 2015). Clarification as to what “preferred” actually means may be warranted, whether it be “natural”, perhaps like barefoot running, or “habitual”, what the person has adapted to (Vanwanseele et al., 2018). If an athlete trains on multiple surfaces or in varying footwear, the “preferred movement path” may be hard to define (Vanwanseele et al., 2018). Nigg et al. (2015) argued that whilst a change in footwear, like barefoot to shod, could lead to changes in the amplitude of the movement path, the pattern would stay the same. Yet there is evidence to the contrary, as there is typically a transition from rearfoot to forefoot strike going from shod to barefoot/minimalist shoes in running (Lieberman et al., 2010, Moore et al., 2015). Nigg et al. do however suggest that that the preferred movement path itself could change with training or fatigue (2017a).

Human movement is inherently variable, so perhaps there is not one preferred pathway, but a range of similar and appropriate pathways (Federolf et al., 2018). The concept of variability within the preferred movement pathway is a potentially important in regards to overuse injury. Using different movement patterns to achieve the same goal could mean reduced injury risk, because load patterns would be varied too, spreading the tissues stress across a wider range of structures over time. Or perhaps there is enough muscle redundancy at the foot and ankle that sufficient variability in muscle activation (to reduce injury risk) can be achieved whilst maintaining the same pathway. In fact, how much deviation from the preferred pathway that would be clinically meaningful has not been quantified (Vanwanseele et al., 2018).

2.2.9.2 *Muscle tuning*

In the muscle tuning paradigm, activation of muscles prior to and during heel contact in running acts as a mechanical damper, reducing the vibration in soft tissues (Nigg et al., 2017a, Nigg, 2001). The goal of altered muscle activation would be to avoid resonance of the soft tissue, whereby the natural frequency of the impact force is similar to the natural frequency of the tissue, causing in large oscillations (Nigg et al., 2017a, Nigg, 2001). The resultant vibrations in the soft tissue would cost energy and be uncomfortable according to Nigg et al. (2001). However, there is no evidence to suggest that soft tissue vibration actually leads to injury (Vanwanseele et al., 2018). Furthermore increased muscle activity to increase damping, seems to contradict the idea of the most metabolically efficient preferred movement path (Vanwanseele et al., 2018).

2.2.9.3 *Muscle activity*

A change in muscle activity as a result of a perturbation such as a running shoe or FO could influence injury potential and treatment in a number of ways. Both a change in magnitude and duration of muscle activity would influence the internal loading of tissues, which could clearly impact on injury risk and recovery. Pre-activation of muscles prior to ground contact likely plays a role in modifying stiffness at a joint, as overall stiffness of a system remains stable during impacts, stiffness at a joint will increase to compensate for a more compliant running surface or cushioned footwear for example (Bishop et al., 2006). The effect of pre-activation of muscles on joint stiffness could be indirect, for example via changing the behaviour of tendons and ligaments that also play a role in joint stiffness (Federolf et al., 2018). Joint stability may be influenced by muscle co-contraction (Federolf et al., 2018). An unstable joint is arguably more prone to an acute injury, e.g. partial dislocation, however high levels of co-contraction to maintain stability could lead to high joint compression forces and perhaps increase the risk of other injuries (Hodges et al., 2015). Thus the concept that altered muscle activity can occur in response to footwear/FO with potential preventative and therapeutic benefits is an attractive one that need not be confined to the concept of tissue vibration, but also include the magnitude and duration of loading, joint stability and joint stiffness and compression.

2.2.9.4 *Internal joint moments*

Previous work showed that a medial shift in centre of pressure due to using a FO was not strongly correlated with change in rearfoot kinematics or ankle eversion moment (Sweeney, 2016).The indirect relationship of external force to resultant motion indicates that the foot does not function as a purely mechanical system, but involves a neuromuscular element. Since the net joint moment is balanced by the internal and external moments, changing the external moment by changing the centre of pressure medially would theoretically alter the demand on internal moment opposing it. The central nervous system can modulate muscle activity based on the degree of plantar sensory feedback, enabling control of posture and locomotion (Nurse and Nigg, 2001). Muscle activation causes joint rotation which determines joint position and moment arm of the external force, again influencing the joint moment.

If a perturbation to the foot altered the external and internal joint moment in opposing directions, then the resultant kinematics may appear largely unchanged. For instance, in one study FO had no effect on peak rearfoot eversion or rearfoot eversion excursion in runners, but peak inversion moment and negative work were reduced compared to controls (Williams et al., 2003). The authors suggested that increased external support to oppose eversion from the inverted orthotic reduced the demand on the internal structures. Unfortunately, no measures of muscle activity, such as EMG, were taken. Regardless, the authors speculated that activity of the invertor muscles would have decreased with an inverted FO. One of the main determinants of force produced by the muscle-tendon unit is the level of muscle activation. Reducing muscle activity could have several implications such as delaying the onset of fatigue (Mills et al., 2009). Work by Telfer and colleagues (2013a, 2013b), however, does not support the concept of reduced muscle activity with FO, as a dose-response to FO, which was found in kinematics, kinetics and plantar pressure, but not in EMG variables. However, they did not record EMG from the tibialis posterior muscle, arguably the major foot invertor muscle.

The neuromuscular system is however exceedingly complex and muscle-tendon unit (MTU) force output also depends on the force-length-velocity relationship and so there is not a linear relationship between muscle activity and dynamic force. The force-length relationship of individual muscles fibres follows a bell-shaped pattern due to the degree of overlap of myofilaments in a sarcomere (Maganaris, 2003). And the force-velocity relationship is such that force decreases with increasing shortening velocity in isotonic contractions due to a decreased number of cross-bridge formations between myofilaments (Hill, 1938). Additionally, because many intrinsic and extrinsic muscles exist there is redundancy in the system, a variety of motor control patterns can achieve the same desired functional outcome, such as walking.

2.2.9.5 *Long-term muscular adaptations*

Given the indirect relationship between muscle activity and muscle force, it is worthwhile to look at the response of muscles to a change in loading with FO from additional perspectives. Altered loading can change the strength, size and morphology of muscle and tendon over time. Adaptations are well established from studies on large muscle groups, but are arguably more extreme than with FO use. Currently the long-term effect of modified footwear and FO on soft tissue morphology is unknown and the necessary duration of a FO intervention to see such adaptations is also unknown.

Changes in plantar loading as a result of changing from conventional footwear to minimalist footwear has demonstrated changes in intrinsic foot muscle morphology. A 7-9% increase in foot muscle volume has been reported following a six month transition from traditional running shoes to minimalist shoes (Chen et al., 2016). A minimalist shoe designed to provide less support and cushioning than a regular shoe theoretically increases the demand for active support from muscles, inducing a training effect (Chen et al., 2016). A limitation of this study, as the authors noted, was no record of pre-post foot strike pattern. We therefore cannot attribute change in muscle volume to the shoe properties directly. The increase in plantar pressure from standard running footwear to minimalist footwear is large, around 50% for the mean maximum pressure of the total plantar surface (Warne et al., 2014) and peak pressure at the medial and lateral heel (Moore et al., 2015). Increased plantar pressure with minimalist footwear is again related to foot-strike pattern, with a more plantar-flexed foot with a forefoot strike comes reduced plantar contact area and reduced distribution of load (Moore et al., 2015, Warne et al., 2014). However, the magnitude of pressure changes with FO are more modest than pressure changes between minimalist and conventional footwear. For example intrinsic medial wedges increased medial rearfoot pressure by 15-29% in the study by Sweeney (2016) and regional plantar pressure changed by 0.58%- 2.48% per 2° of external medial wedging in the study by Telfer et al. (2013a). As such, smaller changes in pressure with FO compared to minimalist footwear may result in smaller changes in foot muscle morphology, if at all.

2.2.10 *Sensory theories: mechanoreceptors of the foot sole*

The role of cutaneous mechanoreceptors in the mechanism of FO effect has not been explored. Mechanoreceptors in the skin, muscle spindles, which detect change in muscle length and velocity of length change and Golgi tendon organ receptors, which are receptors in the tendinous aponeurosis that provide feedback motion small changes in force developed by the muscle, contribute to varying extents to proprioception, (Hillier et al., 2015). Cutaneous mechanoreceptors provide feedback to the central nervous system (CNS), on the load applied to the surface of the foot, which can then be used to alter muscle activity, either via changes in voluntary activation or reflex loops. Under the muscle tuning paradigm, sensing impact forces is necessary in order for muscle "tuning" to occur which alters muscle activation and soft tissue vibrations and possibly injury potential (Nigg et al., 2017a). The relationship between muscle activation and plantar cutaneous receptors was evidenced by a change in muscle activity following ice exposure to inhibit plantar cutaneous receptors (Nurse and Nigg, 2001). For instance an earlier onset of activation and increase in amplitude of the biceps femoris was reported with forefoot and whole foot cooling.

Plantar load and subsequent afferent feedback can be manipulated by varying insole materials (Nurse et al., 2005, Vie et al., 2015). In the study by Vie et al. (2015) forefoot sensitivity (ability to detect a mechanical stimulus) of healthy females increased with a month's use of a hard metatarsal pad compared to use of a soft metatarsal pad. Materials affect the shoe-orthotic-foot interaction and even supposedly sham orthoses influence plantar pressure (McCormick et al., 2013). Furthermore, with a custom FO, often more of the plantar surface of the foot is in contact with the FO than would otherwise be the case and the external force is more evenly dispersed (Sesma et al., 2008). This could mean reduced pressure in areas that would have already been in contact with the shoe and a shift in pressure to areas and cutaneous receptors that would normally not be activated. These factors could all alter any neuromuscular role in determining foot function and thereby stresses experienced by foot and ankle structures.

2.2.10.1 *Cutaneous mechanoreceptors*

There are four types of cutaneous mechanoreceptors in (**Figure 2-2**) the hairless skin of the hands and foot sole, known as glabrous skin (Johnson, 2001, Strzalkowski et al., 2015a, Strzalkowski et al., 2015b). Cutaneous mechanoreceptors are defined based on their firing characteristics in response to stimuli and their receptive field sizes (Johansson et al., 1982, McGlone and Reilly, 2010, Strzalkowski et al., 2015a, Strzalkowski et al., 2015b). The receptive field of a mechanoreceptor can be determined by the area of skin in which a monofilament of 4-5 times the threshold for afferent firing. Threshold is determined as the grams force applied by a monofilament that is perceived 75% of the time (Strzalkowski et al., 2015a, Strzalkowski et al., 2015b). Mechanoreceptors with small receptive fields are classified as *Type I* and mechanoreceptors with large receptive fields are classified as *Type II* (Johnson, 2001, Strzalkowski et al., 2015a, Strzalkowski et al., 2015b). Secondly, mechanoreceptors are classified as *slow adapting* (SA) when they continue to fire with sustained indentation, or *fast adapting* (FA) when they fire only when a stimulus is initially applied or subsequently removed (Kennedy and Inglis, 2002, McGlone and Reilly, 2010). Slow adapting type I (SAI) receptors are known as Merkel discs, slow adapting type II (SAII) receptors are called Ruffini endings, fast adapting type I (FAI) receptors are Meissner corpuscles and fast adapting type I (FAII) receptors are Pacinian corpuscles (Johnson, 2001, McGlone and Reilly, 2010).

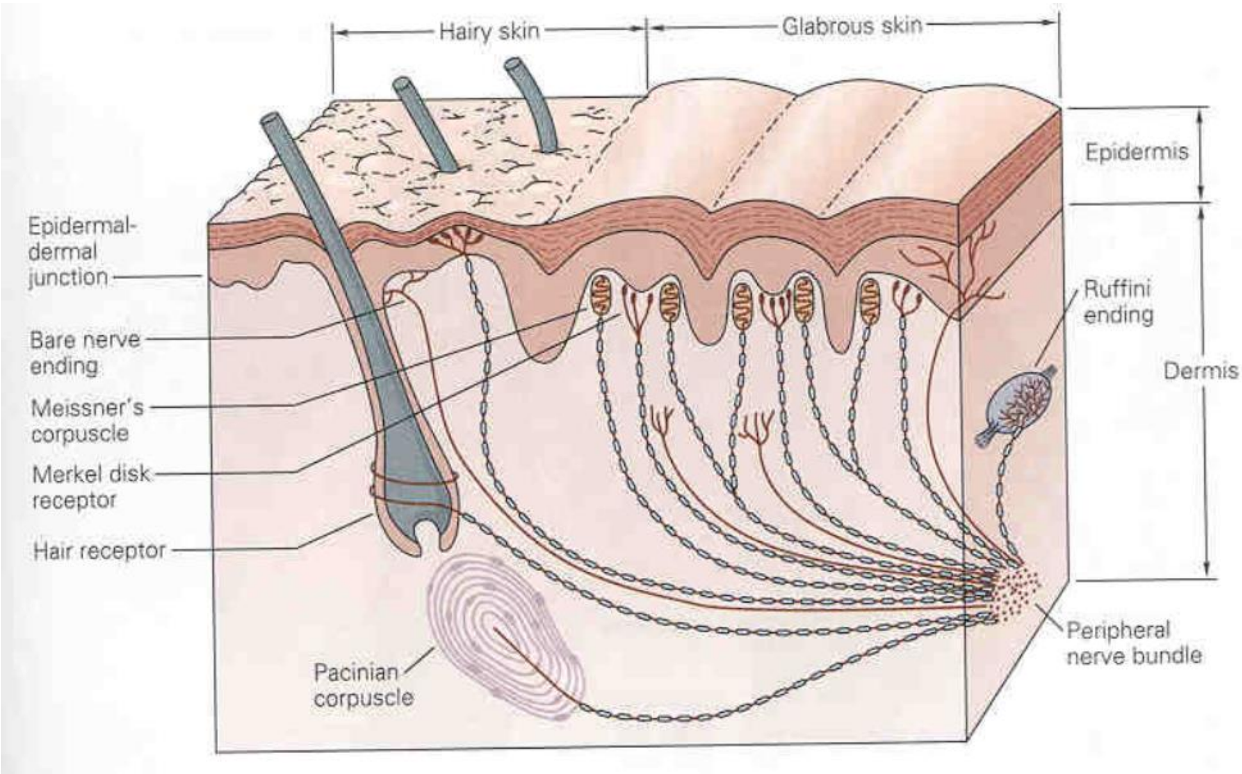


Figure 2-2. Layout of mechanoreceptors in skin (Kandel et al., 2000)

Each of the different classes of cutaneous mechanoreceptor code for different information from the environment and are sensitive to a specific range of vibration frequencies (Strzalkowski et al., 2015a). The slow adapting receptors are most sensitive to low frequencies, 2-32 Hz for SAIs and <8 Hz for SAIIs (Strzalkowski et al., 2015a). The SAIs code for information regarding spatial perception, texture, pressure, edges and curvature (Johnson, 2001), whereas the SAIIs are particularly sensitive to skin stretch (Johansson et al., 1982, Johnson, 2001) owing to their spindle like shape with an inner core suspended by collagenous fibres (Chambers et al., 1972). On the other hand, the fast adapting receptors respond better to high frequencies, 8-64 Hz for FAIs and 64- 400 Hz for FAIIs (Johansson et al., 1982). The FAIs are sensitive to shear forces, including the perception of movement across the skin and they code for rate or velocity of skin indentation and flutter, helping us detect slips, while FAIIs code acceleration and vibration (Johansson et al., 1982, Johnson, 2001, Strzalkowski et al., 2015a).

Given cutaneous mechanoreceptors sense load on the foot sole and FO aim to alter plantar load, mechanistic studies on the nature of mechanoreceptors could be beneficial in understanding the mechanism of FO effect. The activity and distribution of cutaneous mechanoreceptors has been studied using the technique of microneuography (Kennedy and Inglis, 2002, Strzalkowski et al., 2015b). Microneurography involves inserting an electrode directly into a peripheral nerve to record afferent feedback to the CNS (Hagbarth and Vallbo, 1967). Afferent firing threshold, the minimum force necessary for afferent discharge, has been shown to be strongly related to perceptual threshold for fast adapting receptors, while afferent firing threshold was higher than perceptual threshold for slow adapting receptors (Strzalkowski et al., 2015b). Microneurography studies have found cutaneous mechanoreceptors to be spread extensively across the foot sole, with the majority of the receptors identified being FAIs (Kennedy and Inglis, 2002, Strzalkowski et al., 2015b, Strzalkowski et al., 2018) (**Figure 2-3**). A greater density of mechanoreceptors has been found in the toes than the arch and metatarsals (Strzalkowski et al., 2018). Additionally a mediolateral gradient in the forefoot and arch has been established, with the greatest density of mechanoreceptors on the lateral border than the medial border (Strzalkowski et al., 2018).

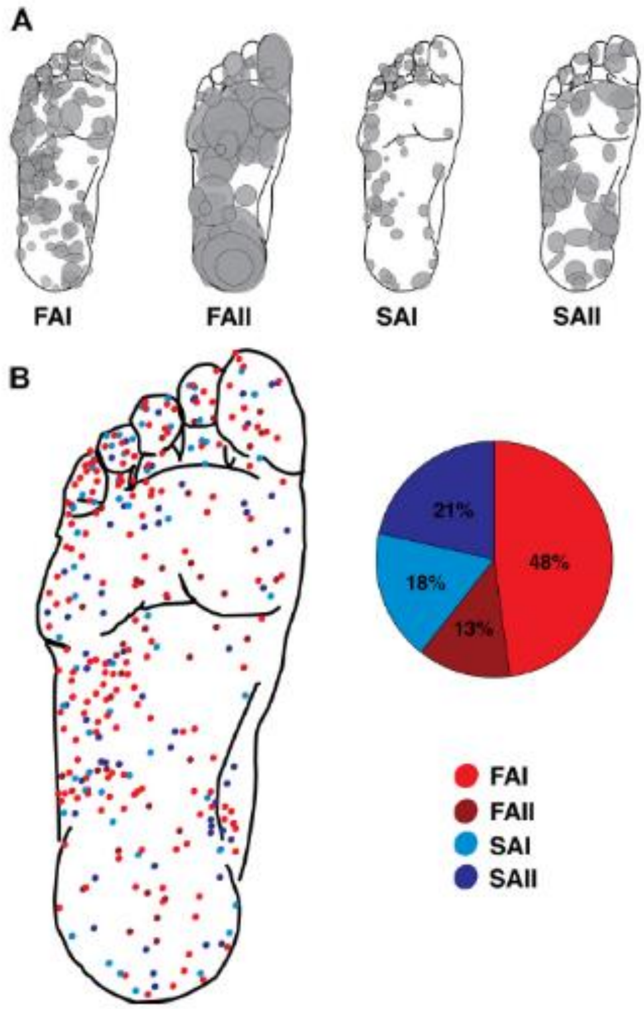


Figure 2-3. Receptive fields of mechanoreceptors. A) Estimated receptive field sizes and locations for fast adapting type I (FAI), fast adapting type II (FAII), slow adapting type I (SAI) and slow adapting type II (SAII) mechanoreceptors. B) Foot sole map shows the centre of receptive fields and pie chart shows the percentages of each mechanoreceptor class in consolidated microneurography data from (Strzalkowski et al., 2018)

Foot orthoses with increased arch support change loading at the medial longitudinal arch, which could influence afferent feedback to the CNS given that the skin is particularly sensitive in that region. Custom FO have been shown to increase peak pressure in the medial arch by 15-23% (Hodgson et al., 2006, McCormick et al., 2013) and increase contact area in the medial arch by 65% (McCormick et al., 2013). Fewer mechanoreceptors have been identified in the medial arch than other regions of the foot (Strzalkowski et al., 2018). However the medial arch area of skin is partly innervated by the saphenous nerve (Moore and Agur, 2007), and not the tibial nerve, which Kennedy and Inglis (2002) and Strzalkowski et al. (2015b) recorded from. As such fewer receptors found in the arch than other areas of the foot to date may be a result of the measurement technique not a true representation of relative mechanoreceptor numbers (Strzalkowski et al., 2018). Nevertheless sufficient data exists to show that the medial arch is the most sensitive area of the foot sole, as it has both the lowest afferent firing threshold and lowest perceptual threshold (Strzalkowski et al., 2015b). The thinner skin in the medial arch, compared to the heel for example, may not entirely account for the medial arch being the area of greatest sensitivity on the foot, because skin thickness and hardness are only strongly correlated to perceptual threshold in FAI receptors (Strzalkowski et al., 2015a).

Changing the external loads applied to the sole of the foot, such as wearing FO, would not change the number or distributions of cutaneous mechanoreceptors, but could potentially alter the firing thresholds and/or the relative weighting of feedback from given receptors in the CNS. An increase in hand sensitivity following anesthetising the forearm is an example of cortical plasticity, which likely arises from increased hand area in the primary somatosensory cortex (Björkman et al., 2009). Furthermore, following space flight it appears that the CNS may decrease the weighting of slow adapting receptors in a response to unloading and increase the weighting of fast adapting receptors when vestibular deficits are present (Lowrey et al., 2014). An increase in loading of a region with a FO intervention could increase the relative weighting given to receptors from that region of the foot, and lead to an increase in sensitivity. Alternatively, increased loading of a region could decrease skin sensitivity if the receptors become saturated.

2.2.11 *Conclusion*

More mechanistic studies on the effect of FO on foot biomechanics are required to explore the theories discussed above and ultimately improve clinical efficacy of FO. Further research into the effect of FO on neuromuscular factors is particularly warranted (Mills et al., 2009). This PhD is an extension of a previous thesis (Sweeney, 2016) that found that systematically manipulating external forces applied to the foot with FO were not well correlated with joint moment and kinematic effects, suggesting a likely contribution of neuromuscular factors to FO mechanisms. The following section reviews soft tissue structures of the lower limb that have the potential to be affected by FO. Additional external perturbations like footwear and taping could affect soft tissue and are thus reviewed together with FO. The literature review of the evidence that changing external forces at the foot and ankle can influence muscle activity is in journal paper format in the next chapter.

2.3 Background of muscle function and EMG and tendon mechanics

The following is an overview of the general principles of muscle activation and muscle-tendon mechanics, which is necessary to understand how to study muscle function and how that might change with FO. Muscles and connective tissue of the foot and ankle are discussed in order to establish the soft tissues most likely to be affected by FO and worthy of investigation in the thesis.

2.3.1 *Muscle activation*

Electromyography is the technique of recording and analysing the electrical signal from contracting muscles and can be used in medical research, rehabilitation, ergonomics and sport science (Konrad, 2005). EMG data allows for the study of muscle activity during gait and allows us to monitor changes in the relative amount of and timing of activation in pathology or as a result of an intervention, like FO.

2.3.1.1 *The motor unit and muscle contraction*

A motor unit (MU) is defined as a motor neuron and all the muscle fibres that it innervates (Liddell and Sherrington, 1925). The EMG signal is a summation of all the motor unit action potentials (MUAPs) within the recording area of the electrode (Hakonen et al., 2015, Konrad, 2005). The process of excitation-contraction coupling is described by Kandel et al. (2000). An action potential occurs when an increase of sodium ions flows into a cell, depolarizing the membrane potential and generating an electric current. The sodium-potassium pump then repolarises the membrane potential after which there is a period of hyperpolarisation, where the membrane potential is below the resting level and it is not possible to generate another action potential. Alpha motor neurons exit the anterior horn of the spinal cord via the ventral root. The central nervous system triggers an action potential in the motor neuron causing a wave of excitation that travels down the axon of the neuron towards the motor end plate. At the neuromuscular junction there is an influx of calcium ions into the nerve cell, a release of vesicles containing the neurotransmitter acetylcholine and proteins that help bind the vesicles to the presynaptic membrane (Kandel et al., 2000). Acetylcholine crosses the synaptic cleft, to bind to receptors on the postsynaptic membrane, which causes an action potential that travels in all directions and through T-tubules in the muscle (Farina et al., 2004, Moritani et al., 2004). Depolarisation of the postsynaptic membrane leads to a cascade of events that produce muscle contraction (excitation-contraction coupling). Calcium ions are released in the sarcolemma, calcium binds to troponin which unlocks the binding site of tropomyosin on actin filaments, to allow cross-bridges to form between actin and myosin, resulting in contraction of the muscle and force generation.

2.3.1.2 *Muscle twitch*

The action potentials in the individual muscle fibres of a MU are considered a functional unit and their summation is the MUAP, despite imperfect synchronisation of the action potentials due to different conduction times in the fibres (Trontelj et al., 2004). The number of muscle fibres innervated by a single MU (the innervation ratio), varies between muscles (Gath and Stalberg, 1981). The MUAP produces a twitch (contraction) of the muscle fibres in which a small amount of force is produced for approximately 50–150 ms (Moritani et al., 2004, Staudenmann et al., 2010). The shape of the twitch will depend on the MU fibre type **(Figure 2-4)**, as fast twitch fibres have a shorter time to peak force than slow twitch fibres (Moritani et al., 2004, Merletti et al., 2004). Multiple MUAP in quick succession will result in the overlapping, or superimposition, of the twitches and a sustained tetanic force, which enables the movement of our joints (De Luca, 2007). The summation of the individual MUAP from multiple MU is the recorded EMG signal. The amplitude, frequency and onset/offset of the EMG signal are analysed in the study of human movement.

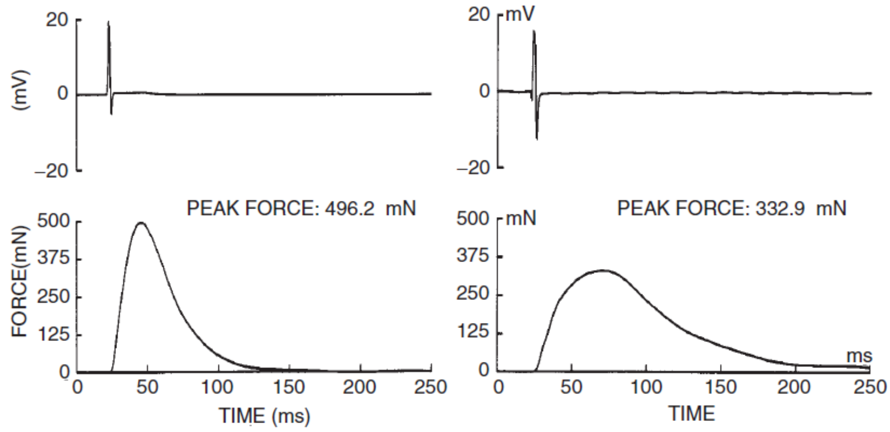


Figure 2-4. The motor unit action potential and corresponding twitch for a fast twitch fibre (left) and slow twitch fibre (right). From (Moritani et al., 2004)

2.3.1.3 *EMG amplitude*

The amplitude of the EMG signal indicates the magnitude of muscle activation, which is related imperfectly to force production. The number of MUs recruited, the size of the MUs and the firing frequency of each MU dictates the amount of force produced as well as the amplitude of the EMG signal although these relationships are non-linear (De Luca and Erim, 1994, Staudenmann et al., 2010). The number of MU recruited increases with neural drive, however according to Henneman's size principle (Henneman, 1957), smaller, slow twitch, MU are recruited at low force levels before larger, fast twitch, MU at greater force levels (Moritani et al., 2004, Staudenmann et al., 2010). An increase in MU firing rate will not linearly increase EMG amplitude or force level due to an increase in the phase cancellation (destructive rather than constructive superimposition of MU) (Staudenmann et al., 2010). Additionally due to the changing in force output with muscle tendon unit length and contraction velocity (Lieber and Friden, 2000), there is not a linear relationship between EMG amplitude and force during dynamic contractions (Staudenmann et al., 2010).

2.3.1.4 *Frequency*

The frequency content of EMG is analysed when interested in muscle fatigue. A Fourier transform produces a power spectrum from which the mean or median power frequency can be obtained (Konrad, 2005, Moritani et al., 2004, Merletti et al., 2004). With fatigue the mean power frequency decreases, partly as a consequence of decreased conduction velocity, and MU recruitment and amplitude increase in an effort to sustain force in submaximal contractions (Merletti et al., 2004). Fibre type influences fatigue as fast twitch Type IIb fibres fatigue quickly, fast twitch Type IIa are relatively fatigue resistant and slow twitch, Type I fibres are the most fatigue resistant (Merletti et al., 2004, Moritani et al., 2004). The EMG study in this thesis investigated the immediate effects of foot orthoses in healthy individuals walking over only a short period of time, so fatigue was not considered a factor.

2.3.1.4.1 *Wavelet analysis*

A form of wavelet analysis can be used to calculate EMG intensity in two frequency bands, where intensity refers to the power of the EMG signal and indicates the timing and strength of muscle activation at various frequencies (von Tscharner, 2000, Mundermann et al., 2006). Wavelet analysis has been used to study the effect of FO on muscle activity in which increases in EMG intensity with FO were greater in the high frequency band than the low frequency band (Mundermann et al., 2006). These high frequency signal components are believed to correspond to firing of fast twitch muscle fibres and low frequency components to slow twitch fibres (Mundermann et al., 2006, Nurse et al., 2005, von Tscharner and Nigg, 2008). However, this idea has been contested (Enoka, 2008, Farina, 2008) and the general consensus is that changes in the spectral properties of the EMG signal cannot be directly attributed to a change in the activation of MUs of a specific fibre type. Therefore because wavelet analysis lacks validation, we are unable to draw any definitive conclusions about the implications of FO studies that have used it and consequently wavelet analysis was not used as a method in this thesis

2.3.1.5 *Timing and duration*

The onset and duration of muscle activity is also of interest in the study of human movement, for instance concerning motor control patterns and the duration of internal loading of a muscle tendon unit. The determination of muscle onset can be purely visual, which although subjective can make allowances for random baseline noise easily (Hodges and Bui, 1996). A common automatic means of establishing onset of EMG activity is based on the point when the EMG signal increases above a certain threshold, usually a predetermined multiple of standard deviations above average baseline activity (Carter and Gutierrez, 2015, Hodges and Bui, 1996, Konrad, 2005), or above a percentage of the peak value (Franettovich et al., 2008, Hodges and Bui, 1996, Kelly et al., 2010, Konrad, 2005). Other more complicated algorithms to detect EMG onset include the approximated generalized likelihood-step, k-means and Teager-Kaiser energy operator (Carter and Gutierrez, 2015, Solnik et al., 2010). Computer based algorithms may be less time consuming than visual detection of EMG onset, however manually inspecting the output value has been recommended (Hodges and Bui, 1996, Konrad, 2005). If there is artefact in the baseline signal it would lead to an overestimation of threshold and a delayed onset time, which may not be accounted for with automated algorithms. The degree of smoothing of a filtered signal will also impact on onset detection and thus is an important consideration (Hodges and Bui, 1996). As it is commonly used and simplistic, a threshold method was used for the determination of onset in Chapter 6. Offset of muscle activity can be determined in the same manner as onset detection and the time between the two points gives the overall duration of muscle activity.

2.3.2 *Muscle-tendon mechanics*

The internal moment generated by a MTU is determined by the product of moment arm and force. The moment arm of a MTU is defined as the perpendicular distance from the joint centre of rotation to the MTU line of action (Maganaris, 2004). The longer the moment arm, the larger the moment generation capacity and also the greater potential muscle-tendon displacements/velocities of the MTU (Maganaris, 2004). Internal moment arm is not constant and can change with joint angle (Lieber and Friden, 2000). Moment arms are often quantified *in vitro*, using geometric measurement, tendon-joint excursion and direct load measurement as outlined by An et al. (1984). In the geometric method muscle and tendon paths can be obtained using x-rays. Orientations and moment arms of the lines of action of the muscles can then be calculated based on the joint co-ordinate system. The direct load measurement involves applying a known weight to a single muscle and registering the moment at the distal segment, from which moment arm can then be obtained. The tendon excursion method uses the relationship between work and tendon excursion to joint rotation ratio and has the advantage that prior knowledge of the location of the centre of joint rotation is not necessary (An et al., 1984, Lee and Piazza, 2008, Maganaris, 2004). Mean moment arms calculated using the tendon excursion method for select muscles at the STJ are presented in **Table 2-1**.

It must be noted that moment arm can change with the presence of muscle contraction, due to factors such as tendon and ligament compliance, so moment arms estimated with cadavers are only an approximation of moment arms *in vivo* (Lee and Piazza, 2008, Maganaris, 2004, Maganaris et al., 1999). Moment arm can also be measured directly *in vivo* using x-ray or magnetic resonance imaging or estimated using the tendon excursion method (Lee and Piazza, 2008, Maganaris, 2004). However the tendon-excursion method is particularly challenging in the frontal and transverse planes, involving a number of assumptions and a means to a control frontal plane movement which has questionable applications to foot function in gait. Consequently it is currently not practical to assess whether using FO directly alters internal moment arms in locomotion. However an awareness of moment arm lengths is necessary in order to determine which muscles contribute the most to particular movements and the most relevant in the study of the effects of FO (2.4 Foot and ankle soft tissues relevant to foot orthoses effects).

Table 2-1. Mean frontal plane moment arm length (mm) at the subtalar joint from (Klein et al., 1996)

Muscle	Mean	Std ind	Std rot
Triceps surae	– 5.3	7.4	10.6
Peroneus longus	21.8	4.3	3.4
Peroneus brevis	20.5	3.9	2.8
Tibialis anterior	– 3.8	4.4	—
Tibialis posterior	– 19.2	3.6	1.3
Flexor hallucis longus	– 7.8	3.0	3.9

std ind: interindividual standard deviation; std rot, standard deviation related to rotation; units, mm; a positive sign stands for an eversion moment; a negative one indicates an inversion moment.

The capacity of a muscle to generate force is naturally in part determined by its size, which can be measured with respect to its thickness, or cross sectional area (CSA) (Blazevich et al., 2009). However muscle function is also somewhat dependent on its structural properties in relation to the individual muscle fibres (Lieber and Friden, 2000). The greater the pennation angle, the angle of the muscle fibres in relation to the line of pull of the MTU, the greater amount of contractile tissue for a given muscle volume, the closer muscle fibres can operate with regards their optimum length and so the greater the force generating potential (Blazevich et al., 2006). Muscles with short muscle fibres and a high pennation angle are optimised for generating high forces, while muscles with long muscle fibres and low pennation angles are suited for a large range of joint excursion (Lieber and Friden, 2000). The ratio of muscle fibre length to muscle length and tendon length impacts MTU force generation capacity due to its influence on sarcomere shortening velocity (Lieber and Friden, 2000, O'Brien et al., 2010, Zajac, 1988). Generally the longer the muscle fibre, the greater the number of sarcomeres in the fibre, so the lesser the displacement per sarcomere and the lesser the sarcomere velocity and the higher the sarcomeres operate on the force velocity curve (Lieber and Friden, 2000). Conversely sarcomere displacement increases with tendon length in order to overcome the compliance of the tendon (Lieber et al., 1992, Zajac, 1988). Force generated in a given contraction will also be influenced by the extent of muscle activation determined by neural drive as discussed above.

The physiological cross-sectional area (PCSA) is directly proportional to the force generation capacity of the muscle (Lieber and Friden, 2000). The PCSA can be defined as follows:

$$\text{PCSA (mm}^2\text{)} = \frac{\text{muscle mass (g)} \cdot \text{cosine } \theta}{\rho \text{ (g/mm}^3\text{)} \cdot \text{fiber length (mm)}}$$

where ρ represents muscle density (1.056 g/cm^3 for mammalian muscle) and θ represents surface pennation angle (Lieber and Friden, 2000, Powell et al., 1984). The PCSA differs from the anatomical cross-sectional area in that it considers the pennation angle of muscle fibres and so by using the cosine component of the equation accounts for the fact that muscle fibres are orientated at an angle to the line of action of the MTU (Lieber and Friden, 2000).

2.3.2.1 *Muscle adaptations to a change in loading*

A muscle can respond to a prolonged change in loading with a change in size, structure or with altered neurological activation (Folland and Williams, 2007). Hypertrophy is an increase in muscle size/thickness through increased myofibrillar size and the activation of satellite cells and is a well-established response to resistance training (Alegre et al., 2014, Folland and Williams, 2007). Atrophy on the other hand is a decrease in muscle size with disuse. After five weeks of bed rest the medial gastrocnemius (MG) decreased in thickness by 12.2 ±8.8% ($p < 0.05$) and the vastus lateralis by 8.0 ±9.1% ($p < 0.005$) in a study by de Boer et al. (2008). However interestingly, no change in muscle thickness occurred in the non-antigravity muscles of tibialis anterior (TA) and biceps brachii in the same study. Pennation angle is a modifiable structural property of muscle, it can increase with strength training and decrease with bed rest (Alegre et al., 2014, de Boer et al., 2008, Morse et al., 2007, Reeves et al., 2004). While many examples of muscle adaptations in the literature are a result of a change in activity load, muscle adaptation could also occur through a change in an external mechanical load applied to the foot, such as via FO. Intrinsic foot muscles can indeed change size in response to modified loading of the foot with a change in footwear (Bruggemann et al., 2005, Johnson et al., 2016). Therefore the approach to studying muscle function in Chapter 7 was to study the effect of FO on muscle morphology by measuring thickness and cross sectional area using ultrasound.

2.3.2.2 *Tendon*

Free tendon and the aponeurosis within the muscle transmit muscle force to the bone which causes rotation about a joint and tendons also facilitate efficient contraction by storing and releasing energy (Magnusson et al., 2008a). The behaviour of a tendon is determined by its mechanical and material properties. Stiffness is a mechanical property of the tendon structure overall, which is determined by the curvilinear slope of the force-elongation curve (Seynnes et al., 2015). The greater the tendon the length the greater its compliance (or the less its stiffness) (Maharaj et al., 2016). Stiffness can be scaled to tendon size, representing the material property Young's Modulus, from the stress-strain curve (Maganaris and Paul, 1999). Young's Modulus or stiffness of the tendon affects the rate of force transmission, where on the force-length curve the muscle fibres have to operate and the chance of strain injury (Narici et al., 2008). An acute, strain injury occurs when elongation of the collagen complex that makes up the tendon occurs. When elongation moves beyond the reversible, elastic region, of the elongation-force curve, to the plastic region, a rupture occurs (Robi et al., 2013). Tendinopathy on the other hand is a chronic, overuse injury, that could arise through a number of proposed mechanisms, including overloading, inflammation and dysregulation of enzymes (Millar et al., 2017). Foot orthoses are used in the treatment of a number of tendinopathies, namely Achilles tendinopathy, patellar tendinopathy and tibialis posterior tendon dysfunction. A change in loading of the foot and ankle with FO could result in tendon adaptations that are clinically beneficial.

Increased tendon stiffness with increased loading has frequently been reported in the Achilles and patellar tendons, particularly with protocols with a high volume of loading or long duration of muscle contraction (Arampatzis et al., 2007, Kubo et al., 2001, Kubo et al., 2003, Reeves et al., 2003a, Reeves et al., 2003b, Seynnes et al., 2009), however a lower intensity programme such as a six month walking intervention has been shown to be insufficient to change tendon stiffness (Kubo et al., 2008). The evidence for tendon hypertrophy with increased loading in humans is less clear. Tendon hypertrophy may only occur after a particularly long period of increased loading (Folland and Williams, 2007, Heinemeier and Kjaer, 2011) and/or loading at high magnitudes (Arampatzis et al., 2007). In cases where tendon hypertrophy has occurred with increased loading, the increase in CSA has been region specific within the tendon of interest (Arampatzis et al., 2007, Kongsgaard et al., 2007, Seynnes et al., 2009). In the study by Seynenes et al. the increase in patellar CSA with resistance training was not related to the increase in muscle strength or muscle PCSA unlike the increase in stiffness, leading the authors to suggest that the mechanisms behind increased tendon stiffness and tendon hypertrophy are different. There is evidence of reduced tendon stiffness and Young's Modulus with unloading, however whether tendon CSA also decreases with unloading is unclear (Maganaris et al., 2006, Reeves et al., 2005). While the effect of FO on tendon properties was not studied directly in this thesis, the relationship between connective tissue in series or in parallel with muscle is important in understanding the implications of potential changes in muscle function with FO.

2.3.3 *Conclusion*

The effect of FO on muscle function and morphology could be studied using EMG and ultrasound respectively. In Chapter 3 the existing evidence for changes in EMG with FO is reviewed. Chapter 6 focuses on the immediate effects of FO on muscles by investigating the effect of FO geometry on the EMG amplitude and onset times. The rationale for the muscles chosen in Chapter 6 is presented in the next section. In Chapter 7 the effect of FO on soft tissue morphology in the long term is studied using ultrasound measures of thickness and cross sectional area.

2.4 Foot and ankle soft tissues relevant to foot orthoses effects

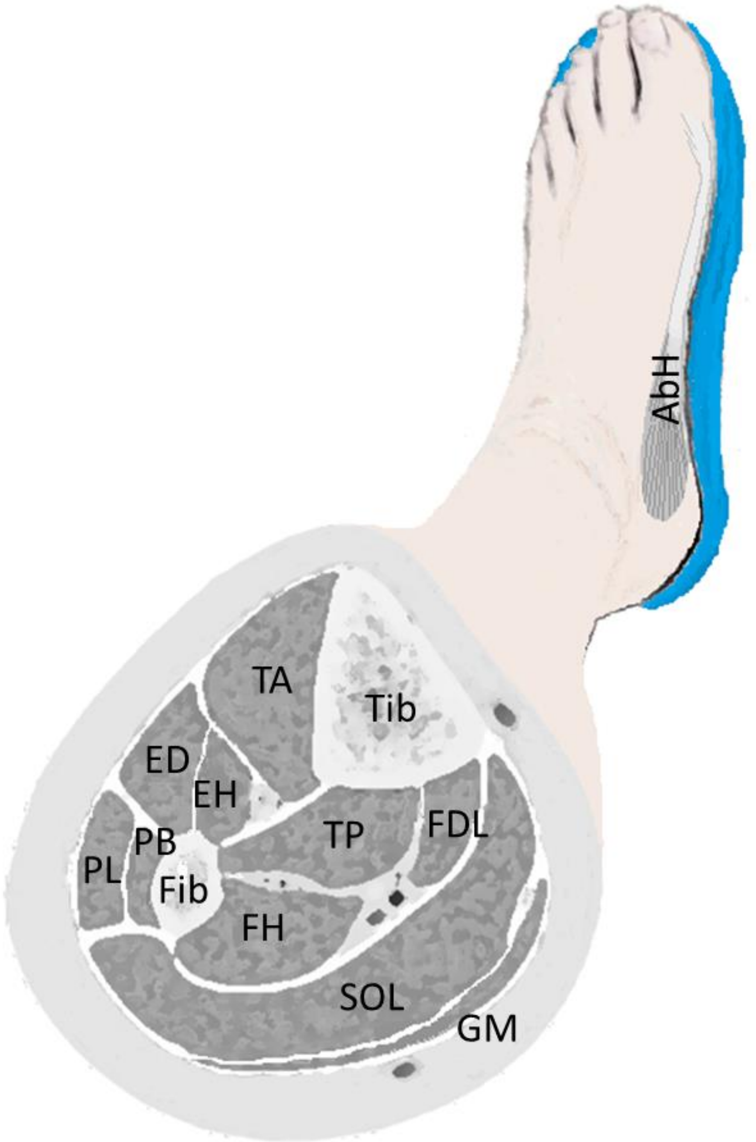


Figure 2-5 Muscles of the shank and abductor hallucis (ABH) in the foot. ED: extensor digitorum; EH: extensor hallucis; Fib: fibula; FDL: flexor digitorum longus; FH: flexor hallucis; GM: gastrocnemius; PL: peroneus longus; PB: brevis; SOL: soleus; Tib: tibia; TA: tibialis anterior; TP: tibialis posterior. Adapted from (Rha et al., 2014)

2.4.1 *Tibialis anterior*

Tibialis anterior is a superficial muscle on the anterior-lateral aspect of the shank that originates on the upper lateral tibia and interosseous membrane and inserts on to the medial plantar surface of the first cuneiform (Hamill and Knutzen, 2009). Onset of TA activity is at the end of stance, as the muscle is active in the swing phase to enable toe clearance, activity peaks at the end of swing phase/initial contact (Bovi et al., 2011, Hof et al., 2002, Hunt et al., 2001, Onmanee, 2016, Van Hedel et al., 2006, Winter and Yack, 1987). The TA is the primary dorsiflexor of the ankle joint. From cadaveric data, the mean dorsiflexor moment arm for TA is 32±6.3 mm, with a PCSA of 10.9± 3.0 cm^2 (Klein et al., 1996, Ward et al., 2009). The tendon of TA is long and begins mid-shank, it changes direction by curving around the malleolar retinacula (Hamill and Knutzen, 2009, Petersen et al., 2000). Ruptures of TA tendon are rare (Aydingöz and Aydingöz, 2002, Gallo et al., 2004, Kausch and Rütt, 1998, Patten and Pun, 2000, Rimoldi et al., 1991).

The TA has a secondary role in inversion. From cadaveric data TA has an inversion moment arm of 3.8 ± 4.4 mm (Klein et al., 1996). An *in* vivo study using ultrasound and a rotating platform showed the TA to have a slight eversion moment arm when the foot was everted 15° and an inversion moment arm when the foot was neutral or inverted 20° (Lee and Piazza, 2008). Moment arm lengths were similar at rest to during light and maximum voluntary contraction. When the foot was neutral the inversion moment arm was approximately 3 mm and when the foot was everted 20° the inversion moment arm was approximately 10 mm. When TA activity is around its greatest at heel contact in walking, the rearfoot is inverted, but only by around 5°, so of a much lesser magnitude than the 20° tested experimentally with the tilting platform (Nester et al., 2014).

Foot-type has been shown to significantly and independently explain 7.1% ($p = 0.017$) of anterior-posterior TA tendon thickness (Murley et al., 2014a). Flat-arched feet (normalized navicular height truncated ratio of less than 0.18 and an arch index ratio of greater than 0.32) have been associated with a thicker TA than those classified as having a normal arched foot type. Greater TA activation at heel contact into the contact phase of stance has been reported in those with pes planus compared to a normal foot posture (Hunt and Smith, 2004, Murley et al., 2009c). Peak TA activation was 19% greater ($p < 0.001$) in those with flat-arched feet (arch index or navicular height measurement greater than two standard deviations from mean values obtained for the normal-arched group) compared to normal-arched feet (Murley et al., 2009c). While it could be speculated that greater TA activation in early stance with a flat-foot could relate to TA resisting foot pronation in this phase, it seems unlikely given that there was a trend towards *less* tibialis posterior activity in early stance in the flat-arched group in the same study and TP is a much more effective invertor (better able to resist pronation).

A FO designed to reduce the eversion moment at the STJ would theoretically decrease the inversion moment required from the invertor muscles that limit eversion. As TA has an inversion moment arm at initial contact, it is conceivable that FO might reduce TA activity.

2.4.2 *Extensor digitorum longus and extensor hallucis longus*

The extensor digitorum longus (EDL) extends the lesser toes and extensor halluces longus (EHL) extends the hallux. The EDL has its origin on the lateral condyle of the tibia and proximal three quarters of the anterior aspect of the interosseous membrane and its insertion on the middle and distal phalanges of the lesser toes (Moore and Agur, 2007). While EHL originates on the middle portion of the anterior aspect of the fibula and interosseous membrane and inserts on the dorsal aspect of the distal phalanx of the hallux (Moore and Agur, 2007). Both EDL and EHL assist TA in dorsiflexion and understandably have similar activation patterns to TA (Hunt et al., 2001, Reeser et al., 1983, Winter and Yack, 1987). However, EDL and EHL have much smaller PCSAs than that of TA (5.6 ±1.7 cm^2 and 2.7 ±1.5 cm^2 respectively) (Ward et al., 2009) and as such are less fundamental in overall foot function than TA.

2.4.3 *Flexor hallucis longus*

The flexor hallucis longus (FHL) is a deep shank muscle located posteriorly and laterally, originating on the distal two thirds of the fibula, the interosseous membrane and the posterior crural septum dividing it from the peroneals (Bianchi and Martinoli, 2007). The insertion of FHL is the distal phalanx of the hallux (Kirane et al., 2008). The FHL is active in mid-late stance, stabilizing the hallux (Reeser et al., 1983) and contributing to ankle plantarflexion (Ward et al., 2009). The plantar flexion moment arm at the ankle joint of FHL is second only to the triceps surae, the mean ± SD of FHL in the study by Klein et al. was 26.6 ± 4.4 mm (1996). In the same study the mean inversion moment arm of FHL was 7.8 ±3.0 mm (vs. 19.2 ±3.6 mm of tibialis posterior (TP)). The FHL is also thought to dynamically support the MLA (Bianchi and Martinoli, 2007, Edama et al., 2017).

2.4.4 *Flexor digitorum longus*

The flexor digitorum longus (FDL) originates on the medial aspect of the mid and distal third of the tibia (Bianchi and Martinoli, 2007, Edama et al., 2017). The insertion of FDL is the plantar surface of the distal phalanges of the lesser toes (Harradine and Bevan, 2009). Similar to FHL, FDL is active mid-late stance (Reeser et al., 1983). By definition FDL flexes the lesser toes, but also contributes to plantar flexion (Ward et al., 2009). If the traction theory of the cause of MTSS is to be believed, whereby repeated muscular contraction results in stress on the periosteum of the tibia bone, then FDL activation may be important in the development and treatment of MTSS as FDL attaches on the medial border of the tibia (Edama et al., 2017). However MTSS may instead be a direct bone stress injury of the tibia (Edama et al., 2017, Moen et al., 2009, Yates and White, 2004). As MTSS is most common in cases of high impact loading, as in the military, it is perhaps not worth investigating in a study of the effect of FO on muscle activity from a general mechanistic perspective (as in this thesis) rather than with respect to MTSS as a specific pathology. In one study that did explore the effect of FO on FDL, there was no difference in FDL between the footwear, FO and barefoot conditions in walking (Akuzawa et al., 2016).

2.4.5 *Tibialis posterior*

The tibialis posterior (TP) is the deepest muscle in the shank, originating from the posterior surfaces of the tibia and fibula and the adjoining interosseous membrane (Bloome et al., 2003, Semple et al., 2009). Distally the TP tendon passes posterior to the medial malleolus and above the sustentaculum tali of the calcaneus and forms anterior, middle and posterior bands which insert at multiple locations (Bloome et al., 2003, Hamilton et al., 2008, Semple et al., 2009). The anterior band is the largest of the three bands, around 65% of the width of the tendon, and inserts on to the navicular tuberosity, often with a fibrocartilaginous or bony sesamoid at the insertion site (Bloome et al., 2003, Semple et al., 2009). The middle and posterior bands insert on to the plantar aspect of the medial and lateral cuneiforms, cuboid and second-fourth metatarsals (Bloome et al., 2003, Semple et al., 2009). Variation in further insertion sites have been observed in some cadavers, including the spring ligament, the sustentaculum tali, the base of the fifth metatarsal, the peroneus longus tendon near the base of the first metatarsal, the flexor hallucis brevis muscle and abductor hallucis muscle (Bloome et al., 2003). It is thought that the presence of multiple insertion sites enables TP to assist in supporting the medial longitudinal arch of the foot (Hamilton et al., 2008, Semple et al., 2009).

The TP is the primary invertor of the foot (Rha et al., 2010). During early stance TP acts eccentrically to generate an inversion moment that opposes the net eversion moment, limiting eversion at the STJ, before acting concentrically to contribute to supination later in stance (Maharaj et al., 2016). Of the extrinsic foot muscles the TP has by far the longest mean inversion moment arm at the STJ of 19.2 mm, as estimated from cadaveric data (**Table 2-1**.)

(Klein et al., 1996). An inversion moment arm of TP of approximately 2 cm has also been reported elsewhere (Piazza et al., 2001, Piazza et al., 2003). Conversely TP had a larger internal rotation moment arm (21.4 ± 2.7 mm) than inversion moment arm (10.2 ± 2.4 mm) in a later study (McCullough et al., 2011). In this case rotation of the cadaver was of the whole foot and not isolated to specific joints. The relatively small sagittal plane moment arm at the ankle (8 mm), demonstrates that TP places a much lesser role in plantar flexion compared to the triceps surae (mean moment arm of 52.8 ± 5.1 mm) (Klein et al., 1996). Activity of TP is worthy of investigation in FO research as we might expect that an anti-pronation FO that decreases the external eversion moment and reduces eversion would mean reduced EMG activity from TP.

Broadly speaking TP has two bursts of activity, the first in early stance around initial contact and the second in midstance (**Figure 2-6**). The onset of the first burst was prior to heel contact around 90% of the gait cycle and peaked around 5% of the gait cycle in a study of 15 participants with a neutral foot posture walking barefoot (Murley et al., 2009a). During early stance TP is acting eccentrically to generate an inversion moment that opposes the net eversion moment, limiting pronation at the STJ. The second peak around 35% of the gait cycle corresponds to active supination of the STJ (Murley et al., 2009a, Maharaj et al., 2016). However the activation profile of TP is extremely variable, with some individuals not displaying a clear double burst pattern (Murley et al., 2009a). In some cases TP activation appears triphasic, with a small magnitude of activation in early swing (Murley et al., 2014b).

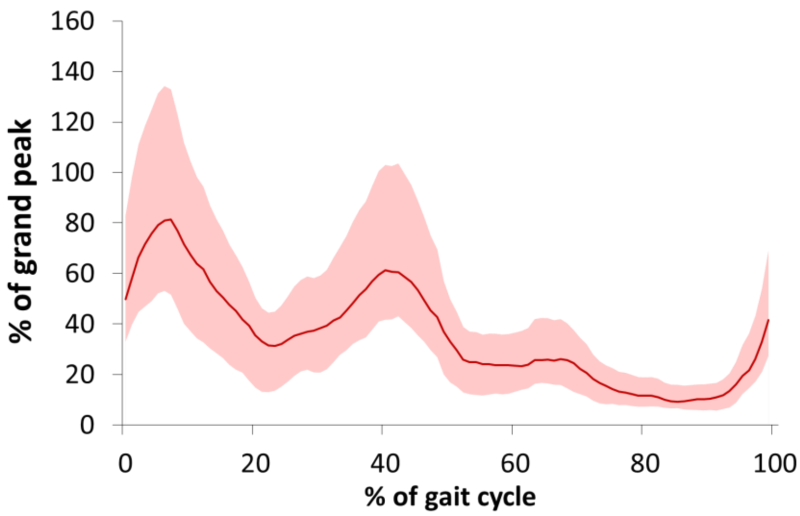


Figure 2-6 Example tibialis posterior route mean square (RMS) EMG, with standard deviation, normalised to the peak

The activation of TP during the gait cycle and its relatively short length of muscle fibres with respect to its total MTU length reflects its capacity for energy storage and release during walking (Maharaj et al., 2016). However the functional anatomy of TP may also explain the relative frequency of TP tendinopathy compared to other muscles of the foot and ankle (Abate et al., 2009, Ward et al., 2009). Indeed TP tendinopathy is thought of as one of the most common tendinopathies (Maganaris et al., 2004).

A cadaveric study found that TP had the shortest muscle fibres in the ankle complex, 3.78 ± 0.49 cm, versus 5.10 ± 0.98 cm in the medial gastrocnemius for example (Ward et al., 2009). The Lf/Lm ratio of the TP was 0.12 ± 0.02, one of the shortest of all 27 muscles of the lower limb studied. Short muscle fibres have less sarcomeres in series than long fibres, so for a given MTU velocity, absolute sarcomere velocity is greater with shorter muscle fibres than longer fibres, because each sarcomere has to change length more to achieve the required change in total fibre length in the given time (Lieber and Friden, 2000). The higher the muscle/sarcomere velocity the lesser the force generating capacity based on the force-velocity relationship (Lieber and Friden, 2000). In order to reduce muscle shortening and thus improve force generating capacity of the MTU, tendon accommodates a lot of the length change during muscle contraction.

By accommodating the majority of the length change during rapid eccentric contractions that involve high forces, the series elastic component (SEC) is thought to act like a mechanical buffer, protecting the muscle fascicles from damage (Maharaj et al., 2016, Reeves and Narici, 2003). A dynamic ultrasound investigation found that as the TP was acting eccentrically in early stance, the muscle fascicles were shortening while the MTU was lengthening (Maharaj et al., 2016). The SEC absorbs energy during early stance, while the TP is opposing the net eversion moment. Energy is then released in midstance for active supination. The recoil of the SEC during active supination of the STJ allows the TP muscle fascicles to shorten slower than the MTU itself. While the power absorption and generation of the SEC improves mechanical efficiency and reduces the risk of injury to the muscle fascicles (Maharaj et al., 2016), the resulting high strain through the SEC may lead to overuse injury to the tendon itself, i.e. TP tendinopathy (Maharaj et al., 2015). During power absorption and generation in the tendon some energy is lost in the form of heat, which can damage tendon over repeated loading and unloading (Maganaris et al., 2004).

The role of TP in maintaining foot function and foot posture is evident as an acquired flat foot deformity is seen with TP tendon dysfunction (Durrant et al., 2011, Squires and Jeng, 2006). A flat foot may mean greater plantar fascia loading than with a normal arch, possibly contributing to the development of plantar fasciitis (Angin et al., 2014).

2.4.5.1 *Medial shin pain: tibialis posterior tendon dysfunction*

If FO reduced the external eversion moment during stance, with medial rearfoot posting for example, we might expect the force required from the tibialis posterior to resist eversion to be less and so a reduced EMG activation would be seen. Reduced activation could mean less force through the tendon which could facilitate healing in the case of tibialis posterior tendon dysfunction. The TP tendon stretches and absorbs energy in early stance, which is returned later in stance during active supination (Maharaj et al., 2016). A reduced external eversion moment in early stance with FO could also mean less force borne directly by the TP tendon which might not necessarily be reflected in reduced EMG activity. Increased external arch support with FO could reduce the force required from the TP muscle to provide active supination in mid-late stance, which may also reduce the internal force through the TP tendon. However there is little evidence to date to support this theory of the mechanism of FO (Murley et al., 2010a).

2.4.6 *The Peroneals*

Peroneus longus (PL) and peroneus brevis (PB), collectively referred to as the peroneals or fibularis muscles, evert (pronate), abduct and plantar flex the tarsal joints and assist in plantar flexing the ankle (Hamilton et al., 2008, Campanini et al., 2007). The peroneals are important in overall foot and ankle function and maintaining joint stiffness and stability (Murley et al., 2010a, O'Connor and Hamill, 2004). Co-activation of PL and its antagonist TP may serve to stabilise the foot in the frontal plane (Murley et al., 2009a).

The peroneals are located on the lateral aspect of the shank (**Figure 2-5**). The origin of PL is on the head and lateral shaft of the fibula, with its distal tendon originating in the middle third of the lateral compartment of the shank (Johnson and Christensen, 1999). The tendon passes behind the lateral malleolus and peroneal tubercle and through the peroneal groove of the cuboid, slants across the sole of the foot and inserts on the plantar lateral aspects of the base of the first metatarsal and the medial cuneiform (Hamilton et al., 2008, Johnson and Christensen, 1999, Shyamsundar et al., 2012). In one cadaver study all 26 cadaveric specimens had a strong band of the PL tendon that inserted on to the base of the first metatarsal and 22 had a branch to the medial cuneiform (Shyamsundar et al., 2012). However substantial differences in the number and location of additional insertion sites existed. Eight specimens had two insertion sites, three specimens had three sites, 13 had four sites and a further three specimens had five or more attachments. The additional sites were the calcaneum or other metatarsals, most commonly the 4/5 metatarsal in nearly half of the sample (12/26). Further variation was reported in the presence of a sesamoid bone distal to the cuboid channel in 16 cases. Differences in morphology between feet from the same individual occurred in eight of the 12 pairs of feet, including whether a sesamoid bone was present. It is disputed whether the origin of a sesamoid in the PL tendon is due to genetic variation or the result of ossification of the tendon due to mechanical stress (Bianchi et al., 2017). The PB is below PL in the distal half of the lateral side of the shank. The tendon of PB also travels behind the lateral malleolus, anterior to the tendon of PL and continues above the PL tendon to attach on the tuberosity of the fifth metatarsal below the attachment of peroneus tertius (Hamilton et al., 2008, Otis et al., 2004). Peroneus tertius is a small muscle that also pronates the tarsal joint, but assists in dorsiflexion, residing lateral to the extensor digitorum longus (Hamilton et al., 2008).

From cadaveric data PL and PB have nearly identical moment arms at the STJ (21.8 and 20.5 mm respectively) (Klein et al. 1996). However the PCSA of PL is approximately twice that of the PB, 10.4 ± 3.8 cm^2 versus 4.9 ± 2.0 cm^2 respectively (Ward et al., 2009). In another *in vitro* study mean eversion moment arm was 31 ± 2.3 mm and 20.5 ± 6.4 mm of the whole foot for PL and PB respectively (McCullough et al., 2011). The difference in PL eversion moment arm lengths between studies is likely due to motion being restricted to individual joints using bone pins in the study by Klein and colleagues, whereas rotation was of the whole foot in the study by McCullough and colleagues (2011). It has been argued that PB is a more effective evertor than PL because on average 2.1° more external rotation of the navicular joint occurred at the talonavicular joint and 0.9° more calcaneal valgus relative to the talus (eversion) occurred at the STJ in the PB than the PL when loaded to the same magnitude in six cadavers (Otis et al., 2004). However owing to the smaller PCSA and so lesser force generating capacity of PB, it is questionable whether equal loading of both peroneal MTU occurs during gait. It could thus be argued that the larger PCSA of PL than PB makes PL more relevant to study with regards eversion moment generation capacity. Furthermore the functional relevance of semi static cadaver foot models such as that used in the study of Otis and colleagues has been questioned (Nester et al., 2007b).

Peronus longus and brevis are active during mid-late stance (Bovi et al., 2011, Murley et al., 2009a, Campanini et al., 2007, Hamilton et al., 2008) (**Figure 2-7**). In mid-late stance the tibia is externally rotating and the foot is supinating (Williams and Nester, 2010b), so presumably the eversion moment generated by the peroneals serves to control supination and maintain stability in the frontal plane. However PL activity in the literature is very variable with an average standard deviation of 49% over the gait cycle (Onmanee, 2016). Large variability in PL EMG may in part be due to measurement issues. Murley and colleagues (2009a) described the EMG profile of PL in walking as having an earlier burst of activity in the contact phase (20% maximum voluntary isometric contraction, MVIC) as well as the burst in midstance (19% MVIC). However this pattern was only apparent when looking at the ensemble average across all fifteen participants. When examining the EMG profiles of each individual there was high between subject variability in their data and only a few participants (3/11 presented) had a profile close to the double bump pattern the authors described. The same double bump pattern was reported in later work by Murley and colleagues (2014b), however only the ensemble averages were plotted for various walking speeds, with no indication of the individual variability in activation. As such it is not possible to conclude that this dual peak pattern of PL activation commonly occurs at an individual level. Poor reliability was reported for PL EMG by Murley and colleagues (2010b) using fine-wire electrodes. It might be that the repeatability of the EMG signal is sensitive to the position of motor units when using fine-wire electrodes in PL (Kadaba et al., 1989). Cross-talk from TA in the contact phase could be erroneously be interpreted as an initial burst of activity of PL when using surface electrodes as the muscle belly of PL is close to TA (Campanini et al., 2007). Improving the reliability of recording PL EMG is warranted. The EMG of PB has been studied less than PL in the literature, but was typically active in midstance and peaked before heel-off (31% MVIC) in walking (Murley et al., 2009a).

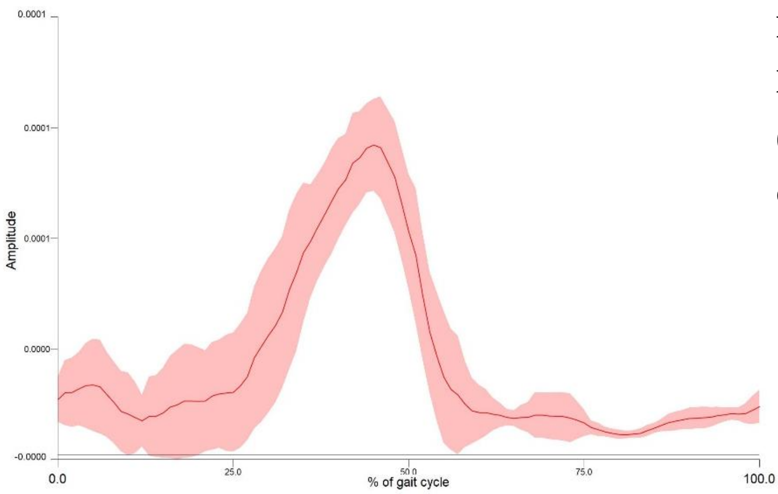


Figure 2-7. Example peroneus longus route mean square (RMS) EMG with standard deviation

Conflicting results in the literature exist regarding an association between PL EMG activity and morphology and foot type. In a group of 30 flat footed individuals PL EMG activity was less across stance compared to 30 individuals defined as having normal arched feet (Murley et al., 2009c). Similarly 15 males with a pronated static foot posture had less PL EMG activity in early stance compared to 18 controls with a normal foot posture (Hunt and Smith, 2004). However because no mention was made of the electrode size or inter-electrode distance, we cannot be certain that what was recorded as PL activity in early stance in the pronated group was not in fact crosstalk from TA. Muscle thickness of PL was greater in the pes planus group than normal foot posture group in one study (Murley et al., 2014a), yet in another study, for PL and PB combined the cross sectional area and muscle thickness were 14.7% and 10% less respectively in the pes planus group compared to the normal foot posture group (Angin et al., 2014). One might expect less activation and an accompanied lesser peroneal muscle volume over time with a pronated foot posture if less of an eversion moment was required from the peroneals for a foot that was already more everted compared to a neutral/normal foot. Nevertheless the associations of peroneal activity and morphology with foot posture reported are only based on cross-sectional studies and no directional or causal relationship between foot type and peroneal activity and morphology can be established. Additionally static classifications of foot type have limited resemblance to foot kinematics and as such may have little relevance to muscle and foot function (Buldt et al., 2015).

2.4.7 *Gastrocnemius*

The medial and lateral heads of the gastrocnemius are located in the superficial compartment of the calf and form the triceps surae in combination with the soleus and plantaris, where present (Dalmau-Pastor et al., 2014). The gastrocnemius is bi-articular and its primary functions are to plantar flex the ankle joint and flex the knee. The knee flexion moment that the gastrocnemius can generate has been shown to be greatest when the knee is fully extended, across a range of ankle angles (Li et al., 2002). The origin of the medial head of the gastrocnemius is an area on the posterior aspect of the distal, medial, epiphysis of the femur (Dalmau-Pastor et al., 2014). The lateral head originates posterior to the lateral femoral epicondyle and proximal to the insertion of the tendon of popliteus in the lateral supracondylar ridge (Dalmau-Pastor et al., 2014). The triceps surae insert onto the calcaneus via the Achilles tendon, which forms approximately halfway along the length of the calf (Dalmau-Pastor et al., 2014, Szaro et al., 2009, Tashjian et al., 2003). The medial head of the gastrocnemius has a

PCSA around twice that of the lateral head (21.1 ± 5.7 cm^2 versus 9.7 ± 3.3 cm^2 respectively from (Ward et al., 2009). The plantar flexion moment arm of the triceps surae (Achilles tendon as a whole) at the ankle was 52.8 ± 5.1 mm in one cadaver study (Klein et al., 1996). The length of the moment arm in the sagittal plane for the Achilles tendon was almost twice that of the next largest plantar flexor moment arm of the flexor hallucis longus (26.6 ± 4.4 mm).

The gastrocnemius may have a secondary role in contributing to movement in the frontal plane. Stimulation of the medial gastrocnemius has been shown to generate torque in the frontal plane, with peak ankle inversion torque equating to approximately 13% of peak plantar flexion torque (Vieira et al., 2013). The mean eversion moment of the triceps surae as a whole in one cadaver study was 5.3 ± 7.4 mm at the STJ in the study by Klein and colleagues (1996). Variation in moment arm was such that in eversion it had an inversion moment arm and in inversion it had an eversion moment. The high standard deviation within the sample suggested that the behaviour of the triceps surae in the frontal plane could be highly individualised (Klein et al., 1996). Large individual variation could in part be due to different patterns of twisting of the tendon fascicles from individual heads of the triceps surae (Edama et al., 2015, Edama et al., 2016). When the whole foot was rotated *in vitro*, the moment arm in the coronal plane was as little as 1.0 ± 4.7 mm for the Achilles tendon as a whole (McCullough et al., 2011). When the three heads of the triceps surae were considered individually, it was found that force applied to the lateral gastrocnemius caused an eversion moment, whereas force applied to the other heads of the triceps surae in insolation and the three together produced an inversion moment (Arndt et al., 1999). The moment arms at the ankle in the frontal plane of the medial and lateral gastrocnemius have been estimated *in vivo* using ultrasound and a rotating platform (Lee and Piazza, 2008). In 15° of eversion at the ankle the lateral head of the gastrocnemius had an eversion moment arm, while the moment arm of the medial head was variable around zero. With the ankle neutral, both heads of the gastrocnemius had an inversion moment arm around 4 mm and at 20° of inversion of the ankle the inversion moment arms increased to around 8 mm. Moment arm lengths were similar at rest to during light contraction and during a maximal voluntary contraction. The ankle inverts at 40-60% of the gait cycle (Nester et al., 2014), which coincides with the peak of medial gastrocnemius activity, around 40-50% of the cycle (Bovi et al., 2011, Hof et al., 2002, Van Hedel et al., 2006, Winter and Yack, 1987). Therefore there is potential for the medial gastrocnemius to play a minor role in inversion during walking.

Electromyography activity of the medial gastrocnemius gradually increases from early contact to peak in midstance, around 40-50% of the gait cycle (Bovi et al., 2011, Hof et al., 2002, Hunt et al., 2001, Onmanee, 2016, Van Hedel et al., 2006, Winter and Yack, 1987).

2.4.8 *Soleus*

The soleus originates on the posterior aspects of both the tibia and fibula where tendinous fibres are attached to the head and upper quarter of the fibula and tibia (Dalmau-Pastor et al., 2014). Based on the high force generating capacity of soleus, it is considered by some as the most important muscle at the ankle (Ward 2009). The separate compartments of the Achilles tendon could allow the three triceps surae muscles to function independently (Bojsen-Moller and Magnusson, 2015). As the soleus consistently inserts on the medial portion of the calcaneus, furthest from the inversion-eversion joint axis, it could potentially contribute more to inversion than the gastrocnemii (Bojsen-Moller and Magnusson, 2015).

Electromyography activity of soleus in walking is similar to the medial gastrocnemius, peaking in midstance, around 40-50% of the gait cycle (Bovi et al., 2011, Hunt et al., 2001, Onmanee, 2016, Winter and Yack, 1987). However the soleus is particularly well suited to postural stabilisation due to its relatively high PCSA and short fibre length, which favours the generation of high forces with small excursions (Lieber and Friden, 2000). It is possible that soleus has more muscle spindles than the gastrocnemius, making the soleus more suited to making small postural adjustments in standing than gastrocnemius (Di Giulio et al., 2009). Additionally the soleus has a high proportion of type I (slow twitch) muscle fibres, optimised for endurance activity, while the gastrocnemius has a greater distribution of type II (fast twitch) fibres, better designed for explosive activity (Di Giulio et al., 2009).

2.4.9 *Plantar intrinsic foot muscles*

The intrinsic foot muscles are found in two layers on the dorsum and four layers on the plantar surface of the foot (**Figure 2-8**) (Kura et al., 1997). Activation of the plantar intrinsic foot muscles is often simultaneous, so they are sometimes considered a functional unit (Kelly et al., 2015). From fine-wire recordings activation of plantar intrinsic muscles is thought to be primarily in mid-late stance (Reeser et al., 1983). The abductor hallucis (ABH), flexor digitorum brevis (FDB) and quadratus plantae (QP) are three plantar intrinsic muscles that have received recent attention in the literature due to their potential roles in postural stability and MLA support in locomotion (Kelly et al., 2012, Kelly et al., 2014, Kelly et al., 2015, Kelly et al., 2016, Farris et al., 2019, Kelly et al., 2018). Recruitment of plantar intrinsic muscles has been shown to increase with greater postural demand (Kelly et al., 2012). However using a nerve block to prevent activation of the plantar intrinsic muscles had little effect on the deformation of the MLA during walking or running, suggesting the plantar intrinsics provide little active contribution to arch support under high loads in locomotion (Farris et al., 2019). The plantar intrinsics are perhaps more important in more demanding dynamic tasks than in steady state walking as activity of FDB and ABH increased in stepping on and off a platform compared to level stepping (Riddick et al., 2019).

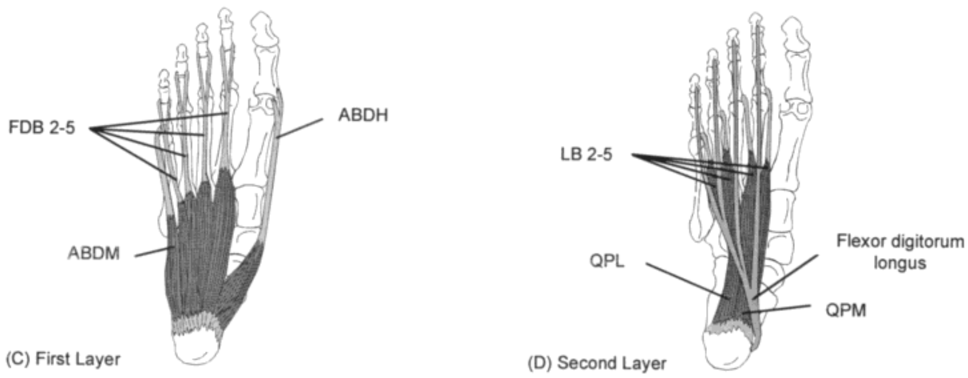


Figure 2-8. First and second layers of plantar intrinsic muscles, from (Kura et al., 1997). FDB 2-5: four flexor digitorum brevis; ABDH: abductor hallucis, ABDM: abductor digiti minimi; LB 2-5: four lumbricals; QPM: quadratus plantar medialis and QPL: quadratus plantar lateralis

2.4.9.1 *Abductor hallucis*

The abductor hallucis (ABH) is an intrinsic foot muscle that originates on the medial calcaneus and inserts on the medial base of the proximal phalanx of the first toe (Hamill and Knutzen, 2009). The function of ABH is to abduct and flex the metatarsophalangeal joint of the big toe (hallux) (Tosovic et al., 2012), however abduction of the hallux is not a common, functional movement (Wong, 2007). It is thought that ABH contributes to actively supporting the MLA (Gray and Basmajian, 1968, Kamiya et al., 2012, Kelly et al., 2014, Reeser et al., 1983, Wong, 2007), but active support of the arch may only occur at high loads (Basmajian and Stecko, 1963, Kelly et al., 2014). The ABH has the largest PCSA of the plantar intrinsic foot muscles (Kura et al., 1997, Tosovic et al., 2012). As the most medial of the plantar intrinsic muscles, the ABH potentially has the greatest moment arm with respect to the axis of rotation in the frontal plane (e.g. STJ axis), therefore the greatest potential to generate an inversion moment, which would support the MLA, as the arch collapses in eversion (Kelly et al., 2014, Tosovic et al., 2012). The position of the ABH means it could be possible to record surface EMG from this muscle if the sensor was small enough, however the feasibility and reliability of this has yet to be established in walking. However activity of the ABH was found to decrease in step descent with rearfoot and rearfoot and forefoot medial wedging when measured with a small surface sensor (Bonifacio et al., 2018).

The relationship between foot posture and ABH size and function is unclear. The CSA and thickness of ABH was significantly smaller (12.8% and 6.8% respectively) in a pes planus group (n=49) compared to a group with normal foot posture group (n=49) (Angin et al., 2014). However earlier work found that activation of ABH in walking measured using fine-wire electrodes was greater in the flat foot group than the normal foot posture group (Gray and Basmajian, 1968). One would expect that if the intrinsic foot muscles of flat footed individuals are consistently more active than those with normal foot posture, that this would be reflected in a greater size of the intrinsic foot muscles in flat feet than normal feet. However there were only 10 participants in each group in the earlier study, which may not give a true picture of ABH EMG activity, as EMG data is typically very variable. Equally, no directional or causal relationship between foot type and either ABH EMG activity nor morphology can be determined from cross-sectional studies.

2.4.9.2 *Flexor digitorum brevis*

The flexor digitorum brevis had a similar activation pattern to ABH in walking in the study by Gray and Basmajian (1968) and mean activation was also greater in the flat footed group than the normal foot posture group. The onset of activity of both FDB and ABH is broadly mid-late stance (Reeser et al., 1983). No difference in FDB CSA or thickness was found between pes planus feet and normal feet by Angin et al. (2014).

2.4.9.3 *Quadratus plantae*

Quadratus plantae (a.k.a. flexor accessorius) has two heads, arising from the medial and lateral aspect of the calcaneus which insert into the tendon of FDL (Ledoux et al., 2001, Sooriakumaran and Sivananthan, 2005). The role of QP is disputed, it may have a small contribution to eversion or may serve to assist in optimising the line of pull of FDL (Sooriakumaran and Sivananthan, 2005).

2.4.10 *Connective tissue*

2.4.10.1 *Achilles tendon*

Tendon fascicles from the two heads of the gastrocnemius are twisted together with fascicles from the soleus to form the Achilles tendon, also known as the calcaneal tendon, which inserts on to the calcaneus (Dalmau-Pastor et al., 2014, Edama et al., 2015). The Achilles tendon is the longest and strongest tendon in the body at 10-15 cm long in adults (Szaro et al., 2009, Tashjian et al., 2003). As a long, compliant tendon, the Achilles tendon is well suited to storage of elastic-strain energy during eccentric contraction which is released during concentric contraction, optimising power output and efficiency and reducing the energy cost of walking (Fukunaga et al., 2001, Lichtwark et al., 2007, Lichtwark and Wilson, 2006, Magnusson et al., 2008b). The Achilles tendon accommodates the majority of the length change of the muscle-tendon unit during eccentric action which enables muscle fascicles to remain relatively isometric and protect them from damage (Fukunaga et al., 2001, Lichtwark et al., 2007, Lichtwark and Wilson, 2006). Repetitive length changes and high strain through the Achilles tendon however predisposes it to injury and Achilles tendinopathy and rupture are common (Albers et al., 2016, Kujala et al., 2005, Lopes et al., 2012, Szaro et al., 2009). Achilles tendinopathy is one of the most common conditions for which FO are prescribed.

Despite individual variation in the pattern of twist and predominant insertion site of the tendon fibre bundles (fascicles) from each of the triceps surae, some degree of twisting occurs in all Achilles tendons (Edama et al., 2015, Edama et al., 2016, Lersch et al., 2012). From proximal to distal fascicles typically twist counter clockwise in the right leg and clockwise in the left leg (Edama et al., 2015, Lersch et al., 2012). In general fascicles from the medial gastrocnemius insert on the lateral-posterior (superficial) side of the calcaneus, fascicles from the lateral gastrocnemius are in the lateral-anterior (deep) part of the tendon and fascicles from the soleus insert on the anterior-medial aspect of the calcaneus (Edama et al., 2015, Edama et al., 2016, Szaro et al., 2009). Dissected tendons have been have been categorised into three types: least, moderate and extreme, depending on their degree of torsion (Edama et al., 2015, Edama et al., 2016).

Material properties throughout the Achilles tendon are not uniform. From its wide origin the tendon narrows in its mid-section, before widening again where it inserts on the calcaneus (Reeves and Cooper, 2017). The CSA of the Achilles tendon was found to be 51% and 85% greater in the distal portion of the tendon (1 cm from the calcaneus) than 7 cm from the calcaneus (Magnusson and Kjaer, 2003). As the total length of the Achilles tendon is around 10-15 cm (Tashjian et al., 2003), the most proximal site in the study by Magnusson et al. can be considered the mid portion of the tendon. Regions of lower CSA would be expected to experience higher stress for a given force than regions of higher CSA (Reeves and Cooper, 2017) and thus be at greater risk of injury. Non-insertional tendinopathy occurs 2-6 cm from the calcaneus, which coincides with this narrow mid region of the tendon (Irwin, 2010).

Tendinopathy is particularly common on the medial side of the mid region of the Achilles tendon (Gibbon et al., 2000). Additionally the most common site of complete rupture of the Achilles tendon is in the mid-portion (Jozsa et al., 1989).

The twisted, compartmentalised structure of the Achilles tendon theoretically enables heterogenous behaviour of the tendon fascicles, although the functional and clinical implications of this feature are complex. It has been proposed that sliding between fascicles facilitates a large capacity for extension in energy storing tendons (Thorpe et al., 2016, Thorpe et al., 2012). Independent stretching of tendon fascicles could allow muscle fibres of the triceps surae to operate at different fibre lengths in order to optimise force generation capacity (Franz et al., 2015). It has also been suggested that the twisting of fibres helps equalize moment arm and strain within tendon in broad muscles that might otherwise occur and again allowing the muscle to operate within the most effective portion of the force-length curve (Dean et al., 2007).

It has been shown with an animal model that the strain in a tendon determines its rupture. As previously outlined, strain is defined as the displacement [of the tendon] divided by the resting length, expressed as a percentage (Almeida-Silveira et al., 2000, Lersch et al., 2012). Achilles tendon stiffness decreased in rats following three weeks of suspension of their hindlimbs compared to control rats (Almeida-Silveira et al., 2000). The tendons were removed from the rats and subjected to mechanical testing and it was found that the tendon ruptured at 20% strain in both groups, but maximal stress was 35% less in the suspended limb group. So less force was required to rupture the tendon in the suspended limb group.

Differences in tendon strain between portions of the Achilles tendon *in vitro* have been shown to be more dependent on calcaneal angle than differences in relative loading between the medial and lateral gastrocnemius (Lersch et al., 2012). Inversion increased strain in the lateral and decreased strain in the medial portions of the distal tendon Achilles tendon respectively, whereas eversion increased strain in the medial portion and decreased strain in the lateral portion. Altering calcaneal angle lead to intra-tendinous strain differences of 15%, whereas altered force distribution resulted in differences of only 2.5%. It would appear that large eversion excursion places highest strain in the medial portion of the Achilles tendon, which is typically composed of fibres from the soleus, however the authors did not vary the loading through the soleus muscle. Non-uniform displacement (and potentially therefore strain) of the Achilles tendon *in vivo* has been observed during passive ankle rotation, eccentric loading and walking (Arndt et al., 2012, Franz et al., 2015, Franz and Thelen, 2015, Slane and Thelen, 2014). During passive movement of the ankle joint displacement of the deep layer was greater than the superficial layer during dorsiflexion (10.4±2.1 mm vs. 8.4±1.9 mm respectively, $p < 0.01$) (Arndt et al., 2012). However displacement was not expressed as a percentage of resting length (strain). Similarly greater displacement was observed in the deeper layer compared to the superficial layers in both passive dorsiflexion and eccentric loading (Slane and Thelen, 2014). Nonetheless the heterogeneity of the structure of the Achilles tendon should be kept in mind when considering how it might respond to a change in loading with an intervention like FO.

2.4.10.1.1 Achilles tendinopathy

Relieving tension through the Achilles tendon with FO would be beneficial in treating Achilles tendinopathy and this could be achieved simply using heel lifts to increase plantar flexion (Rosenbloom, 2010). Foot orthoses with 12 mm heel wedges were shown to increase ankle dorsiflexion moments and reduce plantar flexion moments, however estimated Achilles tendon loading did not reduce, rather there was a shift in distribution from the medial to lateral side (Weinert-Aplin et al., 2016). Conversely as the medial portion of the Achilles tendon, which is the most common region of Achilles tendinopathy (Gibbon et al., 2000), is under most strain in eversion, FO could exert a therapeutic effect through the traditional theory of reducing eversion (Lersch et al., 2012, Rosenbloom, 2010). However there was a mean ± SD improvement in symptoms of 92% ± 16% in runners with Achilles tendinopathy after wearing customised FO designed to reduce eversion, despite an unexpected increase in eversion (Donoghue et al., 2008). A reduction in activation of the triceps surae would theoretically reduce load through the Achilles tendon, however the literature suggests that medial FO do not reduce triceps surae muscle activity in either walking or running (Mills et al., 2012b, Murley and Bird, 2006, Telfer et al., 2013a, Murley et al., 2010a). As such, the mechanism behind the therapeutic benefit of medial FO in treating Achilles tendinopathy is unclear. Recent reviews on the efficacy of FO in treating Achilles tendinopathy have found only weak evidence that FO are equivalent to physical therapy or no treatment(Scott et al., 2015) and recommended against the use of FO to improve pain or function (Wilson et al., 2018).

2.4.10.2 *Plantar fascia*

The plantar fascia has attachments at the calcaneus, talus, navicular, cuneiforms, and 1st-3rd metatarsals and has a medial and lateral portion and a central portion (plantar aponeurosis) (Kalicharan et al., 2017). It is thought that passive structures are more important than muscles in stabilising MLA and that the plantar fascia plays the greatest role (Dawe and Davis, 2011). Sectioning of the plantar fascia in cadaveric feet can decrease arch stiffness by 25% (Huang et al., 1993). According to a "truss model" of the arch, as load is applied to the apex of the arch the stiffness of the plantar fascia prevents collapse of the arch because it restricts the ability of the ends of the arch (calcaneus and metatarsals) to move apart (Dawe and Davis, 2011). As discussed above the plantar fascia influences MLA height and its contribution to foot function can be conceptualised via the windlass mechanism, arch-spring mechanism or some combination of the two (Hicks, 1954, Ker et al., 1987, Welte et al., 2018). The plantar fascia is made up of collagen and fibroblasts (Kalicharan et al., 2017), which is similar to tendon and so it may adapt to increased loading in a similar way to tendon (increased stiffness and maybe increased CSA)

2.4.10.2.1 Plantar fasciitis

One mechanism by which FO may be clinically beneficial in treating plantar fasciitis is through increased arch height that re-distributes plantar load and reduces pressure at the site of pain in the heel. Increases in arch height can increase the peak plantar pressure in the midfoot while reducing the decreasing the peak pressure in the rearfoot (Farzadi et al., 2015, Hodgson et al., 2006, McCormick et al., 2013, Sweeney, 2016). Alternatively FO could directly reduce the strain though the plantar fascia (Kogler et al., 1996). Increasing arch height and external arch support with FO could alleviate symptoms by reducing the internal arch support required, thereby reducing loading of the plantar fascia.

Foot orthoses are often supplied to those with flat feet (pes planus) and it is thought that extreme pes planus (and also pes cavus) may be lead to plantar fasciitis (Neufeld and Cerrato, 2008). However, the evidence for a relationship between foot posture and plantar fasciitis is largely anecdotal (Wearing et al., 2006). A case-control study found a strong association ($p < 0.01$) between a pronated foot posture and chronic heel pain, however those with heel pain also had a significantly greater BMI ($p < 0.01$) (Irving et al., 2007). A cavus foot was a risk factor for developing plantar fasciitis in a prospective study of runners ($p < 0.0005$) (Di Caprio et al., 2010), however a qualitative, subjective assessment of foot posture was made by a single examiner. Consequently whether the benefit of FO for treating plantar fasciitis is derived from changes in foot posture is questionable. . The neuromuscular theory of the therapeutic effect of FO was not supported in a study which found no change in EMG activity in patients with plantar fasciitis after wearing FO for nine weeks, despite reductions in pain (Moyne-Bressand et al., 2018). However the muscles tested were TA, SOL, biceps femoris and vastus medialis, which are not necessarily the most relevant muscles to plantar fasciitis.

Recent systematic reviews on the effectiveness of FO in treating plantar heel pain produced conflicting outcomes, in that one concluded FO were no more effective than sham orthoses (Rasenberg et al., 2018), while the other that there was moderate-quality evidence that FO were effective at reducing pain in the medium term (Whittaker et al., 2018b). The contrast in conclusions was attributed, in an editorial, to the different pain scales used in the studies and the interpretation of the confidence intervals (Whittaker et al., 2018a). Overall clinical consensus supports the use of biomechanical support, including FO, as a safe and effective treatment for plantar fasciitis (Schneider et al., 2018).

2.5 Approach

The focus of this thesis is understanding the effect of FO on soft tissues and below is a rationale for the contributions of muscles and connective that were considered a priority . The reliability of measuring muscle activity in some of these muscles has been identified as a concern, so the reliability of measures was addressed prior to studying the FO effect. Given the knowledge of the conditions for which FO are used, the current evidence on the effect of FO and the uncertainties regarding the underlying mechanisms, four research questions were developed.

2.5.1 Research questions

1) What is the level of evidence that footwear, FO and taping alter lower limb muscle activity during walking and running?

2) Are the EMG signals measured from PL and ABH EMG and the medial longitudinal arch angle based on skin markers reliable in walking?

3) What are the immediate effects of variations in FO geometry on EMG of select muscles of the lower limb and foot and ankle biomechanics?

4) What are the longitudinal effects of FO on soft tissue morphology, plantar pressure and skin sensitivity?

The muscles chosen for investigation of the immediate effects of FO were: tibialis posterior, tibialis anterior, medial gastrocnemius, peroneus longus and abductor hallucis. Due to having the largest inversion moment arm, TP is the largest invertor at the STJ (Klein et al., 1996) and most likely to respond to FO as in theory a reduction in external ankle moment with FO would reduce the requirement of TP to generate a counter internal inversion moment. Although TA has only a small role in inversion, it is important in overall foot function, and can be measured easily with surface EMG. The FHL has a greater inversion moment arm than TA, but it is a deep muscle and it was not considered worth further invasive measurement. Again recording

EMG from FDL at this stage seemed unnecessarily invasive. Although FDL activity may be relevant in MTSS, which may be treated with FO, the etiology of MTSS in disputed. For the plantar flexors the medial gastrocnemius was chosen over lateral gastrocnemius and soleus because the medial gastrocnemius is larger than the lateral gastrocnemius and the soleus has more of a role in standing than walking and the study examined the effects of FO in walking not standing. The peroneals operate in the frontal plane so could be affected by medial FO. In this case PL was chosen over PB because PL has the greater CSA. Finally ABH EMG was recorded as it is relatively accessible with the Delsys Trigno™ Mini sensor unlike FDB and QP. Besides the ABH has the greatest PCSA and moment arm of the plantar intrinsics to support the MLA and has a similar activation pattern to FDB (Gray and Basmajian, 1968).

The study on the long term effect of FO was more exploratory and included muscles and other soft tissue that were considered potentially relevant in overall foot function, there is some evidence for a difference in their morphology with foot posture or may plausibly differ as a result of specific pathologies. See Chapter 7 for further rationale.

Chapter 3 A SYSTEMATIC REVIEW OF THE EFFECT OF FOOTWEAR, FOOT ORTHOSES AND TAPING ON LOWER LIMB MUSCLE ACTIVITY DURING WALKING AND RUNNING

This chapter is an adaptation of a literature review on the effect of footwear, foot orthoses and taping on lower limb muscle activity during walking and running accepted for publication with Prosthetics and Orthotics International (authors: Joanna Reeves, Richard Jones, Anmin Liu, Leah Bent, Emma Plater, Christopher Nester). The three external devices were reviewed together as they can all be used to treat similar musculoskeletal conditions of the lower extremity and it gives us an indication of what change in muscle activity is possible with a change in the distribution of load at the foot and ankle. Additional detail to the published review has been added in relation to the studies that measured TP activity as a lack of high quality studies investigating TP EMG was a rationale for the study in Chapter 6.

3.1 Abstract

3.2 Background:

External devices are used to manage musculoskeletal pathologies by altering loading of the foot, which could result in altered muscle activity that could have therapeutic benefits.

3.3 Objectives:

To establish if evidence exists that footwear, foot orthoses and taping alter lower limb muscle activity during walking and running.

3.4 Study design:

Systematic literature review.

3.5 Methods:

CINAHL, MEDLINE, ScienceDirect, SPORTDiscus and Web of Science databases were searched. Quality assessment was performed using guidelines for assessing healthcare interventions and electromyography methodology.

3.6 Results:

Thirty-one studies were included: 22 related to footwear, eight foot orthoses and one taping. In walking: 1) Rocker footwear apparently decreases tibialis anterior activity and increases triceps surae activity; 2) Orthoses could decrease activity of tibialis posterior and increase activity of peroneus longus; 3) Other footwear and taping effects are unclear.

3.7 Conclusion:

Modifications in shoe or orthosis design in the sagittal or frontal plane can alter activation in walking of muscles acting primarily in these planes. Adequately powered research with kinematic and kinetic data is needed to explain the presence/absence of changes in muscle activation with external devices.

3.8 Clinical relevance:

This review provides some evidence that foot orthoses can reduce tibialis posterior activity, potentially benefiting specific musculoskeletal pathologies.

3.9 Keywords:

Rocker footwear; foot orthoses; tibialis posterior; electromyography

3.10 Word count:

4,981

Background

Musculoskeletal pathologies occur when structures experience more load than they can withstand.[1] If external loads are altered by a therapeutic device there should be a corresponding change in internal muscle-tendon forces, joint loading, the potential for injury, and the rate and likelihood of healing. For example, foot orthoses (FOs) that decrease loading at the rearfoot could decrease activity of the tibialis posterior (TP) muscle and subsequently reduce strain of the TP tendon, a structure vulnerable to tendinopathy.[2] Clinicians can influence the forces applied to feet and muscles/tendons using footwear, FOs and taping.

Footwear that may have therapeutic benefits by altering loading of the foot include 'motion control' shoes, (typically running shoes), and rocker/rollover shoes. Whilst motion control shoes with dual density midsoles reduce calcaneal eversion by 2.77° ($p<0.001$, 95% CI 1.74° to 3.81°),[3] whether these changes impact on muscle function and injury risk is unknown. Under the "preferred movement pathway" theory,[4] footwear or FOs reduce muscle activity and metabolic demand[4, 5] by promoting the path of "least resistance" and reduce injury risk.[4-6] However, muscle activity could also increase to keep foot kinematics within the preferred pathway.[6] Rocker or rollover shoes have outsoles curved in the sagittal plane and alter the contact area between the shoe and floor, plantar load, external sagittal plane joint moments, and thereafter muscular responses and joint motion.[7] A recent review, however, found few statistically significant effects of the Masai Barefoot Technology (MBT) shoes on lower limb muscles.[8] The effect of other rocker and motion control footwear on EMG data has not been reviewed, so the use of these specialised shoes for treatment and injury prevention is unclear.

Foot orthoses redistribute plantar pressure, altering external joint moments, internal joint moments (from muscles and connective tissue), and foot motion. Examples include insoles with rearfoot wedges and arch supports,[9] a.k.a. "anti-pronation" FOs or medial posted FOs. Although FOs reduce peak rearfoot eversion by 2.08° to 2.35° ($p \leq 0.004$) depending on their design,[3] such small changes may not be clinically meaningful.[10] Foot orthoses can change ankle moments[11-13] with peak and mean ankle eversion moments reduced by $1.1 \pm 1.1\%$ ($p= 0.003$) and $2.3 \pm 2.1\%$ ($p< 0.001$) per 2° of medial posting respectively.[13] Such changes would alter the requirements of tissues acting antagonistically to the external moments, including muscles. The evidence for changes in muscle function with FOs that alter joint kinematics and kinetics is important in understanding injury risk and tissue repair.

Low-Dye taping is a temporary intervention for conditions supposedly associated with foot pronation or flat-arched feet.[14, 15] Theoretically applying tension to the skin using tape offloads structures in the medial arch.[3, 14, 16] Taping has been reported to only reduce foot pronation by a non-significant 1.50° ($p = 0.19$, 95% CI=−0.73° to 3.73°).[3] Plantar sensory stimulation is considered an important difference between FOs and taping since changes in afferent feedback due to tape might alter muscle activation.[14] However, there is no evidence to support this theory.

Prior reviews investigating the effects of footwear, FOs and taping did not compare device effects.[8, 16, 17] Also, approaches to searching and appraisal of literature was variable and underpin the need for a more comprehensive review. Indeed limitations of prior studies include low power, inadequate reporting of electromyography (EMG) procedures and low external validity.[16, 17] A review of foot posture, FOs and footwear by Murley et al.[17] allowed comparisons to a barefoot control, which is less clinically generalizable than a shod control and the review was broad, including all types of FOs and inserts and all footwear, not just that intended to alter foot biomechanics. The present review includes only FOs with a medial arch profile and or medial heel/foot wedge in order to improve our understanding of the relationship between medial FOs and muscle activity. Furthermore the recent review of MBT footwear did not assess the quality of EMG data reporting, limiting our understanding of the strength of the evidence identified.[8] Several studies on the effect of external devices on EMG have been published since these reviews, some of which have reported detailed EMG methods,[18] thus further justifying an update on the literature consensus. The aim of this systematic review was to investigate the level of evidence from any study design that investigated whether footwear, FOs and taping alter lower limb EMG during walking and running, irrespective of health status.

3.11 Methods

3.11.1 *Search strategy*

A systematic, electronic database search was performed by reviewer J.R. using CINAHL (1982-2017), MEDLINE (1950-2017), ScienceDirect, SPORTDiscus (1985-2017) and Web of Science (1900-2017) in October 2015 and updated in March 2019. The review conformed to the PRISMA guidelines for systematic reviews, however we were unable to account for biases like publication bias.[19] Searched words are included in **Table 3-1**. Lines 1-3) were combined using "AND" with lines 4) and 5). Additional sources were identified from published reviews and the reference lists of studies that passed the quality screening.

Table 3-1. Inputs to the electronic databases (all databases). Lines 1-3) were combined using "AND" with lines 4) and 5)

Search words
1) orthot* OR insert OR wedge OR orthos$ OR insole OR skive
2) foot* OR feet OR shoe$ OR footwear OR motion control shoe OR Nike free OR pronation control OR heel*
3) tape OR taping OR augmented Dye OR low-dye OR low dye
4) electromyograph$ OR EMG OR IEMG OR muscle function
5) Walk* OR run* OR gait OR locomotion or jog*

3.11.2 *Inclusion criteria*

The search results were assessed for eligibility based on titles and abstracts of original, full text articles using the following inclusion criteria:

1) A clearly defined amplitude, timing or frequency EMG outcome measure from muscles of the lower limb.
2) A fully specified independent variable of any footwear designed with modifications in the shape or material of the sole (including a negative heel, but excluding high heels, ankle braces and ankle destabilisation devices), foot orthoses/insoles (orthosis had a medial arch profile and or medial heel/foot wedge, excluding lateral wedges and ankle-foot orthoses), and taping about the foot/ankle intended to reduce foot pronation (excluding Kinesio taping).
3) Measures were made during level walking or running.
4) The footwear, FOs or taping experimental conditions were compared with a shod control condition.
5) For FOs and taping experimental conditions trials were performed in shoes, not sandals, with all the standard components of a shoe that brace the FOs.
6) Participants were free from conditions affecting the neurological systems.
7) Data was analysed from a minimum of three trials per condition.
8) Full text was published in English, French or German (due to available expertise).
9) Sample size of n > 1.

Only studies on locomotion were included since major theories on mechanisms of therapeutic effect of external devices relate to gait not standing.[20] Studies that only compared the device to barefoot were excluded because EMG amplitude can increase due to shoes alone and FOs versus barefoot (+30% and +30-38% respectively in tibialis anterior (TA)).[21] Articles were excluded if there were less than three trials per condition because without contradictory evidence, this was considered the minimum required for quality data. We did not restrict studies to a specific population as we took a mechanistic approach to understanding potential effects of external devices on muscle activity.

3.11.3 *Quality assessment*

To maintain quality standards in this systematic review the articles that met the inclusion criteria were subject to two levels of quality assessment (**Table 3-2**), performed independently by reviewers J.R. and E.P. After studies were assessed the two reviewers met to discuss discrepancies. When discrepancies persisted these were discussed with a third reviewer (L.B.) and a final score obtained.

The first stage focussed on the quality of the EMG methodology based on external standards of reporting,[22] plus controlling locomotion velocity (since velocity can affect EMG).[23] Studies scored a 1 or 0 depending on whether the criteria was fulfilled or not and the results were summated and expressed as a percentage. Studies achieving less than 50% were excluded.

The second stage of assessment was based on a modified sub set of a checklist for rating clinical interventions.[24] Studies were given 1 or 0 depending on whether each criteria was fulfilled, with the total score expressed as a percentage and studies that scored less than 50% were excluded.

Table 3-2. Quality assessment criteria checklists

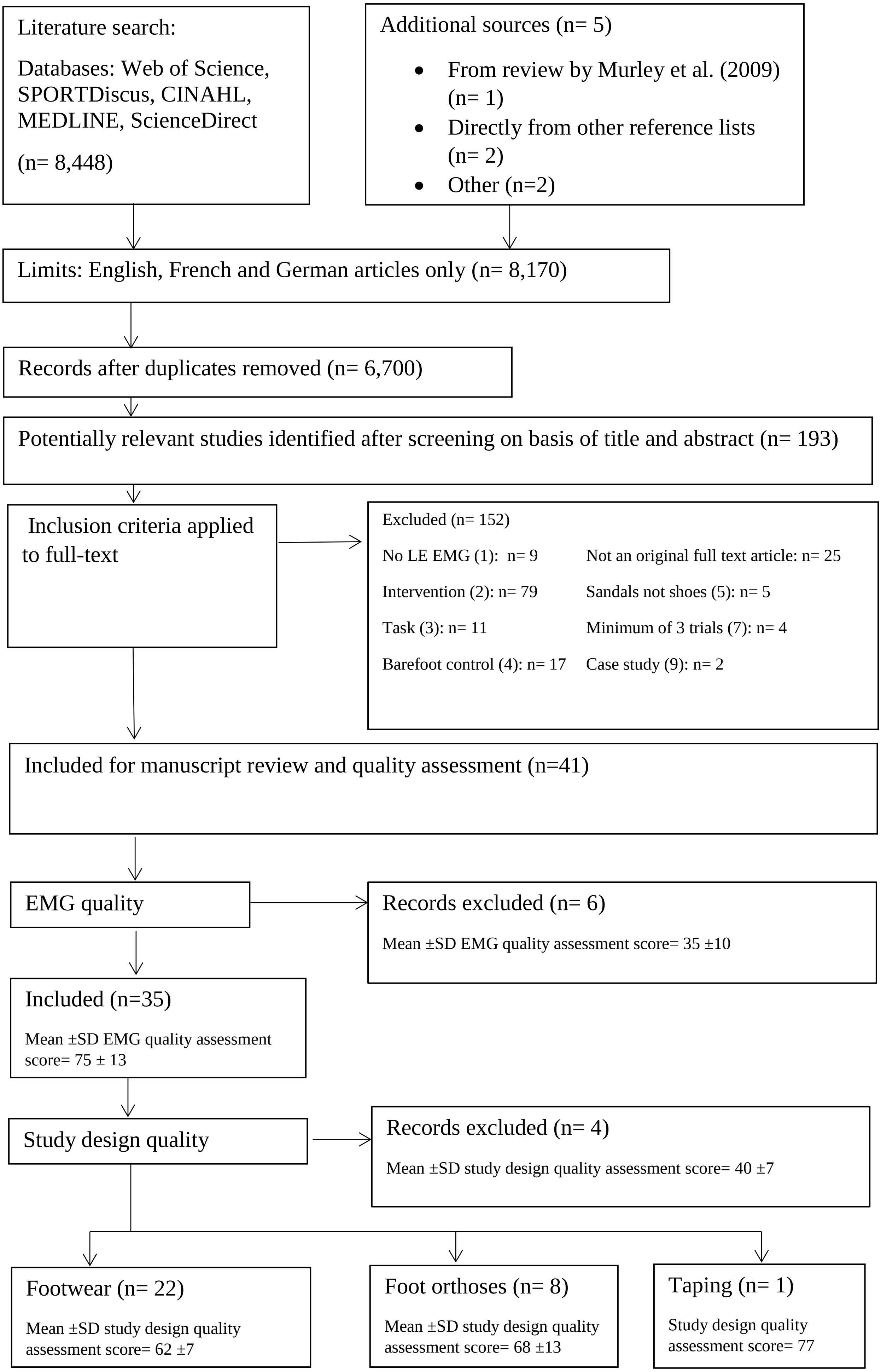
Criteria
First stage: EMG reporting
1) Surface sensors (shape, material, size, inter-electrode distance)
2) Adequate skin preparation
3) Fixation of the sensors was described or reference was made to guidelines for sensor placement[31]
4) Sensor location was based on SENIAM guidelines or a justified alternative (orientation over the muscle belly was made with consideration of fibre direction and with respect to tendons and the motor point, inter-electrode distance was reported)
5) Appropriate signal processing (including specification where applicable of full or half-wave rectification or window size of RMS[22])
6) Walking or running velocity was controlled (not just reported)
7) Adequate description of normalisation procedure if applicable
Fine-wire EMG studies were excluded if insufficient details of the intramuscular wire electrodes (type and material) and procedures (insertion approach and method of establishing a correct insertion) were reported[22]
Second stage
Immediate effects studies
1) Statement of an aim/hypothesis
2) Participant characteristics reported
3) Outcomes described in the introduction or methods
4) Device material reported
5) Clear description of the main findings
6) Variability reported (within written results or figures)
7) Actual p values stated (or <0.001)
8) Participant blinding (e.g. sham FO)
9) Assessor blinding
10) Appropriate statistics (including checking data for normality and sphericity where appropriate)
11) Control condition a true control
12) Randomisation of the order conditions were tested
13) Power calculation performed
Additional criteria for studies on the effects of devices over time
14) A description of the intervention (including duration)
15) Comparable participant characteristics across groups
16) Compliance

3.12 3.4 Results

3.12.1 *Search results*

A flow chart of the selection process from identification to screening and eligibility and inclusion[19] is presented in **Figure 3-1**, six studies were excluded based on EMG quality and four studies were excluded based on study design quality. A total of 31 studies were included, of these, 22 related to footwear and eight to FOs. Key themes of footwear studies were running shoes, rocker footwear, APOS-Therapy shoes[25, 26] and the Reebok EasyTone® shoe. Only one taping study (low-Dye) was identified which passed quality assessment. No study from additional sources met the inclusion criteria. Two studies were translated from German, but did not meet the inclusion criteria. Most studies included healthy, often recreationally active participants, except two studies involving participants with knee pain/knee osteoarthritis,[27, 28] one with running related overuse injuries[29] and one with Achilles tendinopathy.[30] Summaries of included studies are in **Table 3-3** and **Table 3-4**. A summary of excluded studies is included in the Review appendix in **Table 3-5**.

Figure 3-1. Study selection flow chart



3.12.2 *Quality assessment*

The included studies scored 50-100 on EMG quality (mean ±SD: 75 ±13). Unfulfilled criteria for EMG quality spanned categories 1-6. Almost all studies were deemed to have provided adequate details on normalisation where appropriate, although none as detailed as published recommendations e.g. training to produce a maximum voluntary contraction (MVC).[22] Of included studies 14/31 did not specify EMG sensor fixation and 13/31 did not control velocity.

Based on the second stage of quality assessment, the included studies were generally of moderate quality (scored 54-85, mean ±SD: 64 ±9). Excluded studies scored 31-46 (mean ±SD: 40 ±7). Many studies did not report device material, participant and assessor blinding, statistical analysis or power analysis. Variability was not reported in 3/4 of excluded studies. Additionally condition randomisation was absent in 8/31 of the included studies and 2/4 of the excluded studies.

Table 3-3 Summary of studies on the effect of footwear (n=22) and taping (n=1) on lower limb muscle activity

Authors	Participant characteristics	Device	Muscles	Walking or running	Variables	Main findings	QA scores, 1st and 2nd stage (%)
Bucheker et al. (2012)[38]	10 overweight males: 32.0 ±7.9 years, 1.792 ±0.058 m, 91.3 ±7.0 kg	Masai Barefoot Technology, participant's own shoes as control	BF, MG, VL	Walking (15 m walkway)	Intensity (amplitude) and co-activation indices	• In midstance mean intensities of VL ($p< 0.05$) and VL and MG co-activation ($p <0.05$) increased with MBT • In terminal stance mean intensities of VL ($p< 0.05$), MG ($p< 0.05$), and VL and MG co-activation ($p< 0.05$) increased in MBT	71, 62
Burgess and Swinton (2012)[39]	23 healthy, recreationally active females: 20.8 ±1.3 years, 1.654 ±0.056 m, 62.9 ±11.9 kg	Barefoot, Fitflop™ and flip flop treadmill walking, stair climbing and zigzag walk around cones	BF, Glut Med, MG, RF	Treadmill walking (1.34 m/s)	Normalised mean RMS	• No significant differences between footwear conditions	86, 62
Chen et al. (2018)[40]	Ten healthy males: 25.58 ±3.64 years, 1.737 ±0.02 m, 59.86 ± 3.80 kg	Barefoot, sports shoes (Roshe Run, Nike Inc., Oregon, United States) and flip-flops (flat rubber sole, Flipper, Adidas, Germany)	BF, GM, PL, TA, VL	Walking (10 m walkway)	Co-contraction index	• No significant differences between conditions in co-contraction for any muscle pairs	71, 62

Cheung et al. (2009)[36]	20 novice F runners, rearfoot pronation >6°, 25.8 ±3.7 years, BMI: 20.54 ±1.27 kg·m/2	"Supernova control", (Adidas), designed to check excessive pronation; "Supernova cushion" (Adidas, control), designed to reduce impact rate	VL, VM	Treadmill running (10 km)	EMG onset timing and median frequency	• VM activated ~5.3% (95% CI 4.5 to 6.1) of a duty cycle earlier than VL with motion control shoe • Neutral shoe: delay in VM activation by ~4.6% (95% CI 3.9 to 5.3) of a duty cycle compared with VL	83, 54
Cheung et al. (2010)[36]	20 novice F runners, rearfoot pronation >6°, 25.8 ±3.7 years, BMI: 20.54 ±1.27 kg·m/2	"Supernova control", (Adidas, motion control), designed to check excessive pronation; "Supernova cushion" (Adidas, neutral), designed to reduce impact rate	PL, TA	Treadmill running (10 km)	Normalised RMS and median frequency	• Positive correlations between RMS EMG and running mileage in TA and PL in neutral shoe condition ($p< 0.001$) • Median frequency dropped in both shoe conditions with mileage, but significantly larger drop in neutral shoe than motion control shoe ($p< 0.001$ for PL, $p= 0.074$ for TA)	86, 54
Elkjaer et al. (2011)[41]	10 healthy males: 24.5 ± 3.8 years, BMI = 24.03 ± 1.09 kg·m/2	Reebook EasyTone® ET Calibrator; neutral Nike Lunarglide +2 (control)	BF, Glut Max, LG, TA, VL	Treadmill walking	Peaks and integrated	• No significant differences between footwear conditions	57, 62

Forghany et al. (2014)[7]	20 healthy subjects (12 M): 33.1 ± 8.4 years, 1.71 m ±0.04 m, 68.9 kg ±12.1, BMI 23.6 ± 4.1 kg·m/2	Rollover shoe, flat control shoe (same leather upper and last as rollover shoe), flat control footwear weighted to equivalent of the rollover shoe and MBT shoe. All insoles were removed and replaced with a 1.2 mm poron insole.	Lateral BF, ES (right), Glut Max, MG, RF, SOL, TA	Walking (10 m course)	Peak EMG and integral of the signal	• Maximum at initial contact for TA: -29% for MBT, -22% for rollover shoe vs. control • iEMG: TA -17% both MBT and rollover vs. control; SOL +13% MBT, +8% rollover, MG +8% for MBT	50, 62
Franklin et al. (2018)[42]	70 healthy males (age range 20–87 years). YOUNG<40 years (n =20), MID>40 years and<70 years (n =30) and OLD>70 years (n = 20)	Minimalist shoe (Product ID: 2169, Two Barefeet Boarding Co.), control shoe (Style Code: 10001, Hobos Womens, Style Code: 50109, Hobos Mens), barefoot and own shoes	MG, PL, TA	Walking	Mean amplitude acros gait cycle and at separate phases of gait cycle (EMG only normalised in graphs)	• Higher amplitude of GM in minimalist shoe and own shoe vs. control in YOUNG and MID group but not old. • Lower amplitude of PL in minimalist shoe vs. own shoe and control in YOUNG and MID group but not old. • Lower amplitude of TA in minimalist shoe vs. own shoe and control across gait cycle and at initial double support • Slower walking speed in minimalist shoe vs. own shoe and control, but less <5% difference	67, 69

Goryachev et al. (2011)[27]	14 females with symptomatic bilateral medial compartment knee OA for ≥6 months, 59.9 ±6.2 years, 1.607 ±0.06 m, 77.4 ±8.9 kg	APOS-Therapy shoes in "functional neutral configuration", without elements, lateral 1.2 cm, medial 0.8 cm (both elements).	BF, LG, MG, ST, TA, VL, VM	Walking (10 m walkway)	ARV, normalized activity duration and peak	• In less symptomatic leg, almost all muscles varied significantly with COP in at least one phase of stance • In more symptomatic leg, significant differences in ARV across COP configurations for LG in terminal stance, pre-swing and terminal contact, for TA in pre-swing and for VL at contact • Training element of the study did not meet inclusion criteria	50, 69
Goto and Abe (2017)[43]	17 females (19.3 ±0.9 years, 1.577 ±0.04 m, 20.2 ±1.8 kg/m2)	Ladies leather safety footwear (670 g; longitudinal stiffness 35.8 N; MIDORI ANZEN Co., Ltd., Tokyo, Japan, hard sole and hard-resin toe cap). Control sports shoes: (470 g; longitudinal stiffness 14.7 N; Bridgestone Corporation, Tokyo, Japan, soft sole, no toe cap).	BF, LG, TA, VL	Treadmill walking	Mean EMG amplitude of safety shoes normalised to amplitude of control shoes	• Significantly higher amplitude of safety shoes for BF (114.3% ±20.7%, p= 0.01), TA (105.8% ±10.8%, p=0.04) and VL (129.5% ±47.1%, p=0.02) vs. control (100%). No significant difference in LG amplitude in safety shoes (103.3% ±7.7%, p=0.09) vs. control (100%)	83, 69
Horsak and Baca (2013)[44]	7 M, 5 F: 25 ±4 years; 1.72 ±0.11 m; 67 ±11 kg	Reebok Easy Tone® (Reenew model), 2 weeks familiarisation. Participant's own shoes as control	VL, VM	Walking	Mean amplitude	• No significant differences between footwear conditions	67, 54

Horsak et al. (2015)[45]	Reanalysed data: 7 M, 5 F; 25 ± 6 years; 1.74 ±0.07 m; 68 ±10 kg and 7 M, 5 F: 25 ±4 years; 1.72 ±0.11 m; 67 ±11 kg	Reebok Easy Tone® (Reenew model), 2 weeks familiarisation and MBT shoe. Participant's own shoes as control	BF, Glut Med, MG, PL, TA, VM, VL	Walking (10 m walkway)	Mean amplitude, co-contraction indices	• No significant difference in mean muscle activity between unstable shoes and control • Increased co-contraction of vastii and gastrocnemius muscle in MBT, (Cohen's d 0.5-0.9)	71, 54
Kelly et al. (2010)[46]	13 male, recreational runners 31.7 ±4.9 years, 1.817 ±0.046 m, 81.6 ±5.9 kg	Augmented low Dye taping, control taping and Adidas Response Cushion running shoes	Glut Med, VL, VM	Treadmill running (6 mins)	Peak and average EMG signal amplitude, onset time, and burst duration	• Delayed onset of the EMG signal of all muscles with taping, moderate to large effect size	83, 77
Koyama et al. (2012)[47]	6 healthy males: 26.3 ± 5.3 years; 1.72 ± 0.05 m; 68.0 ±6.1 kg	Shape-ups (SKECHERS, USA) vs. normal walking shoe	RF, VL, BF, TA, SOL, MG	Treadmill walking at 3, 4, 5, 6, and 7 km/h (3 mins)	Integrated EMG (iEMG) calculated relative to control shoe	• Significantly higher iEMG of MG (6–16%, p< 0.05) and SOL (8–23%, p< 0.01) in Shape-ups across speeds. • Tendency towards higher iEMG in Shape-ups vs. control in RF, VL, BF and TA	86, 54
Nigg et al. (2006)[48]	5 M and 3 F: 28.0 ±3.6 years, 1.695 m ±0.064 m, 70.1 ±7.5 kg. Free of LE pain/ injury 6+ months prior to testing and never used MBT shoe before	Control: Adidas SuperNova running shoe (mass: 358 g). Experimental: MBT (mass: 650 g), rounded shoe-sole design in AP direction.	BF, Glut Med, MG, TA, VM	Walking (lab) and quiet standing	Intensity, wavelet analysis	• No significant differences changes in EMG intensity • Trend for reduced intensity of TA (-26% ±24%) and BF of (-55% ±60%), • Trend for increased intensity of MG (+52% ±82%), VM (+4% ± 13%) and Glut Med (+16% ± 25%)	86, 62

O'Connor and Hamill (2004)[35]	10 healthy, recreationally active males, 27 ±5 years, 1.72 ±0.07 m, 72.6 ±5.3 kg. Only 4 subjects with full sets of data for TP	EVA with a durometer of 45 (Shore A). Neutral shoes constructed with heel height 2.5 cm. For 8° varus configuration, the medial aspect of the midsole at the heel was 3-cm thick, and lateral aspect 2 cm thick. Dimensions reversed for valgus shoe.	LG, MG, PL, SOL, TA, TP	Running (30 m walkway)	Integrated EMG (iEMG), mean amplitude, onset and offset	• No significant differences between footwear conditions	71, 54
O'Connor et al. (2006)[34]	10 healthy, recreationally active males, 27 ±5 years, 1.72 ±0.07 m, 72.6 ±5.3 kg. Only 4 subjects with full sets of data for TP. Same subjects and materials as O'Connor and Hamill (2004)	EVA with a durometer of 45 (Shore A). Neutral shoes constructed with heel height 2.5 cm. For 8° varus configuration, the medial aspect of the midsole at the heel was 3-cm thick, and lateral aspect 2 cm thick. Dimensions reversed for valgus shoe.	LG, MG, PL, SOL, TA, TP	Treadmill running (5 mins)	Mean amplitude, onset and offset	• Significantly less mean EMG activity in the TA and SOL in the neutral shoe vs. either wedged shoe • TA amplitude increased 16% in varus (medial wedge) shoe vs. neutral shoe	71, 69

Price et al. (2013)[50]	15 healthy females: 29 ±6.7 years, 1.671 ±0.042 m, 62.6 ±6.9 kg,	Earth sandal (control), FitFlop, Masai Barefoot Technology, Reebok Easy-Tone and Skechers Tone-Ups	BF, MG, PL, RF, SOL, TA	Walking (lab)	Median RMS for phase	• Fitflop, Reebok and Skechers increased PL activity during pre-swing, whereas MBT increased MG and decreased TA activity in loading response and mid-stance • Increased PL activity in loading response in MBT vs. control • SOL activation during midstance was lower in Fitflop and Skechers than MBT and control	86, 77
Price et al. (2014)[49]	15 M: 30 ±8 years, BMI: 25.9 ±4.5 kg·m/2; 13 F: 37.8 ±12.4 years, BMI: 23.0 ±4.7 kg·m/2	Barefoot. Flip-flop (Havaiana, Brazil), EVA midsole. Fit-flop: Walkstar I. for females, Dass for males, multi-density EVA in heel, midfoot and toe. Rubber outsole	PL, TA	Walking (lab)	Amplitude	• No significant differences between footwear conditions	83, 62
Sacco (2012)[51]	25 healthy females with no experience of MBT (21.8 ±3.0 years, 1.610 ± 00.4 m, 52.6 ±5.3 kg)	Barefoot, MBT (501 g), and standard tennis shoe (Rainha System, Alpargatas, Brazil, 171 g, neutral strike).	BF, Glut Med, LG, TA, VL	Walking (10 m walkway)	Peak, time of peak and integral of the envelope	• Less peak TA amplitude in MBT vs. standard shoe and barefoot ($p < 0.01$) • Walking with the MBT shoe did not increase muscle activity when compared to walking with the standard shoe	83, 62

Scott et al. (2012)[32]	28 adults with flat feet (14 M/F), 21.2 ±3.8 years, 1.71 ±0.1 m, 73.3 ±16.0 kg	Standard flexible shoe (Dunlop Volley), stability running shoe (Nike Air Structure Triax +10, range of features aimed at controlling moderate pronation) and barefoot	MG, TA (surface), PL, TP (fine-wire)	Walking (9 m walkway)	Time of peak amplitude and peak amplitude	• Both styles of footwear increased TA peak amplitude and decreased PL peak amplitude vs. barefoot • Little difference between footwear conditions	83, 69
Sobhani et al. (2013)[52]	16 healthy runners (8 M/F), 29 ±9 years, 1.771 ±0.093 m, 69.8 ±11 kg	Standard shoe (apex/rolling point 53% of shoe length, proximal to metatarsal region, 467 ± 87 g). Modified rocker shoe (rolling point 65%, 805 ±157 g)	LG, MG, SOL, TA	Slow running and walking (10 m lab)	Peak and time of peak (%)	• Significant delay of EMG peak, ~2% ($p < 0.001$) in triceps surae walking with rocker shoes • No change in peak amplitude of triceps surae in running/walking • Peak amplitude of TA increased 20%, 64.7 mV, $p < 0.001$) walking with rocker shoes	67, 69
Sobhani et al. (2015)[30]	13 Achilles tendinopathy patients (11 F), 48 ±14.5 years, 1.72 ±0.07 m, 77 ±14 kg. Achilles tendinopathy 4 months to 9 years (mean 22.5 months, median 11.5 months)	Standard shoe (apex/rolling point 53% of shoe length, proximal to metatarsal region, 467 ± 87 g). Modified rocker shoe (rolling point 65%, 805 ±157 g)	LG, MG, SOL, TA	Slow running and walking (10 m lab)	Peak and time of peak (%)	• Peak activity of TA increased (61.77 µV, 35%) for walking with rocker shoes ($p = 0.015$)• Delay of ~4% of the gait cycle in time of peak activity of LG ($p = 0.001$) in running	67, 62

*ARV= average rectified value, LE= lower extremity, M= male, F= female, RMS= root mean square, TP= tibialis posterior, TA= tibialis anterior, SOL= soleus, PL= peroneus longus, MG= medial gastrocnemius, LG= lateral gastrocnemius, AT= Achilles tendinopathy, VL= vastus lateralis, VM= vastus medialis, BF= biceps femoris, Glut Med= gluteus medius, Glut Max= gluteus maximus, ES= erector spinae, RF= rectus femoris, ST= semitendinosus, MBT= Masai Barefoot Technology, OA= osteoarthritis. QA= quality assessment, for first and second stage ((number of satisfied criteria/number of applicable criteria)*100).*

Table 3-4. Summary of studies on the effect of foot orthoses (n=8) on lower limb muscle activity

Authors	Participant characteristics	Device	Muscles	Walking or running	Variables	Main findings	QA scores, 1st and 2nd stage (%)
Akuzawa et al. (2016)[53]	10 healthy males: 25 ± 5.0 years, 1.68 ± 0.06 m, 61.5 ± 7.8 kg	Shoe (Calcetto Le3, Asics, Japan), shoe + prefabricated orthosis (Athlete grip7, Winning One Inc., Japan), barefoot	TP, FDL, PL	Walking	Amplitude as %MVC in contact, midstance and propulsion phase	• Significant reduction ($p<0.036$) in TP activity in propulsion phase with orthoses relative to barefoot but not relative to shoe • No significant difference in FDL and PL EMG between conditions	57, 62
Baur et al. (2011)[29]	99 runners with running-related overuse symptoms. 50 M, 49 F. CO: 37.1 ±8.3 years, 1.74 ±0.09 m, 68.8 ±13.6 kg. FOs: 37.3 ± 8.2 years, 1.73 ±0.09 m, 66.8 ±11.6 kg	Custom, MLA support (25 mm), a detorsion wedge in the forefoot (lateral post, 3 mm), and a bowl-shaped heel. 8 week intervention	PL	Treadmill running	Activation time and mean amplitude	• Sig ($p= 0.001$) increase in preactivation amplitude of 22% ± 48% (95% CI = 9%–32%) in OR compared with CO	71, 75
Kelly et al. (2011)[54]	12 male recreational athletes (31.2 ±3.8 years, 76 ±3.9 kg, 1.808 ±0.04 m)	Prefabricated Formthotics (Foot Science International)	MG, PL, TA, VM	Treadmill running	Burst duration and average RMS amplitude	• Lower RMS signal amplitude VM (-13.3%, $p< 0.02$) and MG (-10.7%, $p< 0.05$), increased PL burst duration (+14.7%, $p< 0.05$), running with orthoses	86, 54

Maharaj et al. (2018)[2]	18 adults with flat feet recruited: 5 F, 13 M (14 included in analysis) 26 ±5 years, 1.70 ±0.11 m, 71.3 ±12.6 kg	Shoe:(Gel Lyte 33, Ascics, Japan), shoe + Custom FO: ¾ length semi-rigid 4 mm polypropylene thermoplastic shell with vinyl covering, 4 mm medial skive at 15° and a 5° extrinsic rear foot post, barefoot	TP	Treadmill walking	Amplitude as % of max at preferred walking speed	• Reduced TP activity with shoe and shoe + FO vs. barefoot in early stance (1-12%) and late stance (19-22%), main effect of condition ($p \leq 0.01$), but no significant difference between shoe and shoe + FO.	100, 54
Mills et al. (2012)[28]	40 patients with knee pain. 27 mobile (foot): 28.67 ±6.13 years, 1.696 ±0.149 m, 71.03 ± 11.97 kg. 13 less mobile: 31.15 ±4.41 years, 1.71 ±0.0841 m, 71.15 ± 11.22 kg	Prefabricated EVA FOs with varying hardnesses	BF, Glut Med, MG, RF, SOL, TA, VM, VL	Treadmill jogging (3 min intervals)	Peaks and temporal (only offset reported)	• Orthoses, regardless of comfort, had no immediate effect on lower limb EMG or kinematics compared with baseline shoe conditions • Moderate difference in VL peak amplitude ($p = 0.007$) between most and least comfortable orthosis, greatest increase in peak amplitude in least comfortable	71, 69

Murley et al. (2010)[9]	30 adults with flat feet, 21.8 ±4.3 years, 1.71 ±0.1 m, 73.3 ±15.5 kg	Modified prefabricated FO: ¾ length, medial heel wedge under heel, arch support heat-moulded to individual. Custom FO: ¾ length, posted at 20° inverted, heel supported by EVA wedge, plaster cast modifications to contour shell to arch	MG, PL, TA, TP	Walking (9 m walkway)	Time of peak amplitude; (RMS); peak amplitude	• In contact phase TP amplitude decreased with prefabricated orthosis (peak amplitude -19%, p= 0.007; RMS amplitude -22%, p= 0.002) and custom orthosis (peak amplitude -12%, p= 0.001, RMS amplitude -13%, p= 0.001), vs. shoe-only • During midstance/ propulsive phase PL EMG amplitude increased with prefabricated orthosis, vs. shoe-only (peak amplitude +21%, p= 0.024; RMS amplitude +24%, p= 0.019) and custom orthosis (peak amplitude +16%, p= 0.028)	57, 85
Murley and Bird (2006)[21]	Pronated foot type: 10 F, 5 M, 23 ±5 years, 1.702 ±0.09 m and 69.9 ±14.4 kg	3 pairs of rigid custom-made foot orthoses (posted at 0°, 15° and 30° inverted)	MG, PL, SOL, TA	Walking (walkway)	Maximum amplitude as % of MVC, onset	• Increased maximum TA amplitude using shoe only (+30%), 0° (+33%), 15° (+38%) and 30° (+30%) inverted orthoses conditions vs. barefoot ($p < 0.01$)• PL maximum amplitude increased using the 15° inverted orthosis condition vs. barefoot (+21%, p= 0.04), trend for an increase vs. shoe only	57, 62

Telfer et al. (2013)[18]	12 pronated and 12 gender matched controls 29.9 ±8.7years, 1.71 m ± 0.08, 71.6 ±10.7 kg	9 variations: level of external rearfoot posting modified from 6° lateral to 10° medial in 2° increments.	BF, LG, MG, PL, SOL, TA, VL, VM	Walking (indoor walkway)	Peaks and means	• No main effects due to posting level • Group effects customised FOs reducing above knee muscle activity in pronated foot types compared to normal foot types (BF mean p= 0.022; VL peak p< 0.001; VM peak p= 0.009; VM mean p= 0.001) • Interaction effect peak MG (p= 0.034) and peak SOL p= 0.015)	100, 85

*LE= lower extremity, M= male, F= female, RMS= root mean square, TP= tibialis posterior, TA= tibialis anterior, SOL= soleus, PL= peroneus longus, MG= medial gastrocnemius, LG= lateral gastrocnemius, AT= Achilles tendinopathy, VL= vastus lateralis, VM= vastus medialis, BF = biceps femoris, Glut Med= gluteus medius, ES= erector spinae, RF= rectus femoris, CO= control, FO= foot orthosis, MLA= medial longitudinal arch, EVA= Ethyl Vinyl Acetate. QA= quality assessment, first and second stage ((number of satisfied criteria/number of applicable criteria)*100).*

3.12.3 <u>*Overview of included studies*</u>

3.12.3.1 *Footwear*

3.12.3.1.1 Running shoes

A stability running shoe with a dual-density medial post and foot bridge had no effect on EMG activity during walking versus a standard flexible shoe for the peroneus longus (PL), medial gastrocnemius (MG), soleus, tibialis anterior (TA) and TP.[32] However a motion control running shoe with a dual-density midsole (firmer material on the medial versus lateral side), reduced mean TA and PL activity and delayed fatigue during running[33] versus a cushioned running shoe. Delayed fatigue was demonstrated by maintained median frequency of TA and PL during a 10 km run.

A shoe with a medial wedge (3 cm thick on the medial side, 2 cm thick on the lateral side) increased mean TA amplitude by 16% during treadmill running versus a neutral shoe,[34] but not during overground running.[35] The activity of the gastrocnemii, soleus, PL and TP were not affected by medial or lateral wedges.[34, 35] However TP EMG data was only available for four participants due to measurement difficulties.[34, 35]

The relative timing of vastus medialis (VM) and vastus lateralis (VL) activation during running was compared between a motion control running shoe with a dual-density midsole and a cushioning running shoe.[36] This comparison was made based on the premise that delayed onset of VM with respect to VL is associated with patellofemoral pain.[36] The authors normalised EMG signals to a “duty cycle” (defined in the animal literature as stance expressed as a percentage of step cycle, i.e. stance + swing).[37] In the motion control shoe activation of VM occurred ~5.3% (95% CI 4.5% to 6.1%) of a duty cycle earlier than VL (during a 10 km run). In contrast VM activation occurred ~4.6% (95% CI 3.9% to 5.3%) later than VL in the neutral shoe.[36] The implications of the findings are limited by the ambiguity of the reporting of the methods.

3.12.3.1.2 Rocker footwear

Tibialis anterior amplitude in early stance in walking reduced by ~30-40% ($p < 0.05$) with rocker footwear like MBT, which is curved under the heel in the sagittal plane, versus flat heeled conventional footwear.[7, 50, 51] There was also a trend towards reduced TA EMG intensity when walking in MBT versus a running shoe.[48] A modified shoe with a forefoot rocker only *increased* peak TA activity in walking (by 20-35%, $p < 0.001$, $p = 0.015$ respectively), but not in running.[30, 52] The shoe did not alter triceps surae (TS) activity in late stance during either walking or running.[30, 52]

Increased PL activity throughout stance (e.g. 50% at loading response, $p = 0.02$) has been shown with MBT.[50] Other footwear in that study, including FitFlops™, designed to be unstable in the sagittal plane, increased peroneal activity during pre-swing. However, later work found no difference in PL activity with FitFlops™[49] or difference in co-contraction with regular flip-flops.[40] A minimalist shoe reduced TA activity and increased plantar flexion in early stance relative to control footwear, however walking speed was also slower.[42]

Changes in TS activity during loading response with rocker footwear are opposite to that of TA.[7, 50] The integral of the EMG profile in rocker footwear was 8-13% ($p < 0.05$) greater than the control shoe for the soleus and 5.5-8% for the MG (significant for MBT, $p < 0.05$, but not other rocker shoe, $p > 0.05$).[7] Similarly, integrated EMG of MG was 6-16% ($p < 0.05$) higher and 8-23% ($p < 0.01$) higher in soleus in a rocker shoe compared with a regular walking shoe in treadmill walking.[47] There was also a trend towards increased MG EMG intensity walking in MBT compared with a running shoe in another study.[48] Activation of MG was unaffected when wearing a FitFlop™ sandal with a variable density sole.[39, 50]

Studies recording quadriceps activation during walking have generally found no effect of rocker footwear.[7, 48, 50, 51] However increased activation of VL and greater co-contraction of vastii and MG across stance was found in MBT.[38, 45] Activation of biceps femoris or rectus femoris was unaltered by a FitFlop™ sandal.[39] Stiff soled safety shoes significantly increased VL, biceps femoris and TA activity relative to a soft soled trainer.[43]

3.12.3.1.3 APOS-Therapy shoes

APOS-Therapy shoes have adjustable domes on the sole allowing manipulation of COP position and external joint moments.[25, 26] In females with knee osteoarthritis a lateral shift in the sole domes and COP reduced averaged TA EMG amplitude in pre-swing versus a neutral dome configuration.[27] The EMG amplitude of the lateral gastrocnemius increased with a medial shift in COP and decreased with a lateral shift in COP due to APOS-Therapy shoes (compared with neutral).[27]

3.12.3.1.4 Reebok EasyTone® shoe

The Reebok EasyTone® shoe, designed to be unstable with balance pods, did not alter muscle activation in walking of any thigh, shank or gluteal muscles.[41, 44, 45]

3.12.3.2 *Foot orthoses*

There is some limited evidence that FOs decrease activity of TP in early stance and increase activity of PL in mid-late stance.[9, 21] Peak amplitude and RMS amplitude of TP during loading response was shown to reduce by 19% ($p= 0.007$) and 22% ($p= 0.002$) respectively with custom FOs, and 12% ($p< 0.001$) and 13% ($p= 0.001$) respectively with prefabricated FOs.[9] Whereas PL activity increased in midstance with a prefabricated FOs (peak amplitude +21%, $p= 0.024$; RMS amplitude +24%, $p= 0.019$) and a custom FOs (peak amplitude +16%, $p= 0.028$) compared with a shoe only. Maximum PL amplitude has also been shown to increase in walking by 19% for pronated individuals when wearing 15° inverted FOs versus shoes alone ($p< 0.05$).[21] However, PL amplitude does not appear to increase linearly with wedging magnitude.[18, 21] Another two studies found TP activity was not significantly different between the footwear and FOs conditions ($p> 0.05$), although there was a decrease of around 10% ($p< 0.05$) from barefoot to shod and shod with either a prefabricated or custom FOs, which was not considered clinically generalizable in this review.[2, 53] However in the study that recorded kinematics and kinetics, there was also no effect of FOs on subtalar joint displacement or supination moment relative to the shoe condition.[2] There was no difference in flexor digitorum longus or PL activity between conditions.[53]

As for TA and TS, most evidence indicates magnitude and timing of activation is unchanged by wearing FOs during walking and running.[18, 21, 28, 54] One study found there was a tendency for FOs to decrease TA activation during walking versus a shoe only (effect size 0.18-0.29, for custom and prefabricated FOs respectively), although the result was not statistically significant.[9]

Activity of PL may also increase with FOs during running. In one study 99 runners with an overuse injury were assigned to customised FOs or no FOs.[29] In treadmill running, there was a significant increase ($p= 0.003$) in PL pre-activation amplitude (EMG activity prior to foot contact) after two months wearing FOs, but not in the control patient group.[29] It is unclear whether change was in barefoot running or the running shoes or both. Another study reported a 14.7% ($p< 0.05$) greater duration of PL activity (the muscle was active for longer) during running with prefabricated FOs compared with no FOs as well as lower average MG and VM RMS amplitude with FOs versus no FOs.[54]

3.12.3.3 *Low-Dye taping*

In the only study included involving taping a significant delay (5-7%, $p= 0.001$) in onset times of VM, VL and gluteus medius was found during shod running with low-Dye tape compared with control taping.[46]

3.13 Discussion

The aim of this review was to establish if there is evidence that footwear, FOs and taping alter muscle activity of the lower limb during walking and running. The effect of running shoe design, FitFlops[TM] sandals and low-Dye taping on muscle activity is unclear, while rocker/rollover shoes appear to affect muscle activity of MG and TS.[7, 47, 50] There is evidence, albeit limited, that FOs decrease activity of TP in early stance,[9] which could be beneficial in treating posterior tibial tendon dysfunction (PTTD). Activity of PL may increase in mid-late stance,[9, 21] otherwise FOs do not appear to alter EMG of lower limb muscles.[9, 18, 21, 28]

3.13.1 <u>*Footwear*</u>

3.13.1.1 *Running shoes*

The effect of running shoe design on muscle activity remains unclear due to uncertainty regarding whether the shoes tested were effective in changing loading. No study investigating the effects of running shoe design on EMG during walking or running collected simultaneous kinematics or kinetics.[32, 33, 36] Without kinetic data we cannot determine if the footwear changed loading of the foot, which might sometimes explain the absence of change in EMG. Concurrent collection of kinetic and EMG data would also be useful to establish if the difference between a nil effect of motion control shoes in walking and a reduction in fatigue during running are due to the greater forces in running, foot strike patterns, or different shoe properties.[32, 33]

Similarly sagittal plane kinematics were not reported in studies involving a medial wedged shoe.[34, 35] Wedging could increase ankle plantar flexion and increase demand on TA, potentially explaining the 16% increase in TA amplitude during treadmill running compared with a neutral shoe.[34] Perhaps a medial wedged shoe is substantially different to a motion control running shoe with a dual-density midsole if the effects of the wedge are not isolated to the frontal plane.

3.13.1.2 *Rocker footwear*

Footwear that shifts the COP anteriorly at heel contact reduces TA amplitude between initial contact and into midstance.[7, 50, 51] An anterior shift in the COP increased the external dorsiflexion moment, resulting in a more dorsiflexed ankle and less work required from TA to control plantarflexion after initial contact.[7] Increased external dorsiflexion moment/increased internal plantar flexion moment in early stance would also account for the increases in TS activity with some rocker footwear.[7, 47, 50] Potentially the increased PL activity in MBT shoes[50] is due to the need for PL to contribute to sagittal plane moments. In contrast the *increase* in TA activity in walking with the shoe with the modified forefoot rocker[30, 52] might be explained by the greater mass and sole thickness of the modified shoe versus the control. As TA is active in swing, greater activity during early stance could result from the greater moment of inertia not the sole curvature.

Rocker footwear have been shown to reduce internal plantar flexion moment in late stance, which could be beneficial for offloading the Achilles tendon when treating Achilles tendinopathy.[8, 30, 52] However reduced internal plantar flexion moment in late stance is not necessarily coupled to reduced TS activity in the same phase.[30, 50, 52] This could be because peak o MG activity is earlier in stance than the peak of the internal plantar flexion moment[7] and the energy recoil of the Achilles tendon is in terminal stance. Thus reduced loading of the Achilles tendon suggested by a reduced internal plantar flexion moment may still be beneficial in treating Achilles tendinopathy.

The curved sole of rocker footwear purportedly reduces contact area with the ground and thus reduces stability. Increased co-activation from TA and TS in early stance with MBT may increase ankle stability to compensate.[7] The induced instability is assumed to increase movement variability and activate muscles required to maintain balance and control movement.[55] Greater movement variability could be beneficial in managing chronic injury if it reduced the repetitive loading of injured structures.[56] Conversely increased co-activation increases joint loading.[57] Consequently the clinical implications of altered muscle activation and reduced internal plantar flexion with rocker footwear remains unestablished. Additionally there was no effect of the modified rocker shoe on pain in individuals with chronic Achilles tendinopathy and randomised clinical trials are necessary to evaluate the therapeutic effect of rocker footwear.[30]

3.13.1.3 *APOS-Therapy shoes*

Reduced TA and increased TS activation with APOS-Therapy shoes could be relevant to those with TS and Achilles injury, anterior compartment syndrome and intermittent claudication, but any implications remain speculative.

3.13.2 *Foot orthoses*

There is some limited evidence that FOs decrease activity of TP in early stance and increase activity of PL in mid-late stance,[9, 21] but otherwise there appears to be a lack of effect of FOs on lower limb muscle activity during walking.[9, 18, 21, 28, 54]

In two studies the activity of TP reduced with FOs with respect to barefoot, but not shoes alone in early stance (Maharaj et al., 2018) and late stance (Akuzawa et al., 2016) respectively. There are a few explanations as to why a lack of effect of FOs with respect to footwear was found. Firstly, there could have been a genuine lack of effect of the FOs geometry, or that it was too subtle to be detected as a significant difference within the variable TP EMG pattern. Secondly, in the case of the study by Maharaj et al. (2018), the stiffer insole material (semi rigid 4-mm polypropylene) compared to the EVA shoe liner may have negated any effect of the FOs geometry (Maharaj et al., 2018), or affected an alternative aspect of muscle function than activation. Thirdly, it is possible that measurement error due to a change in the recording capacity of the fine-wire electrode was larger than any small effect of the FOs.

A reduction in indwelling EMG amplitude can occur over time, as described in 6.2.4.3, and the Appendix, although this has yet to be documented in the literature at the time of this review. In this thesis and from previous work in the department, it can be estimated that there is around 30 minutes before an initial drop in the amplitude occurs as an artefact of the experiment, and the amplitude can drop again as the duration of data collection increases, irrespective of any condition being tested. The order of experimental conditions was not randomised in either study (Akuzawa et al., 2016, Maharaj et al., 2018) and was reported as being tested in the order of barefoot, footwear alone and footwear plus FOs. Without knowledge of the within session reliability of the EMG recordings nor the duration of the sessions, the results of these studies need to be interpreted with caution. Footwear may have reduced TP amplitude in early stance (Maharaj et al., 2018) and FOs may have reduced TP amplitude in late stance relative to barefoot (Akuzawa et al., 2016) , however we cannot rule out an order effect.

Three other studies in which TP EMG was recorded were excluded due to having only a barefoot control (Barn et al., 2014) and because the control condition was in a sandal not a shoe, and thus lacking the supportive elements of a shoe (Garbalosa et al., 2015, Stacoff et al., 2007). If these studies were included in the review the conclusion would not have been changed. Two of the studies indicated there is a possibility to reduce TP EMG with FOs (Barn et al., 2014, Garbalosa et al., 2015) and the third indicated no clear effect , although data was only available from three participants (Stacoff et al., 2007). A trend towards reduced TP activity in the contact period with custom FOs compared to barefoot (p = 0.09) was found in participants with a progressive flat foot deformity, rheumatoid arthritis and TP tenosynovitis. The absence of a significant effect may be due to an insufficient habituation period, a lack of power due to a small sample size (n=10) and large variation in pathology within the sample (Barn et al., 2014), as varying levels of inflammatory cytokines could affect muscle function (Barn et al., 2012). In the other study maximum EMG activity decreased wearing the maximal arch subtalar stabilisation (MASS) orthotic (22.8 ±17.4 to 20.9 ±19.6 peak normalized EMG ((mV/mV)*s)), but increased wearing the full contact FO (18.5 ±22.2 to 27.4 ±48.7 peak normalized EMG ((mV/mV)*s)) with respect to a sandal alone (Garbalosa et al., 2015). This discrepancy could be due to the different lengths and possibly bending stiffness of the FOs. As the primary invertor of the foot tibialis posterior (TP) acts eccentrically during early stance to generate an inversion moment that opposes the external eversion moment, and helps control rearfoot eversion. It also acts concentrically to support foot supination later in stance.[58] If FOs increase the external inversion moment, they might reduce required internal inversion moments, reducing TP activity. Reduced TP activation could mean less force through the TP tendon which could facilitate healing in pathologies like PTTD. The limited amount of evidence on the effect of FOs on TP is likely because indwelling EMG is required to measure TP activity. Further research with adequate power and concurrent collection of kinematic and kinetic data is needed to relate kinetic and kinematic changes to muscle activation.

A linear dose-response to extrinsic rearfoot posting during walking has been demonstrated in kinematic, kinetic and plantar pressure variables, but without a corresponding effect on any EMG related muscle activity in the calf muscles (including PL), quadriceps or hamstrings.[13, 18] Maximum PL amplitude did increase in walking by 19% for pronated individuals when wearing 15° inverted FOs versus shoes alone, but again without a linear dose-response to magnitude of wedging.[21] The lack of a dose response to medial rearfoot wedging could infer that the FOs exert their effect on PL due to changes in load under the medial longitudinal arch rather than the rearfoot.[21] The midfoot is in contact with the ground during midstance and the heel is unloading.[59] Similarly in later work by Murley and colleagues, flat-footed participants increased PL activity in midstance with prefabricated FOs (peak amplitude +21%, p= 0.024;

RMS amplitude +24%, p= 0.019) and custom FOs (peak amplitude +16%, p= 0.028) compared with a shoe only.[9] The original authors speculated that increased PL EMG amplitude resulted from the foot being more laterally unstable. If FOs increased the external inversion moment, greater PL EMG activation may be needed to maintain equilibrium. As PL is the antagonist of TP, if FOs reduced TP EMG activity this would possibly be accompanied by increased PL EMG activity. However, TP and PL activity do not necessarily represent equal opposing inversion and eversion moments respectively, due to additional muscle tendon parameters[60] and different moment arms.

Although FOs may increase amplitude and duration of PL activity during running,[29, 54] the literature is limited by low between-session reliability of PL EMG.[61-63] Reported poor inter-session reliability of EMG data from PL reduces confidence in EMG results collected weeks apart.[61-63] Amplitude of an EMG signal varies not only due to the detection of different motor units, but because of variable skin-electrode impedance between sessions.[64] Variability in amplitude between sessions could affect the ability to detect changes in duration of muscle activity due to FOs using threshold methods. As measurements were taken in separate sessions without mention of normalisation[29, 54] comparing EMG measures could be beyond this technique. Additionally electrode placement in one study[29] followed the methods of Winter and Yack[65] (50% of the distance between the fibular head and lateral malleolus, rather than 25% of the distance recommended by SENIAM).[31] A distal shift in surface electrode placement over PL of 2 cm increases the presence of crosstalk, likely from TA.[66] Given that PL is most active in mid-late stance, while TA is active prior to foot contact to enable a dorsiflexed ankle position at initial contact, potentially muscle activity reported as pre-activation of PL was actually crosstalk from TA.

Foot orthoses designed to reduce the external eversion moment at the subtalar joint would theoretically decrease the internal inversion moment required from the invertor muscles limiting eversion. The TA has an inversion moment arm when the foot is inverted, as at initial contact,[67] therefore FOs that reduce the eversion moment might also reduce TA activity. The conclusion of the review by Murley et al.[17] preceding the work of Telfer[13, 18] that FOs may *increase* activation of the TA should be reconsidered. Studies that found FOs increase TA activity had notable limitations. As Murley et al.[17] identified, a significant increase in EMG activation was not always supported by confidence intervals. Many studies did not simultaneously collect kinematic and kinetic data so we cannot relate any change (or lack of) in EMG to other changes in biomechanics, or evaluate the intervention in the context of the "preferred movement pathway" theory. For example, an extrinsic medial rearfoot wedge could place the foot in a more plantar flexed position since the heel is lifted in the shoe, perhaps increasing demand on the TA in the sagittal plane after initial contact.[34] Without kinematic and kinetic data and with variable changes in EMG, the implications of this finding are limited. As TA is not the principal invertor of the foot and its main role is dorsiflexion, perhaps any effect is too small to detect, or too variable depending upon the action of the other invertor muscles (i.e. posterior leg muscles passing medial to the ankle) and foot position. Also, as the only ankle dorsiflexor, TA function is unlikely to be compromised with more alternative invertor muscles available. Overall the majority of studies have found FOs do not change TA activity significantly, in some cases FOs may decrease TA activity, but any effect is subtle.

3.13.3 *Literature limitations*

Tibialis posterior is the largest invertor, but given fine-wire EMG can be challenging few studies have investigated its function, or intrinsic muscle activity. Furthermore, the magnitude of change in muscle activity that is clinically meaningful is unknown, thus significant effects of external devices on EMG does not reveal clinically beneficial effects. Electromyography is only a measure of electrical activity not force production, nor mechanical work in the muscle-tendon unit. Additionally, differences in electrode types, signal processing, normalisation and outcome variables make establishing a consensus regarding the meaning of changes in EMG difficult. Guidelines describe methods of EMG processing, but are not a universal best practice.[22]

Research investigating FOs used various materials and designs, and the descriptions of FOs were limited (no excluded study provided detail on this criteria). Studies used a mixture of customised and prefabricated FOs and both FOs with modifications only in the rearfoot and FOs with additional modifications in the arch and forefoot. Whether isolated modifications in specific FOs geometry could lead to specific changes in EMG is unclear. Additionally several studies may have also been inadequately powered. A final observation is that studies generally focus on the immediate effect of external devices, yet muscle function could change over time. Longitudinal EMG studies are difficult, but other approaches such as muscle morphology have proven sensitive to footwear.[68, 69]

3.13.4 *Review limitations*

While the quality assessment allowed the review to be based on studies of at least moderate quality, failing criteria could reflect inadequacy in reporting and not whether appropriate procedures were followed.[17] Additionally, database searches from one reviewer and the score of 50% as a threshold for inclusion could be considered subjective. Furthermore each criteria was given equal weighting when some could be more influential than others. For instance blinding might be unrealistic, as so-called sham FOs can exert mechanical effects[70] and potentially influence EMG. Conversely, lack of randomisation could invalidate results due to an order effect and be grounds for exclusion alone. Nonetheless, outcomes of the excluded studies were largely in agreement with those included, except one which found a significantly longer duration ($p < 0.05$) of TA activity following foot contact with FOs versus control.[71]

Footwear outside the inclusion criteria could alter loading of the foot and subsequent muscle activity. However a general review of footwear would be far broader and by restricting our search to footwear that aims to alter foot/ankle motion with modifications in sole construction, findings can be more directly related to the other devices reviewed. This review focused on muscle activation, however devices could have other effects on soft tissue, like the capacity of the series elastic element of TP to absorb energy in early stance.[58]

The review included studies with heterogeneous injury status and foot postures. The response to an intervention may vary with pathology. However the evidence that Achilles tendinopathy for example alters muscle activation is conflicting.[72, 73] Few of the studies included patient populations and those that did did not provide healthy controls and foot posture was often not reported. Consequently sub-group comparisons were not possible.

3.14 Conclusion

Modifications in shoe or FOs design in the sagittal or frontal plane can alter activation in walking of muscles acting primarily in these planes. Adequately powered research with kinematic and kinetic data is needed to explain the presence/absence of changes in muscle activation with external devices.

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3.16 Declaration of conflicting of interests

C.N. owns equity in a company that manufactures foot orthoses. Other Authors have no conflicts of interest to declare.

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3.18 Review appendix

Table 3-5. Excluded studies

Authors	Participant characteristics	Device	Muscles	Walking or running	Variables	Main findings	QA Score (%), First stage; second stage
Excluded at first stage							
Bourgit et al. (2008)[1]	12 healthy females: 24 ±4 years, 1.67 ±0.06 m, 60 ± 7kg	Standard fitness shoe (Reebok Revent Mid DMX, around 4° plantar flexion) and dorsiflexion shoes with a curvature plate in the midsole: Springboost B-Fit 2°dorsiflexion, Springboost B-Fit 4° dorsiflexion, and Meridian 10° dorsiflexion	BF, Glut Max, MG, LG, TA, RF, SOL, VL, VM	Treadmill walking and running	RMS as %MVC and integrated EMG (iEMG)	• Significant increase in plantar flexor RMS EMG (average of MG, LG and SOL) with dorsiflexion shoes vs standard shoe in walking and running	43; 54
Branthwaite et al. (2012)[2]	Males and females, n = 15 (age = 25.2 ±5.2 years; height = 1.68 ±0.07 m; weight = 66 ±10.6 kg)	Masai Barefoot Technology, participant's own shoes as control	SOL, MG, LG, TA, PL, RF, BF and Glut Med	Walking (10 m walkway)	Mean, maximum EMG (mV) and percentage values (tip toe exercise?)	• No significant difference between conditions. Large inter-individual differences in muscle activity	43; 54
Demura and Demura (2012)[3]	10 healthy males: 24.1 ±4.1 years, 1.72 ± 0.05 m, 69.2 ±8.4 kg	Stretch Walker (Nosaka, Ltd, Japan), Masai Barefoot Technology and flat-bottomed shoe	Glut Med, MG, LG, ST, SOL, TA, vastus tibialis?	Walking (laps of ~40 m walkway)	Integrated EMG	• No significant difference between conditions	33; 54

Gu et al. (2014)[4]	22 healthy males: 23.5 ±1.3 years, 1.73 ±0.01 m, 66 ±2.4 kg	Flat-soled control shoe, experimental shoe with adjustable hemispheres of 1.5 cm and diameter of 5.5 cm in the heel and forefoot (neutral, medial and lateral configuration)	BF, MG, LG, RF, TA, PL, VL, VM	Walking (10 m walkway)	RMS	• Increase in LG, PL and TA activity in experimental shoes vs. control	33; 54
Ritchie et al. (2011)[5]	21 males: 21 ±4 years, 1.768 ±0.05 m, 73.3 ±6.5 kg	Prefabricated 3/4 length Formthotic (Foot Science International)	MG, PL, TA	Walking (15 m walkway)	Activation ratio with respect to control	• No significant differences	43; 62
Tomaro and Burdett (1993)[6]	10 adults with history of leg pathology, treated with orthotics in last 6 months (3 M, 7 F, 25-30 years)	Prefabricated Sporthotic® (Langer Biomechanics, Deer Park, NY), own athletic shoes	LG, PL, TA	Treadmill walking	RMS amplitude, duration of activity	• Significantly ($p < 0.05$) longer duration of TA activity following heel strike with FO vs. control • No significant difference in EMG amplitude between conditions for all muscles	17; 46
Excluded at second stage							

Boergers et al. (2000)[7]	University track team athletes (n = 8; 5 F, 3 M, 20 ±1.1 years) with self-reported anterior shin pain.	Traditional arch tape Perrin (1995), experimental arch tape (over dorsal rather than plantar surface) and presumably barefoot	TA	Walking and running	Peaks and means	• No significant difference between conditions	71; 46
Burke et al. (2012)[8]	6 endurance-trained recreational runners (3 M, 3 F; 32.3 ±10.07 years)	Flexible, custom-made UltraStep orthotics, (Foot Levelers, Inc., Roanoke, VA)	BF, MG, TA, VL	Treadmill running	Peaks	• No significant difference between conditions	57; 31
Romkes et al. (2006)[9]	12 healthy adults: 6 M, 6 F, 38.6 ±13.2 years, 1.73 ±0.06 m, 77.4 ±12.3 kg	Masai Barefoot Technology and own shoe	MG, LG, RF, ST, TA, VL, VM	Walking	Amplitude as % of peak in barefoot	• TA activity decreased in early stance and increased in swing MBT vs. control ($p< 0.05$) • MG and LG increased activity from late swing to mid-stance MBT vs. control ($p< 0.05$) • VM and VL increased activity mid-stance to toe-off MBT vs. control ($p< 0.05$) • RF increased activity in mid-stance and decreased activity in early swing MBT vs. control ($p< 0.05$)	86; 38

Stöggl et al. (2010)[10]	12 healthy students: 6 M, 6F, 25 ±2 years, 1.72 ±0.07 m, 65 ±9 kg	Masai Barefoot Technology and running shoe (Adidas SuperNova. Herzogenaurach, Germany), before and after 10 week training intervention	BF, MG, TA, PL, VL, VM	Treadmill walking	Variability (SD over 15 consecutive gait cycles) for: RMS, integrated EMG (iEMG) and median power frequency	• No difference between shoes and no change by training in EMG variability	100; 44

*M= male, F= female, RMS= root mean square, TA= tibialis anterior, SOL= soleus, PL= peroneus longus, MG= medial gastrocnemius, LG= lateral gastrocnemius, VL= vastus lateralis, VM= vastus medialis, BF = biceps femoris, Glut Med= gluteus medius, RF= rectus femoris, FO= foot orthosis, QA= quality assessment, second stage ((number of satisfied criteria/number of applicable criteria)*100).*

3.19 Review references of excluded studies

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3.20 Outcome of literature review

The research question of this chapter was: what is the level of evidence that footwear, FO and taping alter lower limb muscle activity during walking and running? The review concluded that modifications in shoe or FOs design in the sagittal or frontal plane can alter activation in walking of muscles acting primarily in these planes. Also, that adequately powered research with kinematic and kinetic data is needed to explain the presence/absence of changes in muscle activation with external devices. The conclusion that FO can affect muscles acting in the frontal plane was based on only a few high quality studies on TP and interpreting the results of the effect of FO on PL was limited by previous reports of poor reliability of PL EMG. Additionally the review found a lack of research on the effect of FO on ABH EMG. Consequently in Chapter 5 is a reliability study of protocols for recording EMG from PL and ABH for subsequent study of FO effects. The review also provided rationale for the study in Chapter 6, which recorded indwelling EMG from TP in combination with surface EMG of ABH and PL, using the protocols from Chapter 5, surface EMG of MG and TA and kinematics and kinetics.

Chapter 4 METHODS

4.1 Introduction

This chapter outlines the main methods used in this thesis: electromyography, motion and force assessment and musculoskeletal ultrasound. The theory, protocol, processing and analysis of the data are presented for each method.

4.2 Electromyography

Electromyography is the technique of recording and analysing the electrical signal from contracting muscles (Muscle activation) and can be used in medical research, rehabilitation, ergonomics and sport science (Konrad, 2005).

4.2.1 *Sensors*

Electromyography can be recorded for superficial muscles using surface sensors on the skin (surface EMG). For muscles that are deep or small and narrowly bordering with other muscles it is necessary to use fine-wire EMG sensors, i.e. to insert an electrode directly into the muscle belly to obtain higher fidelity. Typically in the study of human movement a bipolar arrangement of electrodes/sensors is used (Konrad, 2005, Merletti and Hermens, 2004). In a bipolar arrangement a wave of depolarisation-repolarisation propagates along the muscle and a potential difference (micro volt) is measured between the two electrodes (Merletti and Hermens, 2004). The raw EMG signal has both positive and negative components because as the wave propagates the area of depolarisation will be closer to the first electrode at some points and closer to the second electrode at other points. The common signal at both recording electrodes is subtracted from the potential difference between the two and a ground/reference electrode, which is traditionally placed on a bony landmark and can be used to remove the noise from external sources (Hermens et al., 2000). The higher the common mode, the better the rejection ratio (Konrad, 2005). With active sensors pre-amplifiers near the recording site can be used to apply a 500 fold gain to the signal, which reduces the sensitivity of the signal to cable artefacts (Konrad, 2005).

4.2.1.1 *Surface EMG*

Guidelines for EMG data collection were produced in a European collaborative project: The Surface EMG for a Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000).

The SENIAM website has recommendations on the optimal location of sensors on specific muscles and general guidelines on collecting surface EMG relating to:

1. Selection of the surface EMG sensor;

2. Preparation of the skin;

3. Positioning the patient in a starting posture;

4. Determination of the sensor location;

5. Placement and fixation of the sensor;

6. Testing of the connection.

The SENIAM guidelines recommended an inter-electrode distance of 20 mm or ≤ ¼ of muscle length for smaller muscles. However De Luca and colleagues (2012) demonstrated that there could be significant crosstalk (activity from other muscles) during walking when recording from the tibialis anterior and instead advocated a 10 mm inter-electrode distance with a double differential bar configuration, as used in Delsys sensors. A sensor should be placed parallel to muscle fibres over the muscle belly, i.e. in the middle of the muscle not near the innervation zone or the tendon where the signal changes shape (Merletti and Hermens, 2004). Skin preparation is recommended to optimise adhesion of the sensors to the skin and to improve the EMG signal quality by reducing skin impedance (Hermens et al., 2000, Konrad, 2005). Preparing the skin involves shaving hair off the skin at the site, if present, and removing oils and dead skin cells with an alcohol wipe and an abrasive gel, although the latter step in particular is not necessary with Delsys electrodes (Delsys Inc., 2019). The use of a conductive gel on the skin will also improve skin impedance, although many electrodes come pre-gelled and additional gel may induce artefacts when sweating and/or mechanical disruption to the sensors are likely (Roy et al., 2007).

Recommendations for the overall reporting of EMG procedures have been outlined by Merletti (1999), which have been endorsed by the International Society of Electrophysiology and Kinesiology (ISEK). The details included in the recommendations such as specific properties of the sensors used as well as processing and normalisation methods could impact the EMG output, which was why they were incorporated into the quality assessment in the literature review in this thesis.

4.2.1.2 *Indwelling EMG*

For the study of movement fine-wire electrodes are typically used for recording from deep muscles (Konrad, 2005). Recording indwelling EMG is invasive and challenging in comparison with surface EMG (O'Connor et al., 2006, Semple et al., 2009, Stacoff et al., 2007). A smaller number of motor units are detected with fine-wire electrodes compared to surface electrodes and so the placement of fine-wire electrodes is more sensitive to position with respect to active motor units than surface electrodes (Kadaba et al., 1985). Fine-wire electrodes consist of insulated wires, with uninsulated recording tips, that are inserted into the muscle with a hypodermic needle (Konrad, 2005). The needle is removed after insertion and the protruding wires are attached to a sensor on the skin surface. Fine-wire electrodes can be purchased pre-made, for instance: a 50 mm long and 25 gauge (Chalgren Enterprises Inc., USA) was used in this study. These electrodes have a 2 mm uninsulated recording tip and have previously been used in the literature (Barn et al., 2014, Kelly et al., 2016). Other research groups manufacture their own fine-wire electrodes, for example electrodes with 1 mm of uninsulated recording surface (Murley et al., 2014b, Stacoff et al., 2007). No literature was identified comparing the consistency of the recording capabilities of electrodes with recording surfaces of different length. Therefore, it is unclear whether the EMG signals would be significantly affected by the length of the uninsulated recording surface.

The amplitude and frequency of indwelling EMG is larger than surface EMG because skin has low-pass filtering properties (Kamen and Gabriel, 2009). The necessary sampling frequency is thus higher for indwelling EMG than surface EMG as the sampling frequency should be at least twice the expected maximum frequency of the signal according to the Nyquist theorem (Konrad, 2005). Surface EMG is often sampled at 1000 Hz, while indwelling EMG is typically sampled at a minimum of 1500-2000 Hz (Barn et al., 2012, Burden, 2007, Konrad, 2005). However, all EMG data (including surface EMG) were collected at 2000 Hz in this thesis as is the Delsys EMG sensor specifications.

4.2.1.3 *Delsys EMG sensors*

The surface EMG data in this study were collected using sensors manufactured by Delsys Inc. The Trigno™ standard sensors are active, wireless sensors that are reusable and include a triaxial accelerometer. The sensor dimensions are 27 mm x 37 mm x 15 mm. The sensors have a patented design consisting of four 1 mm x 5 mm parallel bars (contacts). The contacts are 99.9% silver and have a fixed inter-electrode spacing of 10 mm. The electrodes record two EMG inputs, one from the different pair and another from two stabilizing references. This method of reducing the noise in the signal eliminates the need for aground electrode. The bandwidth of the EMG signal was 20-450 Hz.

The Delsys Trigno™ Mini sensors (**Figure 4-1**.) have a sensing head with dimensions 25 mm x 12 mm x 7 mm, which makes them suitable for recording from small muscles and muscles that are difficult to isolate (Delsys Inc., 2014). A 200 mm long cable connects the sensing head to the main sensor, which serves as a stabilizing reference and has the same dimensions as the standard sensors.

Figure 4-1. Delsys Trigno™ Mini sensors to scale

The skin preparation guidelines that Delsys recommend were followed, which were to shave excessive hair from the detection site, wipe the skin with isopropyl alcohol, then allow the skin to dry for a few seconds before application. Delsys Adhesive Sensor Interface (double sided sticker) was used to affix the sensors. The particular Delsys adhesive has slits in it that can be directly aligned with the electrode contacts and has been developed to have high peel adhesion to maintain contact with the skin during mechanical perturbations (Roy et al., 2007).

4.2.2 *Development of fine-wire EMG technique for tibialis posterior*

Using fine-wire electrodes is a challenging and invasive method to record EMG, which unlike surface EMG does not have established guidelines. As such during the development of the protocol the technique for insertion evolved after seeking advice from researchers who are experienced in this procedure.

4.2.2.1 *Electrodes*

Insertion of the bipolar fine-wire electrodes (0.051 mm diameter, paired-hook wires, Teflon-coated stainless-steel wire) is performed with an unused, sterile, hypodermic needle (50 mm long and 25 gauge, Chalgren Enterprises Inc., USA).

4.2.2.2 *Anatomical approaches*

Needle insertions into the tibialis posterior (TP) can be performed using either the *anterior* approach or the *posterior* approach, sometimes referred to as the *posterior-medial* approach (Rha et al., 2014, Semple et al., 2009, Yang et al., 2008). In the anterior approach the needle is inserted in the mid or upper third of the length of the tibia, halfway between the tibia and the fibula, passing through the tibialis anterior, or the extensor digitorum longus and then the interosseous membrane, before reaching TP (Rha et al., 2010, Rha et al., 2014, Yang et al., 2008). With the anterior approach care must be taken to avoid the tibia and the deep anterior neurovascular bundle (Rha et al., 2014, Semple et al., 2009, Yang et al., 2008). In the posterior approach, the insertion is made at 50% of the distance between the medial malleolus and the tibial tuberosity, posterior to the medial border of the tibia and angled towards the fibula (Murley et al., 2009a, Semple et al., 2009). In the posterior approach care must be taken to avoid the tibia bone, the saphenous nerve and the posterior neurovascular bundle (Rha et al., 2014, Semple et al., 2009, Won et al., 2011, Yang et al., 2008).

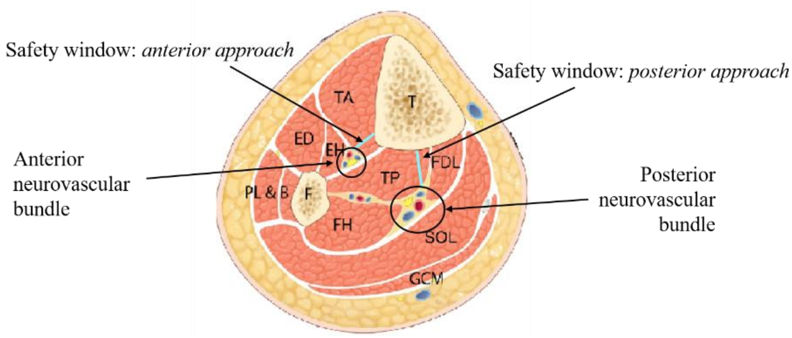


Figure 4-2. Cross-sectional drawing of the calf indicating the safety windows (blue lines) of the anterior and posterior approaches. Adapted from Rha et al. (2014), ED indicates extensor digitorum; EH, extensor hallucis; F, fibula; FDL, flexor digitorum longus; FH, flexor hallucis; GCM, gastrocnemius; PL &B, peroneus longus and brevis; SOL, soleus; T, tibia; TA, tibialis anterior; TP, tibialis posterior.

The choice of either the anterior or posterior approach for insertion into TP depends on safety and practicality. It was thought that the safety window, the region in which the needle could be inserted while avoiding bone and neurovascular bundles, was greater in the anterior approach than then posterior approach (**Figure 4-2**.) (Lee et al., 1990). However the relative safety of the two approaches appears to be dependent on the longitudinal location of insertion, with the anterior approach being safest in the upper third of the length of the tibia and the posterior approach being safest at the midpoint of the tibia (Rha et al., 2010, Rha et al., 2014, Won et al., 2011). In adults the safety window at the optimal location using either approach is approximately 1-1.5 cm (Rha et al., 2010, Won et al., 2011). The disadvantage of the anterior approach is that the EMG signal during walking can be unstable with this technique, often because the electrode tips retract through the interosseous membrane into the tibialis anterior (Semple et al., 2009). Therefore while the anterior approach may remain an option for clinical applications of needle insertions into tibialis posterior, such as diagnoses of neuropathies with EMG and botulinum toxin injection in cerebral palsy patients (Rha et al., 2010, Rha et al., 2014, Won et al., 2011, Yang et al., 2008), the posterior approach is advocated for EMG studies of locomotion.

4.2.2.3 *First technique*

The participant is either sitting or lying supine on a physio plinth with both legs initially extended. Testing for this study was performed on the right leg because it is most often the dominant leg and there was no reason to believe the effect of the FO would differ between legs in healthy participants. The right leg is slightly flexed and everted with a towel roll to prop the knee to keep the leg in position. The tibial tuberosity and medial malleolus are palpated and marked with a marker pen or eyeliner pencil and the distance between these marks measured with a tape measure. The point 50% of the distance between these marks is then located and an additional mark made level to this, slightly more posterior, to avoid the insertion site itself.

In the first technique water-soluble gel applied to an ultrasound probe and the site mid-shank is scanned in the transverse plane with the probe perpendicular to the skin and then the area is cleaned of gel prior to insertion (**Figure 4-3**). From the ultrasound image (B-mode) the intention is to firstly identify TP itself and ascertain its depth. The second purpose of the ultrasound scan is to locate the neurovascular bundle and border of the tibia (**Figure 4-4**). A thin metal pointer probe under the ultrasound probe can be used to gently compress the skin over the site of the neurovascular bundle and border of the tibia bone. The ultrasound probe can then be removed, with the pointer probe still in place allowing the sites to be marked, either directly with a surgical pen that writes through gel or slightly proximal or distal to said locations.

Figure 4-3. Ultrasound probe in the position of scanning midshank in the transverse plane (Onmanee, 2016)

The quality of the ultrasound image varies considerably between individuals and as such the ease with which the neurovascular bundle can be located differs. A broadband transducer (e.g. 6-15 MHz) is advantageous in cases like this, when superficial and deep structures are of interest because the spectrum of frequency distribution means it is possible to have different frequencies within a single scanning plane (Wave theory) (Bianchi and Martinoli, 2007). The frequency of the ultrasound was a compromise between a low frequency being preferable for greater depths (i.e. locating TP) and greater axial resolution at higher frequencies for ease of imaging the neurovascular bundle (Schmidt, 2016). It was useful to ask the participant to perform inversion and then toe flexion to visualise the contraction of TP and flexor digitorum longus (FDL) respectively. The ease with observing isolated movements varies between individuals, as sometimes each muscle can be seen to contract in isolation, while sometimes the whole area appears to move with one or other movement, possibly due to an inability to isolate the movement or transfer of pressure between the muscles. Doppler ultrasound imaging can be useful in locating the neurovascular bundle, with a bright blue/red spot indicating blood flow at the bundle location (**Figure 4-5**). In Doppler ultrasound the wavelength/frequency of the sound wave changes when it is reflected by an object in motion (e.g. a red blood cell) and the difference between the emitted and received frequencies are translated into colour information and overlaid on the grayscale image (Hammer and Terslev, 2016). The extent of vascularisation in the muscles also appears to vary, as in some cases the location of the bundle is not clear and the whole area flashes blue/red.

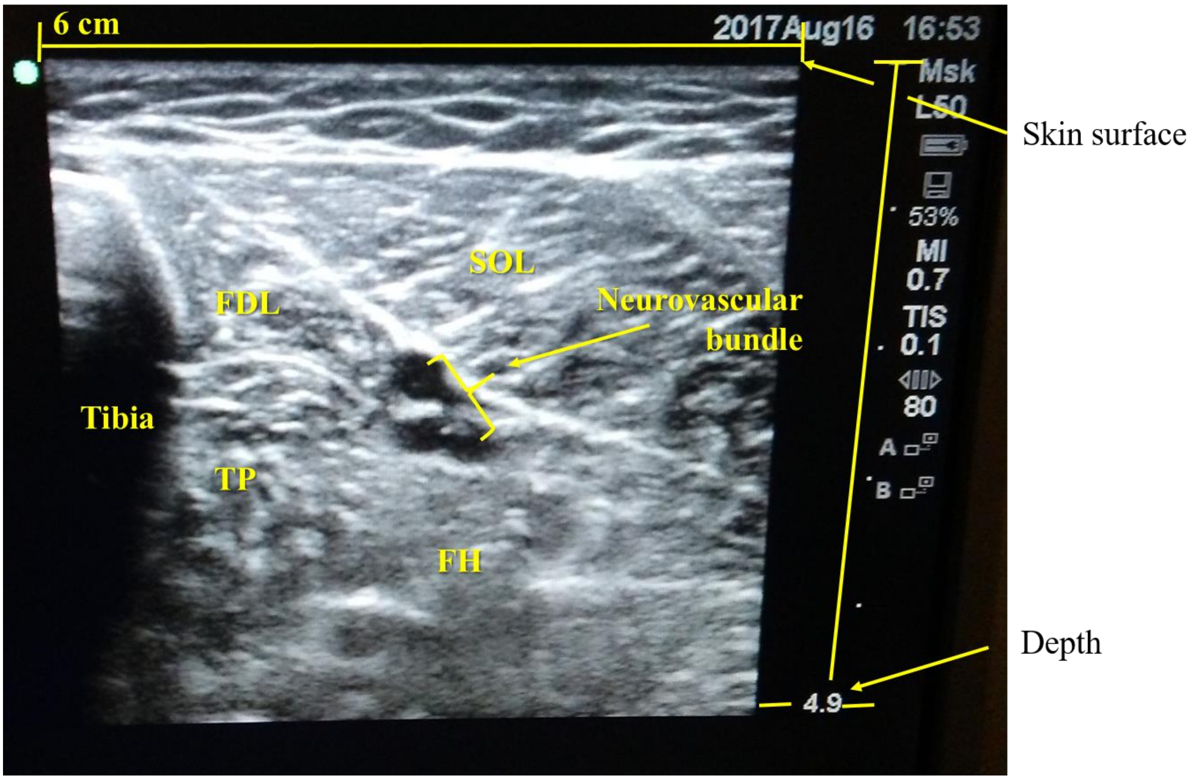


Figure 4-4. Ultrasound scan of right leg at midpoint of calf using the posterior approach. FDL= flexor digitorum longus, SOL= soleus, TP= tibialis posterior, FH= flexor hallucis.

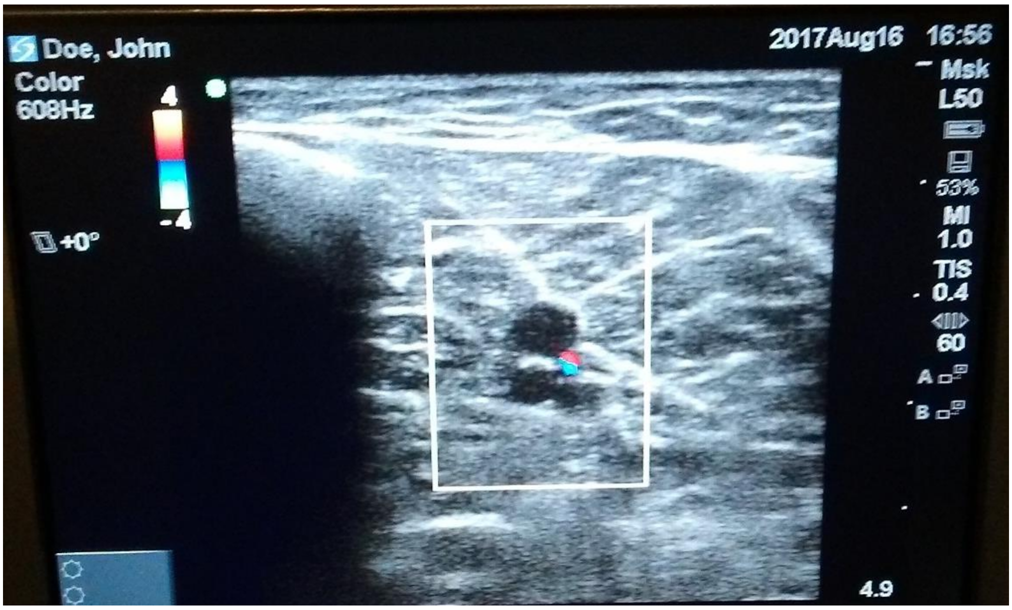


Figure 4-5. Doppler image of right calf showing blood flow in the neurovascular bundle (note Doppler imaging is more useful when the neuromuscular bundle is less clearly visible using B-mode ultrasound)

Once the insertion site is wiped of gel and cleaned with an alcohol wipe, the needle can be removed from its packaging with sterile gloves on. Care must be taken to ensure that the uninsulated tips of the needle are pointing in opposite directions before insertion. The needle is inserted in the direction of the fibula. The depth of the insertion is determined from the ultrasound image, with additional depth required to allow for the electrode tips, which are around 0.5 cm, pointing back towards the surface of the skin. In practice, the depth required is commonly all or most of the length of a 5 cm needle for a participant of average height. The needle is then removed and disposed of in a sharps bin and the electrodes are left in place. The participant is then asked to make several inversion movements to encourage the electrode to imbed into the muscle.

Electrical stimulation (Dantec Clavis, Natus Neurology Inc., USA) is used to ascertain whether the electrodes are correctly inserted into TP (**Figure 4-6**). A ground (contact) is placed on a patch of skin cleaned with an alcohol wipe distal to the insertion site. Each electrode (wire) is tested in turn. If the insertion has been successful gentle stimulation will result in slight inversion of the foot only. When the insertion is unsuccessful the most common location of the electrodes is in the FDL and so slight movement of the lesser toes will be seen instead of (or in addition to) inversion. If the electrodes are in the FDL they are removed and the insertion is repeated with a new needle (Murley et al., 2009a). If the insertion is successful a loop is taped in the wire electrodes to allow for skin movement (Onmanee, 2016) and the electrode tips are then connected to a spring contact sensor (**Figure 4-7**).

Figure 4-6. Stimulation of electrodes

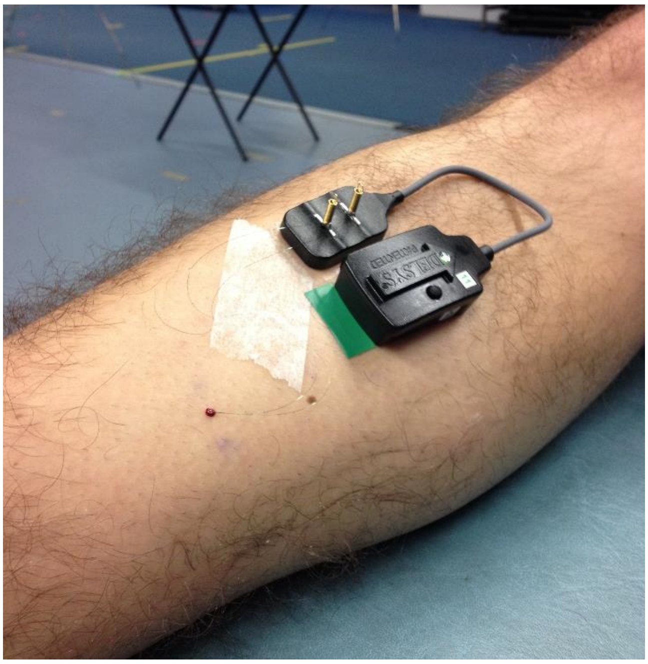


Figure 4-7. Fine-wire electrodes with a loop taped in the wires and the tips attached to a spring contact sensor

The first technique was challenging and met with limited success as frequently the electrodes proved to be in the FDL and not TP. The difficulty was not purely a question of depth (FDL being more superficial than TP), but also of the angle of insertion, as the needle could be directed across the muscle rather than directly through it. Advice was sought from Dr. Ruth

Barn (née Semple) who has published work using fine-wire EMG of TP (Barn et al., 2012, Barn et al., 2014, Semple et al., 2009).

4.2.2.4 *Second technique*

The second technique followed much the same procedures as the first technique with the exception of using a sterile probe cover and sterile gel to perform insertion of the needle while the ultrasound probe is still in contact with the skin. In this way it is possible to visualise the needle passing through the tissue until it reaches TP. This was the method used by Murley and co-workers (2009a, 2009c) and depicted in a video accessible at:

https://www.biomedcentral.com/content/supplementary/1757-1146-2-35-S1.m4v (**Figure 4-8**).

In practice it was found to be necessary to have ultrasound gel between the probe cover and probe.

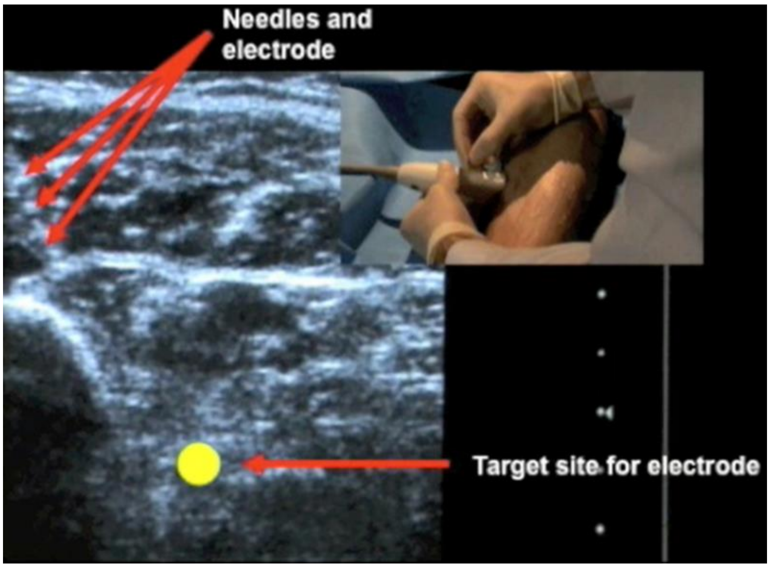


Figure 4-8. Still frame from the above video depicting the progression of the needle through the tissue during insertion with simultaneous ultrasound guidance

Theoretically the second technique is advantageous in that the progression of the needle can continue until TP is reached and conversely it can be removed early if the trajectory is clearly incorrect, without the need of stimulation and reducing the risk of hitting the neurovascular bundle. However in practice it was found that the ease of which the needle could be seen on the ultrasound image passing through the tissue was again highly individual. Sometimes a clear line could be seen on the ultrasound image as in **Figure 4-8**, otherwise only a varying extent of tissue deformation around the needle as it progressed could be observed.

Needle insertion with simultaneous ultrasound guidance did not eliminate the challenge of inserting at the correct angle, in fact it presented with an additional challenge of compromising keeping both the ultrasound probe and needle direction as perpendicular to the skin as possible at the same time, while getting the needle under the probe. It was necessary to ask an assistant to hold the ultrasound probe after the correct location was found, but there was still often some slipping of the probe during the time it took to prepare the needle and ensure the electrode tips were pointing in opposite directions. As a result, the success rate using the second technique did not greatly improve on the first technique.

4.2.2.5 *Adapted technique*

Further training was gained from Prof. Juan Garbalosa, who has also published literature using fine-wire EMG of TP (Garbalosa et al., 2015), which aided the technique. Prof Garbalosa does not use ultrasound guidance, but inserts the needle approximately a thumb’s width from the border of the tibia at the midpoint between the medial malleolus and tibial tuberosity (Garbalosa, personal communication 21st July, 2017). However, for greatest confidence in insertion location ultrasound scanning of the insertion site and marking the safety window prior to insertion was deemed to be the most applicable (as per the first technique).

However, in the adapted technique, the position of the participant’s leg was adapted so that it is flexed more than the other techniques and raised off the bed, supported by the knee of the person inserting the needle (**Figure 4-9a**). In this way the needle casing can be placed on the lateral aspect of the leg opposite to the needle insertion site and so it is easier to visualise a slice through the leg and the correct angle of insertion (**Figure 4-9b**). This modification of the technique appears to offer the best success rate and was thus used for subsequent data collection.

Figure 4-9. Preferred position for fine-wire insertion showing a) the shank is raised off the bed and b) the needle casing on the lateral side to visualise a line through the shank

4.2.3 EMG processing and analysis

Processing of the EMG signal included rectification, filtering and averaging. The profile of average muscle activation could be calculated using a moving average, a root mean squared (RMS) calculation or as a linear envelope with a low pass filter (e.g. second order Butterworth filter at 6 Hz) (Konrad, 2005). The RMS has been advocated over a moving average due to the RMS having lesser variability and greater meaning as a quantification of signal power rather than simply the area under the curve (Burden, 2007). The RMS calculates the mean power of the raw EMG over a specific epoch (Burden, 2007, Konrad, 2005). The window size over which the calculation is made in the RMS is important as it affects the level of smoothing of the signal, with between 50-100 ms being common in gait studies (Burden, 2007, Konrad, 2005). A 75 ms window was used to calculate the RMS per trial and averaged from all included gait cycles. Window size was chosen after piloting with commonly used window sizes between 50-100 ms, which are considered to perform well during most movement studies (Konrad, 2005).

For each condition and muscle RMS signals were exported to ASCII files normalised to the gait cycle for further analysis in a MATLAB (R2016b) script. In the MATLAB script the EMG profiles over each individual gait cycle were plotted so that gait cycles that were not representative of the overall pattern of activation could be removed. A minimum of six good gait cycles were sought for further analysis.

4.2.3.1 *Normalisation*

Normalisation of the EMG allows the EMG signal to be expressed as a percentage of a set value, often an amplitude obtained from a maximum voluntary isometric contraction (MVC). Normalisation was necessary to calculate an average across a sample, as factors like subcutaneous tissue would vary significantly between participants (Staudenmann et al., 2010). As skin impedance would also vary day by day, the normalisation allowed the comparisons could be made within individuals in repeated measurement studies.

Previous work within the group found that EMG normalisation to the mean or the peak was superior to normalising to MVCs in muscles of the shank with regards reducing variability (Onmanee, 2016). Thus, in the MATLAB script the EMG signal amplitude was normalized to the peak per gait cycle. Peak normalisation was easier to interpret than the mean normalisation and was used for normalising all EMG in this thesis for consistency. Peak normalisation is also appropriate for within-subject comparisons across multiple conditions.

4.3 Motion capture

4.3.1 *Camera system*

Both motion and ground reaction force data were recorded with a 15-Camera wall mounted infra-red cameras (Oqus 400, Qualysis, Gothenburg, Sweden) operated with Qualysis Track Manager Software (QTM, version 2.13, Qualisys AB, Sweden) and captured kinematic data at 100 Hz. The four synchronised force plates (BP400600, AMTI, USA) and a 16-bit analog-to-digital convertor, i.e. the USB-2533 Analog board (Qualisys AB, Sweden) was used to collect the ground reaction forces at a sampling rate of 1000 Hz. Analog and digital data were synced using a trigger in which Delsys Trigno was the master. EMG data were collected with the synchronised Delsys system at a sampling rate of 2000 Hz.

The motion capture system was calibrated prior to data collection. The manufacturer's "L-shaped" calibration frame with four reflective markers was positioned in the corner of the designated force plate to set the origin and define the axes of the lab (global) co-ordinate system. During the calibration process a "T-shaped" wand with a known distance of (601.7 mm) between its markers was waved through the calibration volume. A 70 second period of calibration allowed for at least a minute of calibration with time to move into the calibration volume after pushing the trigger to start the calibration from the computer. The volume was set up as 6 meters long, 2 meters wide and 2 meters high volume, which covered a walking distance with a few steps before and after the force plates and the wand was moved evenly through the volume. The calibration was accepted if the residual error from each camera was below 0.8 mm, if not the process was repeated.

4.3.2 *Marker setup*

Reflective markers were placed on anatomical landmarks to track three-dimensional movement of the shank and feet using a six degree of freedom model. In a six degree of freedom model body segments are tracked independently about unfixed joint centres with three translations and three rotations permitted about each joint (Baker, 2013).

Individual markers (9 mm diameter) were placed bilaterally on the medial and lateral femoral epicondyles (the knee), the medial and lateral malleoli and on the 1st metatarsal head (MTP1) and the 5th metatarsal head (MTP5) of the left foot for anatomical calibration (**Figure 4-10a, b and Figure 4-11**). Markers were placed on the right and left leg to enable proper functioning of the automatic gait event detection pipeline command (see Definition of gait events). A small rigid cluster of four markers was attached to the shank to track the shank segment in dynamic trials. A bandage was fixed over the shank cluster in the reliability study of Chapter 5, but not in the immediate effects of FO study in Chapter 6 because the fine-wire electrode wires and sensors were in the way. All markers remained on the participant during the walking trials.

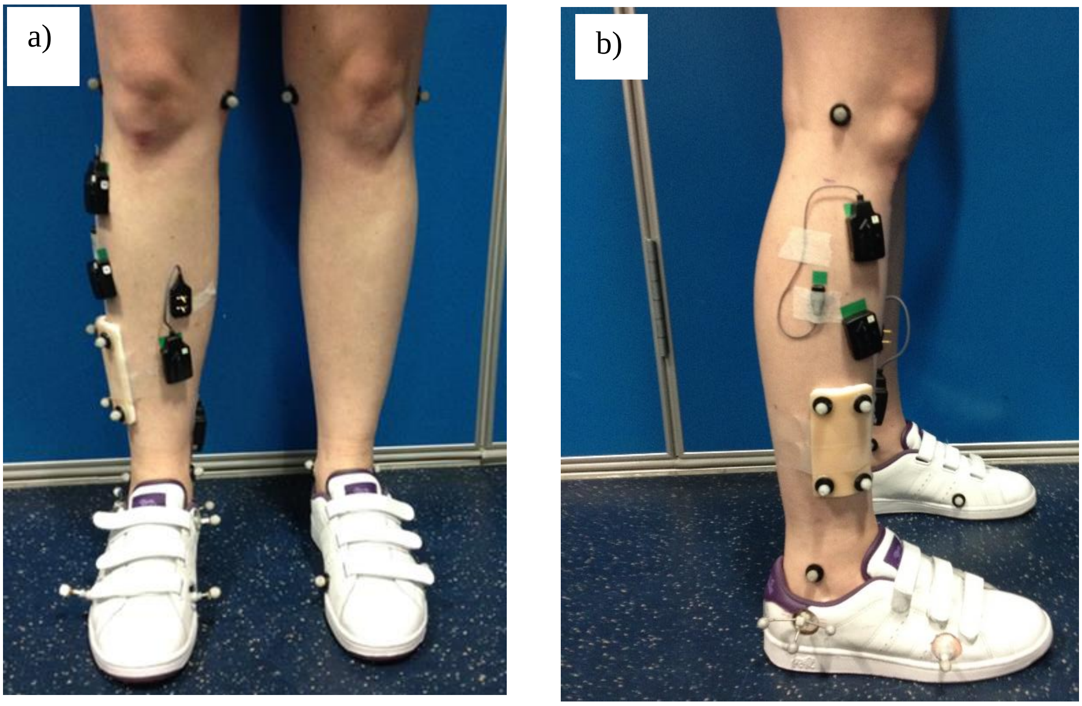


Figure 4-10. Experimental marker set up showing EMG electrodes and reflective markers bilaterally on: medial and lateral femoral epicondyles, medial and lateral malleoli, 1st metatarsal head, 5th metatarsal head and on the right leg: a shank cluster, the medial aspect of the calcaneus, navicular tuberosity and a triad wand cluster on the lateral heel. The 1st metatarsal head and 5th metatarsal head on the right leg, the medial aspect of the calcaneus and the navicular tuberosity markers were on a wand.

Figure 4-11.Marker set up showing the wand markers on the 1st metatarsal head, navicular tuberosity and the medial aspect of the calcaneus for medial longitudinal arch angle calculation

Figure 4-12 Triad marker cluster on the calcaneus (from another marker set up)

The reflective markers on wands were attached to the mounting plates through the holes of the right shoe at MTP1, MTP5, the navicular tuberosity and medial aspect of the calcaneus. The Lonsdale Leyton trainers were deliberately chosen because 1) the VELCRO® design enabled a quick change between conditions and 2) the firm heel cup meant that enough of the trainer's integrity could be maintained after cutting a hole in it. Holes in the shoes allowed tracking the movement of the foot itself rather than the shoe. The mounting plates remained on the skin in between trials while the wands could be unscrewed to take the shoe on and off. Tape was placed over the top of the mounting plates to help them stay on the skin while removing the shoe. A triad wand cluster was attached through a 25 mm hole at the lateral aspect of the calcaneus to allow tracking of the calcaneus (**Figure 4-12**).

In an investigation on the effect of shoe-hole size diameter on the collision of marker wands with a shoe, it was found that a 25 mm diameter was sufficiently large to prevent marker movement at all sites (Bishop et al., 2015).

Static trials are used in motion capture to establish a neutral standing reference position in quiet standing from which joint angles are calculated from and joint centres can be defined (Milner, 2007). A static trial was collected before each condition in which participants were asked to stay still with their arms either folded or out to the side for a few seconds. The static trial from the shoes with manufacturer's inlay (reliability study) or flat inlay (immediate effects study) was used to calibrate the anatomical reference position for the dynamic trials except for the barefoot condition, for which a separate static was used.

4.3.3 *Force data*

Ground reaction force (GRF) data were recorded using two AMTI force plates (Type: BP400600, dimensions: 600mm x 400mm) at 1000 Hz. The orientation of the force plates in the lab is end to end which is intended to maximise the likelihood of a successful contact of the right foot on one of the plates during dynamic trials. Force plates were reset before each new condition to reduce the potential for signal drift which is an issue with piezoelectric force plates.

4.3.4 *Procedure*

Before the start of the first condition, each subject was allowed to take a few walking practice trials to establish an appropriate starting position to further encourage single foot contact on a force plate. The starting position was sufficiently far from the plates to permit an even and comfortable gait. Participants were encouraged to walk naturally without focusing on achieving a clean contact with a force plate.

4.3.5 *Controlling for speed*

Walking speed was controlled in processing using the mean stride time in each session from the barefoot and shoe only conditions for each participant. Stride time was measured from right heel strike to right heel strike excluding the interval in which the participants changed direction of walking. Strides ±5% of the mean in all conditions were excluded from analysis.

4.3.6 *Data processing and analysis*

Dynamic trials were cropped to allow a few seconds before the right foot step prior to contact with the force plate and after at least one right step after force plate contact at the end of the trial. Static trials were cropped to a region with minimal body sway and good visibility of all markers for a minimum of 10 frames. Trials with only partial foot contact with the force plate or in which it appeared the participant had deliberately targeted the force plate were excluded from the analysis. Markers were identified and labelled in QTM version 2.13. An Automatic Identification of Markers model (AIM model) was generated for each participant. The AIM model used the pattern of angles and distances between markers in a movement trial to automatically label markers in subsequent trials. Data were added to each existing model from a couple of trials after model creation to improve the model. Labelling in trials in which the model was applied was visually checked and manually edited where necessary. Files were exported as C3D files without a zero force baseline correction.

4.3.7 *Model building*

The C3D files were imported into Visual 3D (version 6, C-Motion, Inc., Germantown, USA) for analysis. In the Visual3D workspace a model was created by adding the static calibration file, which was used as the anatomical reference point from which motion was calculated relative to. Dynamic files were assigned to the static file and a kinematic model was built. A model template was created, which was applied to subsequent participants. Mass and height was adjusted in the model in "Subject Data/Metrics" for each participant. The model consisted of the following segments: lab, left shank, left foot, left virtual foot, right shank (**Figure 4-14**), right foot (**Figure 4-13**), right virtual foot and right heel (**Figure 4-15**). The local co-ordinate system for each segment consisted of three orthogonal axes, with a medio-lateral x-axis, an anterio-posterior y-axis and a vertical z-axis around which sagittal, frontal and transverse motion were calculated respectively. The right shank was defined using the anatomical markers: the medial and lateral epicondyles of the knee and the medial and lateral malleoli.

The midpoint between the epicondyles and midpoint between the malleoli defined the z-axis. The foot segment was defined using the medial and lateral malleoli and the first and fifth metatarsals. The virtual foot for the right and left were created using virtual landmarks projected to the floor from the malleoli and metatarsal markers (**Figure 4-15**). Using a virtual foot enables the reference position of the foot to be neutral as supposed to in plantar flexion as it is with the foot defined using the malleoli.

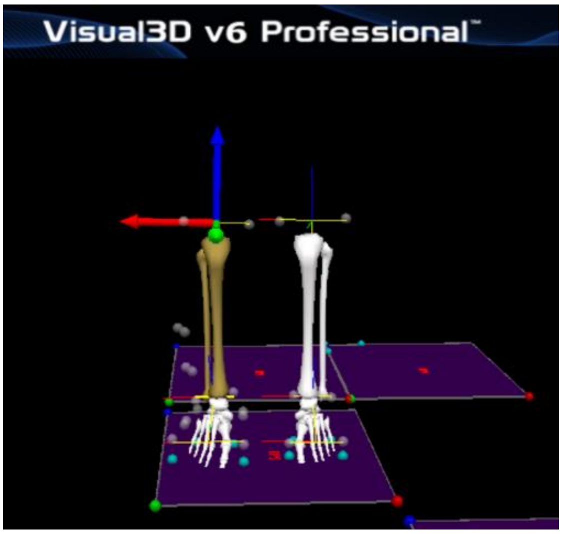


Figure 4-14. Right shank segment with co-ordinate system

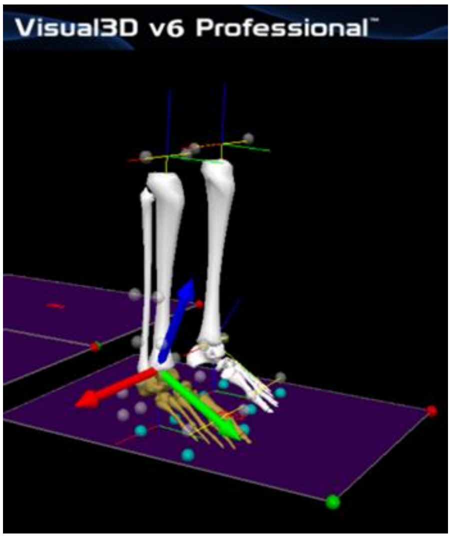


Figure 4-13. Right foot segment with co-ordinate system

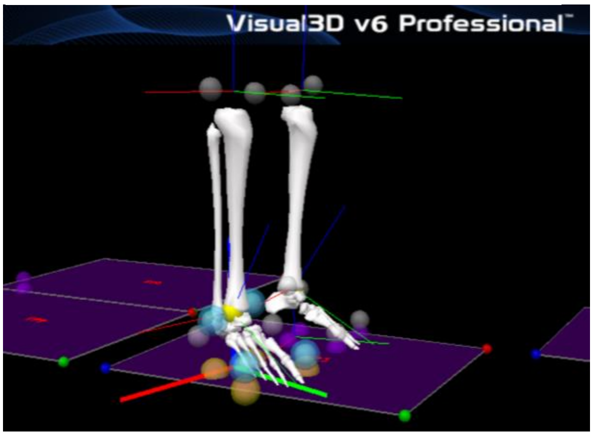


Figure 4-15. Kinematic model with the offset virtual markers (highlighted brown) and the co-ordinate system of the virtual foot and right heel

A pipeline of processing steps was created which included the following steps:

4.3.8 *Filtering and interpolation*

Filtered trajectories were interpolated with a third order polynomial with a maximum gap size of 10. The intention of using a low pass filter is to remove noise associated with skin movement artefact (Milner, 2007). Marker trajectories were low pass filtered using a bi-directional Butterworth filter with a 6 Hz frequency cut-off, which is typical for walking data (Milner, 2007).

4.3.9 *Definition of gait events*

The "Automatic_Gait_Events" command pipeline was used to define gait events. The command uses pattern recognition of marker trajectories to label heel strike and toe-off events that correspond to ON and OFF events on the force plate (Stanhope et al., 1990). The ON/OFF events are based on force plate assignments, when force rises above a threshold (10 N) which is set as the minimum force platform value. (C-Motion, 2015).

4.3.10 *Construction of angular joint data*

Ankle inversion/eversion angle, plantar flexion angle and the rearfoot inversion/eversion angle were calculated in Visual3D using the Compute_Model_Based_Data pipeline command. The right shank was tracked using three markers on the right shank cluster in the study in Chapter 5. In the study in Chapter 6 the position of the fine-wire sensor meant it was not possible to bandage over a shank cluster, so the anatomical markers were used for tracking. The right virtual foot (Model building) was used to calculate ankle dorsiflexion and plantar flexion angle and rearfoot inversion/eversion angle based on tracking markers on both shank and foot (using the malleolus and metatarsal markers projected to the floor). The ankle angle and rearfoot angle were presented with the right shank as the reference segment and an X-Y-Z Cardan sequence of rotations.

4.3.11 *Medial longitudinal arch angle*

The medial longitudinal arch (MLA) angle was calculated as the angle in the sagittal plane between a vector from a marker on the medial aspect of the calcaneus to the navicular tuberosity and a vector from a marker on the navicular tuberosity to a marker on the first metatarsal head (Balsdon et al., 2016) (**Figure 4-16b**) using the Compute_Planar_Angle command pipeline (**Figure 4-16a**).

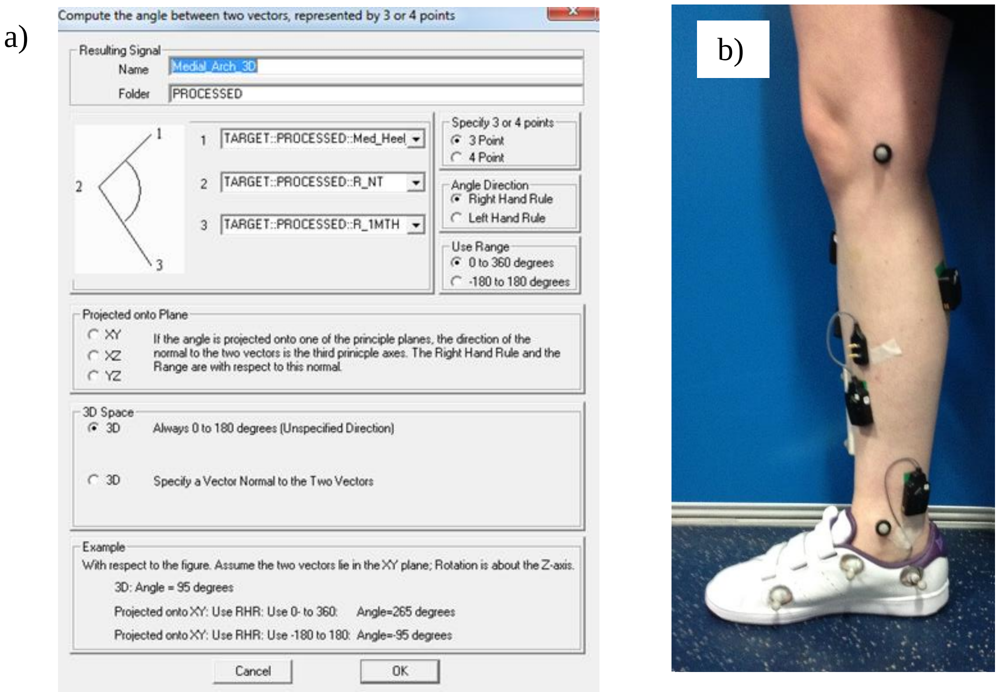


Figure 4-16. Calculation of the medial longitudinal arch angle a) Pipeline in Visual3D and b) marker setup showing markers on the first metatarsal head, the navicular tuberosity and medial aspect of the calcaneus

4.4 Kinetics: joint moments

Inverse dynamics uses the recorded motion about a joint, ground reaction force data and the moment of inertia of the segment (from estimated segment mass, proximal and distal radii and segment geometry) together with ground reaction force data to calculate the net joint moment (torque) required to produce the joint movement (Yeadon and King, 2007). The ankle inversion/eversion moment and ankle dorsi-/plantar flexion moment were calculated using inverse dynamics in Visual3D with the inbuilt Compute_Model_Based_Data pipeline command. For each segment Visual3D calculates the mass, moments of inertia and centre of gravity location using a classic model (Hanavan Jr, 1964). The calculations use participants' body weight and the regression equations and anthropometrics of Dempster (1955).

As the segment proximal to the ankle joint, the right shank was used as the reference coordinate system. The net internal moment calculated using Visual3D followed the right-hand rule by default (C-Motion, 2017). As the effect of FO on the external ankle joint moment was of primary interest, the default internal moment values were negated to calculate the external moment, as has been reported previously (Nester et al., 2003b). Ankle inversion/eversion moment represented an external inversion when positive and an external eversion moment when negative. Dorsi flexion moment represented dorsiflexion when positive and plantar flexion when negative. Moments were normalised to each individual's body mass and data were presented time-normalised to 100% of stance, as moment calculated using GRF, i.e. when the foot is contact with the ground in stance and not swing. Multi-segment joint moments within foot were not calculated as such computation are very complex requiring joint definition and plantar pressure data at individual joints within the foot (Baker and Robb, 2006) or consecutive force platforms and targetted stepping (Bruening et al., 2012, Bruening et al., 2010). As data was obtained from a single force platform, the GRF represents the force from a combination of the rearfoot and forefoot segments, thus for ankle moment calculations the foot was represented as a single rigid segment (Hsu et al., 2014).The whole foot rather than the rearfoot segment was thus used for ankle moment calculations.

4.5 Ultrasound

Ultrasound waves can be used to image musculoskeletal structures such as those in the foot and ankle (Angin et al., 2014, Angin et al., 2018, Mickle et al., 2013, Murley et al., 2014a). Musculoskeletal ultrasound was used in in Chapter 5 to aid EMG sensor placement of the peroneus longus, in Chapter 6 for preparation for fine-wire insertion into the tibialis posterior and in Chapter 7 to measure the thickness and cross-sectional area of the soft tissue around the foot and ankle before and after three months of wearing a FO.

4.5.1 *Wave theory*

The term "ultrasound" is derived from the fact that the frequency of sound waves emitted (1-5 MHz in medical imaging and typically 5- 15 MHz in musculoskeletal applications) is above the frequency perceptible by the human ear (20 Hz- 2 MHz) (McDicken and Anderson, 2014, Schmidt, 2016). An ultrasound probe is a transducer, which contains piezoelectric crystals emitting sound waves into the soft tissue, and when there is a change in the density of the medium the wave is travelling through it is then reflected back towards the probe (McDicken and Anderson, 2014). Ultrasound gel is required for imaging to overcome the difference in density between air and skin and allow the transmission of signal through the skin (Lieu, 2010, Schmidt, 2016). B-mode ultrasound ("brightness"-mode), is the most common in musculoskeletal imaging and is also known as grayscale ultrasound (Lieu, 2010, Schmidt, 2016). The "brightness" of a structure on the constructed image is determined by the amplitude of the signal it reflects (i.e. echo amplitude), which is governed by its density in relation to surrounding tissues, this is known as echogenicity (Ihnatsenka and Boezaart, 2010, Lieu, 2010, McDicken and Anderson, 2014, Schmidt, 2016). Bone is *anechoic* and appears black, as ultrasound waves cannot pass through it, cartilage and muscle is *hypoechoic* (grey), and connective tissue is *hyperechoic* (white) (Ihnatsenka and Boezaart, 2010).

Ultrasound waves are longitudinal in nature because particles in the medium they travel through move in the same direction of travel as the wave itself, they are also called compression waves due to the pressure fluctuations they create (McDicken and Anderson, 2014). In a longitudinal wave the amplitude is the difference between the maximum pressure (compression) or minimum pressure (rarefaction) and the standard pressure in the tissue if the wave was absent (**Figure 4-15**). Thus, the amplitude and echogenicity are governed by material density due to how much the particles are able to compress. The wavelength (λ) in longitudinal waves is determined by the length between two consecutive compressions or two consecutive expansions/rarefactions (McDicken and Anderson, 2014). Frequency (f) is defined by the number of wavelengths per unit of time (per second or Hz, f=1/time) and is within a range set on the transducer (Lieu, 2010, Mohamed et al., 2010). So for a given period of time, as frequency increases, wavelength decreases (**Figure 4-15**) demonstrating an inverse relationship between the two (Lieu, 2010, Mohamed et al., 2010, Schmidt, 2016). In order to both emit and receive sound waves to create an image, the ultrasound wave is pulsed rather than continuous (**Figure 4-16**) (Lieu, 2010, Mohamed et al., 2010).

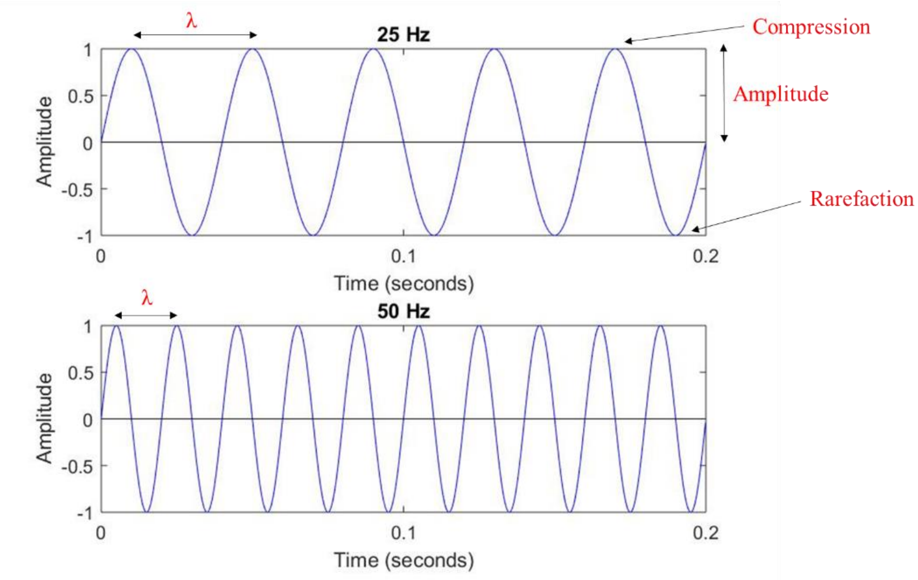


Figure 4-17. Sine wave of 25 Hz and 50 Hz. Compression is the maximum pressure and rarefaction is the minimum pressure. Amplitude is the height of the wave, which is the difference between compression or rarefaction and the pressure without the ultrasound wave (zero). Wavelength (λ) is the length between two consecutive compressions or two consecutive rarefactions. It can be seen that as frequency increases from 25 Hz to 50 Hz wavelength decreases.

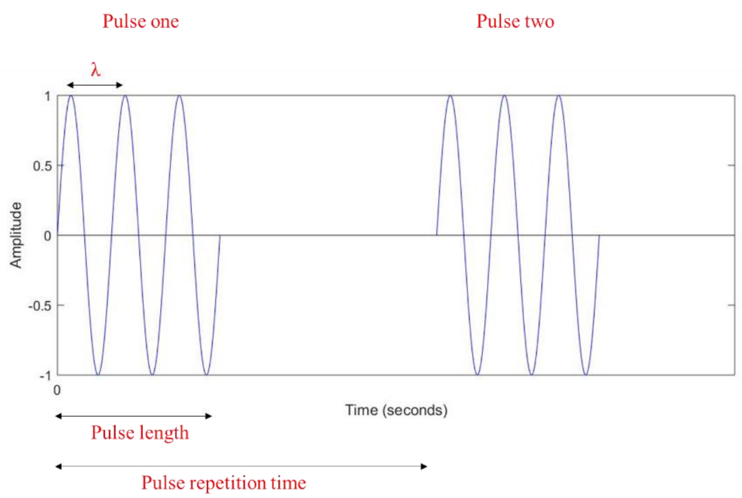


Figure 4-18. Ultrasound pulses of three cycles in length. Wavelength (λ) is the length between two consecutive compressions or two consecutive rarefactions. The spatial pulse length is the product of wavelength and the number of cycles in a pulse. Pulse repetition time is the time from the start of the first pulse to the start of the second pulse.

Wave speed (v) is the product of wavelength and frequency ($v = \lambda \cdot f$) (Lieu, 2010, Mohamed et al., 2010). However because of the inverse relationship between frequency and wavelength, as frequency increases wavelength decreases and speed remains constant. As such the speed the ultrasound wave travels through the tissue is dependent on the properties of the tissue itself (rigidity and particle density) and independent of changes in frequency, a property of the wave (Lieu, 2010, McDicken and Anderson, 2014). Ultrasound machines use the average speed of 1540 m/s for which ultrasound travels through soft tissue, along with the time it takes for the sound wave to return to the transducer after echoing off the structure (half the overall transit time) to calculate the depth of the structure (McDicken and Anderson, 2014).

Given:

$v = d/t$ then $d = v \cdot t$ (where v= velocity, d = distance and t = time)

Depth = 1540 m/s · time

The frequency of ultrasound scanning is important because it influences the penetration depth of the signal and the resolution, or clarity, of the image. The higher the frequency, the better the resolution, but the lower the penetration depth of the ultrasound waves (Ihnatsenka and Boezaart, 2010, Lieu, 2010, McDicken and Anderson, 2014, Mohamed et al., 2010, Schmidt, 2016). With higher frequency comes a greater amount of vibration through tissue lost as heat, which progressively attenuates the signal amplitude with increasing depth (McDicken and Anderson, 2014). As $f=1/t$ and $d = v \cdot t$ then depth = v/f (so penetration depth decreases as frequency increases). Broadband transducers transmit high frequencies for clear imaging of superficial structures, as well as low frequencies to enable identification of deep structures (Lieu, 2010).

In B-mode ultrasound spatial resolution (axial, lateral and elevation) and contrast resolution are of particular relevance. Spatial resolution is the minimum distance between structures that is required to distinguish them within the image, whereas contrast resolution is the minimum difference in the grayscale of the pixels, or echo amplitude, required to distinguish between two structures (Lieu, 2010, McDicken and Anderson, 2014). Axial resolution represents the clarity of the image in the direction of the ultrasound pulse (vertical in the image) and is determined by spatial pulse length, which is governed by the frequency (Mohamed et al., 2010). The spatial pulse length is the product of wavelength and number of cycles in a pulse (**Figure 4-16**), which is usually 2-4 for most ultrasound machines (Lieu, 2010, McDicken and

Anderson, 2014). As an increase in frequency means a decrease in wavelength (**Figure 4-15**), then an increase in frequency will also mean a decrease in spatial pulse length for a given number of cycles in a pulse. In order for two structures to be distinguishable on the image they must be at least half the spatial pulse length apart (**Figure 4-17**) (Lieu, 2010), so that when the sound wave hits the first structure and some is reflected back to the transducer, there is enough time for the wave transmitted to the second structure to be reflected back without the echo from the second structure coalescing with the echo from the first structure.

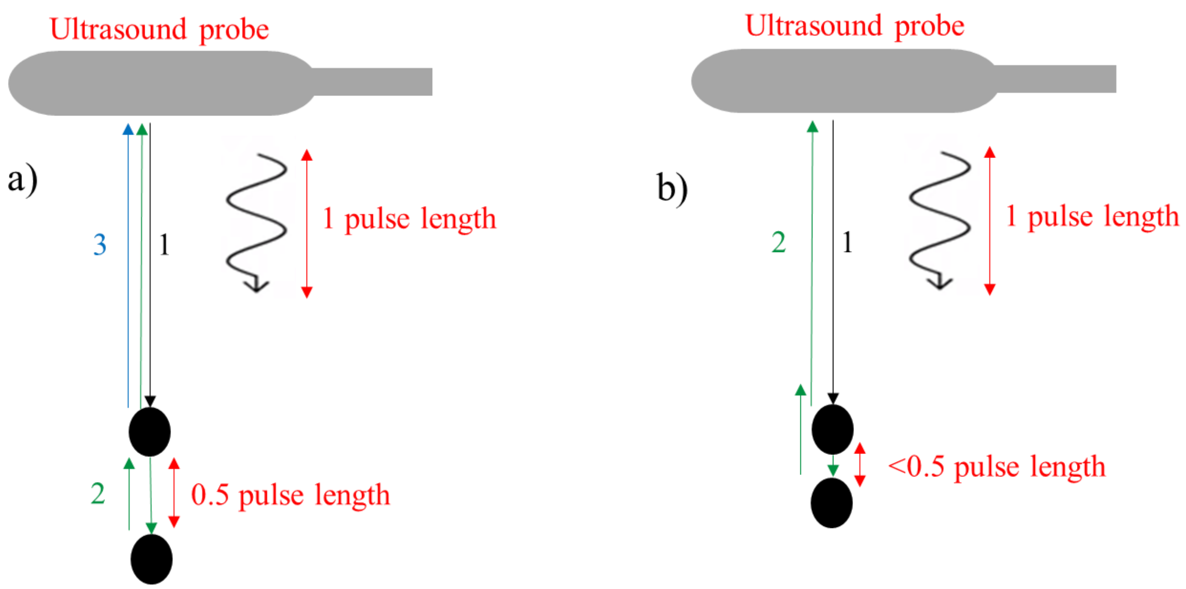


Figure 4-19. For two structures to be distinguished separately in an image the second structure must be at least half the pulse length from the first structure. Step 1 (black): the initial signal is transmitted. Step 2 (green): some of the signal is reflected off the first structure and some is transmitted through the first structure to reflect off the second structure. In a) by the time the echo from the second structure has returned to the level of the first structure, the echo from the first structure has returned to the transducer. So in step 3 (blue) the echo from the second structure is recorded separately from the echo of the first structure. In b) the distance between the two structures is less than half the pulse length so the echo from the second structure takes less time to return to the level of the first structure, so the returning signals are combined and the two structures are recorded as one.

Ultrasound imaging is relatively cheap and accessible, whilst also being less invasive when compared to other imaging modalities. Conventional x-rays, used only to image bone, along with computed tomography (CT), used to image multiple tissues, involve exposure to radiation (Ihnatsenka and Boezaart, 2010), and although magnetic resonance imaging (MRI) is a safe way to image soft tissues, it is expensive and often not feasible to use in repeated measures studies (Crofts et al., 2014). Ultrasound imaging is ideal in many ways, however it is somewhat user dependent with respect to adjusting settings to optimise image quality and concerning the use of the probe itself. Care should be taken to avoid applying excessive pressure to the skin with the probe that would compress the soft tissue leading to inaccurate measurements and may be uncomfortable for the participant (Ihnatsenka and Boezaart, 2010). The probe should be rotated to be in the correct orientation relative to the longitudinal or transverse axis of the segment (dependent on the measurement of interest) and the probe should be tilted to be perpendicular to the surface of the skin (Ihnatsenka and Boezaart, 2010). Correct probe position and orientation requires practice and is important so as not to affect the appearance of a structure in an image (Ihnatsenka and Boezaart, 2010, Martinoli, 2010). Additionally determining the thickest portion of the muscle for muscle thickness measurements is subjective (Crofts et al., 2014). Nonetheless with appropriate training reliable images of the soft tissue of the foot and ankle can be obtained (Crofts et al., 2014, Mickle et al., 2013).

4.5.2 *Piloting and training*

For Chapter 7 the thickness and cross-sectional area (CSA) of the following structures were piloted prior to data collection: abductor hallucis (ABH); flexor hallucis brevis (FHB); flexor digitorum brevis (FDB); flexor digitorum longus (FDL); flexor halluces longus (FHL); proximal, mid and distal plantar fascia and peroneus longus (PL) and peroneus brevis (PB). The protocol followed was as outlined previously (Angin et al., 2014, Crofts et al., 2014). Images of the Achilles tendon were also taken at the insertion (ATINS) on the calcaneus and the mid-portion (ATMID).

The following structures were not clear in piloting so were not included in the intervention in Chapter 7: FHL (thickness or CSA), distal plantar fascia and CSA of FHB. The compromise between axial resolution and depth explains why a clear image was not feasible for FHL because it is a deep muscle.

4.5.3 *Reliability*

An inter-rater reliability test for the dependent variables was performed on a group of seven participants between the candidate and a second researcher, because the second researcher collected some of the data in Chapter 7 and it was important to ensure consistent standards. The second researcher undertook the same piloting and training as the candidate. Reliability was assessed only from images from which a clear border of the deep and superficial aponeurosis in the central portion of the image could be distinguished (n=4-7). The CSA and thickness of the following structures were measured with at least a day between sessions: ABH, ATINS, ATMID, FDB, FHB, PL, PB, FHL and FDL and the thickness of proximal plantar fascia (PFINS) and mid-portion plantar fascia (PFMID).

The inter-rater and intra-rater reliability of the ultrasound measurements are presented in **Table 4-1**. The standard error of measurement (SEM) was calculated as the square root of the sum of squares of the between subjects standard deviation. The minimal detectable difference (MDD) was calculated as 1.5*SEM (Hopkins, 2000). Ultrasound imaging of the extrinsic foot muscles: PL, PB and FHL and FDL were not considered reliable based on our own measurements (**Table 4-1**) and on the large limits of agreement (LoA) in previous reliability work (Crofts et al., 2014), so not tested in the intervention. The peroneals were measured together in the previous paper and had some of the largest LoA as a percent of muscle size of the structures studied with 17.5% and 19.7% for thickness and CSA respectively. For FDL the LoA as a percent of muscle size were 15% and 26% for thickness and CSA respectively and for FHL 11% for both thickness and CSA (Crofts et al., 2014). The MDD difference of FDB was also high in our reliability study, likely due to possible differences in the positioning of the probe in the middle of the foot between measurements.

4.5.3.1 *Outcome of reliability work*

From the reliability and pilot work the thickness and cross sectional area of FDL, PB and PL were considered too unreliable to image and were not used in the long term effect of FO study in Chapter 7. Lower reliability was also found in extrinsic foot muscles compared to intrinsic muscles previously (Crofts et al., 2014) as the limits of agreement were high for both CSA for FDL (26%) and the peroneals (19.7%) and thickness (15% for FDL and 17.5% for the peroneals). The final scanning protocol for Chapter 7 was: thickness of ABH, FDB, FHB, PFINS and PFMID, ATINS, ATMID and CSA of: ABH, ATINS, ATMID and FDB. The MDD established in this chapter were used to interpret any differences between pre and post FO use in Chapter 7 in the context of measurement error.

Table 4-1. Ultrasound reliability data, n=4-7 due to some errors in data collection

		Rater 1				Rater 2				Intra-rater	
		Mean	**SD**	**SEM**	**MDD**	**Mean**	**SD**	**SEM**	**MDD**	**SEM**	**MDD**
ABH (n=7)	Thickness (mm)	10.24	2.49	0.12	0.18	9.75	2.44	0.49	0.73	0.49	0.74
(n=7)	CSA (mm^2)	193.59	46.57	11.03	16.55	193.41	44.55	23.41	35.12	4.82	7.23
ATINS (n=6)	Thickness (mm)	4.53	0.62	0.46	0.68	4.42	0.56	0.22	0.33	0.22	0.32
(n=5)	CSA (mm^2)	87.49	12.67	7.90	11.85	85.25	15.09	5.67	8.50	7.98	11.96
ATMID (n=5)	Thickness (mm)	4.38	0.72	0.24	0.36	4.65	0.57	0.36	0.54	0.30	0.45
(n=5)	CSA (mm^2)	57.88	7.19	2.30	3.45	61.94	17.78	2.35	3.53	9.45	14.17
FDB (n=4)	Thickness (mm)	12.03	2.57	0.95	1.43	10.14	2.18	1.32	1.98	1.43	2.14
(n=4)	CSA (mm^2)	218.07	39.17	9.66	14.50	209.15	71.50	27.89	41.84	26.91	40.36
FDL (n=7)	Thickness (mm)	13.86	3.57	0.90	1.35	13.36	2.84	2.53	3.80	1.74	2.61
(n=4)	CSA (mm^2)	137.09	51.46	59.76	89.63	160.10	86.50	78.22	117.32	38.11	57.16
FHB (n=5)	Thickness (mm)	12.48	2.78	0.49	0.73	12.47	2.33	0.44	0.66	0.69	1.03
PB (n=5)	Thickness (mm)	10.10	3.73	0.23	0.34	8.56	2.64	1.15	1.72	2.00	2.99
(n=5)	CSA (mm^2)	282.28	107.61	54.47	81.70	344.27	78.42	28.70	43.06	61.85	92.77
PL (n=5)	Thickness (mm)	15.87	5.98	0.76	1.14	12.84	4.60	1.67	2.50	2.62	3.93
(n=4)	CSA (mm^2)	477.01	133.82	34.35	51.53	482.39	116.66	12.24	18.35	13.55	20.33
PFINS (n=6)	Thickness (mm)	2.35	0.40	0.26	0.39	2.36	0.33	0.42	0.62	0.33	0.50
PFMID (n=7)	Thickness (mm)	1.74	0.22	0.27	0.41	1.57	0.16	0.05	0.08	0.20	0.30

SEM= standard error of measurement; MDD= minimal detectable difference

4.6 Summary

This chapter outlined the main methods used in this thesis: electromyography, motion and force assessment and musculoskeletal ultrasound. The theory and justification of the protocol, processing and analysis of the data were presented for each method. The development of the fine-wire electrode insertion technique was described in detail, which was necessary to perform the study in Chapter 6. The fine-wire EMG protocol outlined can be used to aid future recordings from tibialis posterior, as the technique is challenging and there is a lack of detailed guidelines available. Establishing the MDD of measures of soft tissue thickness and cross-sectional area was a necessary prerequisite for Chapter 7.

Chapter 5 RELIABILITY OF SURFACE EMG OF SELECT INTRINSIC AND EXTRINSIC FOOT MUSCLES AND FOOT AND ANKLE BIOMECHANICS

5.1 Introduction

In this chapter the reliability of peroneus longus (PL) and abductor hallucis (ABH) EMG and the medial longitudinal arch angle in walking was investigated primarily as reliable methods of collection had not been established. The PL EMG element was accepted for publication:

Reeves J, Jones R, Liu A, Bent L, Nester C. The between-day reliability of peroneus longus EMG during walking. Journal of Biomechanics 2019.

Reliability data was collected both shod with a flat inlay and barefoot for the sake of comparison to other reliability studies that were collected barefoot (Barn et al., 2012, Leardini et al., 2007, Murley et al., 2010b, Rabbito et al., 2011). For ABH EMG there was concern that the shoe may interfere with the EMG sensor which was another reason for collecting both barefoot and shod data, to have barefoot data as a reference. Additionally, it was thought that added material in the medial arch area with FO may interfere further with the EMG sensor, more than a shoe alone. For this reason, the relative reliability was assessed for PL EMG and ABH EMG by comparing whether the effect of FO, if any, was consistent between days.

Additionally, the reliability of tibialis anterior (TA) and medial gastrocnemius (MG) were also assessed so that measures of PL and ABH EMG could be compared to those from more commonly recorded shank muscles. Lastly the reliability of discrete kinematic and kinetic variables were assessed in the shoe only condition, in order to obtain an indication of the typical error of the measurements for the interpretation of possible effects of FO in Chapter 6.

5.1.1 <u>*Rationale*</u>

The PL and ABH are important in foot function as they control frontal plane foot motion and assist with medial arch support respectively. Investigating the function of these muscles with EMG is challenging. Due to its close proximity to adjacent muscles, EMG measures of PL may be susceptible to crosstalk, thus correct electrode placement is vital (Campanini et al., 2007). Isolating the signal from ABH is difficult due to its small size. Consequently, measurement error is an issue in recording EMG from PL and ABH. Previous electromyography (EMG) studies have reported poor between-day reliability of PL during walking, which could affect our ability to study intervention effects in this muscle (Barn et al., 2012, Murley et al., 2010b).

In patients with rheumatoid arthritis (RA), for example, peak PL amplitude in the combined midstance/propulsion phase had an ICC of 0.03-0.19 and standard error of measurement (SEM) of 17-18% (Barn et al., 2012). Test and retest reliability can be considered one of the most important factors regarding measurement accuracy (Hopkins, 2000). A measure cannot be considered valid if it is has significantly high error and is not reliable, as reliability is a requisite of validity (Thomas et al., 2011).

Surface EMG signal of PL is susceptible to crosstalk and a shift in electrode position of approximately 2-3 cm could change peak PL activation relative to the central location by up to 29% (±13%) (Campanini et al., 2007). Crosstalk with tibialis anterior (TA) particularly, could explain poor between-day reliability of PL surface EMG in the study by Barn and colleagues (2012) if electrodes were inadvertently placed closer to the TA during one session than another. With a small sample size (n=5), the mean difference between sessions would be particularly susceptible to extreme errors in electrode placement. Identifying the borders of the PL with prior ultrasound scanning may facilitate sensor placement. Additionally, with small EMG sensors muscle borders can be more easily avoided. The small head (25 mm x 12 mm x 7 mm) of the Delsys Trigno™ Mini sensor (Fig. 1., Delsys, Inc., Boston, USA) makes it suitable for recording signals from muscles that are difficult to isolate and susceptible to crosstalk. The presence of rheumatoid arthritis (RA) might also have added variability by influencing muscle function in the study by Barn et al (2012). Thus, the reliability of surface PL EMG in a healthy population is unknown.

Using fine-wire EMG (Murley et al., 2010b), eliminates the problem of crosstalk, however it has been challenged that intramuscular wire electrodes are sensitive to their positioning regarding the proximity to active motor units (Kadaba et al., 1985). Furthermore, if good reliability of PL EMG can be achieved using surface electrodes it would be advantageous because fine-wire EMG is invasive and requires specialist skills and equipment.

Potential variation in PL and TP EMG between sessions did not appear to be overcome by normalising to maximum voluntary contractions or dynamic normalisation to a self-selected fast walking speed (MVCs) (Murley et al., 2010b). Performing an MVC itself introduces a potential source of error, particularly when participants are unlikely familiar with performing the expected maximal movements in the frontal plane. Isolating the contraction of only PL or TP may also not be possible and the uncontrolled co-contraction from other muscles in the shank is probably inevitable (Onmanee, 2016). In fine-wire studies participants may also not truly produce maximum effort if a forceful contraction causes discomfort with fine-wire in the muscle. Previous work in our group found that normalising EMG signals from shank muscles, including TP, to MVCs had greater between-session variability than normalising to the peak (average of the maximum values from six included gait cycles) or mean (average over the gait cycle, averaged over the six included gait cycles) (Onmanee, 2016). Normalising to the peak (maximum value from each gait cycle then averaged across trials) is the method used in this thesis as it showed good between-day reliability of EMG and is more intuitive to interpret than normalising to the mean (Onmanee, 2016). Given the good between day reliability of TP normalised to the peak, it was considered unnecessarily invasive to perform fine-wire EMG for further study of reliability, particularly not additional MVCs as anecdotally these were reported as uncomfortable.

The reliability of ABH EMG during walking is unknown as EMG studies of this muscle in vivo have typically been in free exercises or running (Incel et al., 2003, Jung et al., 2011, Kelly et al., 2012, Kelly et al., 2016). A few historic EMG studies of ABH in walking were performed using in-dwelling electrodes and found high between-subject variation and no consistent pattern of activation (Reeser et al., 1983). It has yet to be established whether ABH EMG during walking is reliable using modern and small surface EMG sensors.

The behaviour of the MLA during walking is potentially advantageous in understanding foot function (Welte et al., 2018) and may be particularly important in understanding the effect of FO, given that many FO have added arch support. The MLA angle has been tracked dynamically in barefoot walking (Leardini et al., 2007, Rabbito et al., 2011), however the reliability of the MLA during shod walking in unknown. Reflective markers placed on the skin likely represent motion of the foot within the shoe better than markers placed on the shoe surface (Bishop et al., 2015). A simplified multi-segment model of in-shoe foot kinematics was developed to extend the work of Bishop and colleagues on a multi-segment kinematic model of the foot-shoe complex and the size of holes in the shoe required to avoid collision of markers with the shoe (Bishop et al., 2013, Bishop et al., 2015, Bishop et al., 2017). It is necessary to understand the reliability of such a model so that any variables from the model, such as the deformation of the MLA, can be confidently linked with the patterns of muscle activation.

Based on the above discussions the following research questions and aims were determined for this part of the study:

5.1.2 *Research question*

Is peroneus longus and abductor hallucis EMG and medial longitudinal arch angle reliable in walking?

5.1.3 *Aim:*

1) To use ultrasound to aid placement of small EMG electrodes and determine the reliability of PL and ABH in walking.
2) To determine the between-day reliability of MLA angle using a foot-shoe model.
3) To establish reference values of the between-day reliability of TA and MG EMG and discrete kinematic and kinetic variables.

5.2 Methods

The methods used in this study for data collection, data processing, biomechanical modelling based on marker data are same as those presented in Chapter 4.

5.2.1 *Participants*

Software for estimating sample size does not accommodate studies on the reliability of measures (Hopkins, 2006). However, Hopkins did state that very high reliability could be demonstrated with as few as 10 participants and that 50-100 participants would be needed for more moderate values of reliability. Fifty participants are not feasible within the time constraints of these studies. In the developmental paper on the reliability, accuracy and minimal detectable difference of the multi-segment model of Bishop et al. (2013), 12 participants were recruited for the reliability study. Therefore, it was deemed realistic and justified to recruit 10 for this study.

A convenience sample of 11 healthy participants were recruited from the university population. Exclusion criteria were 1) any recent lower limb injury or foot or ankle pathology including hallux valgus; 2) any cardiovascular, musculoskeletal or neurological conditions or disease 3) walks with an aid and 4) have been wearing prescribed orthotics during the last six months. Participants attended two data collection sessions separated by at least a day. Data collection was not completed for one participant due to persistent difficulty in affixing markers to their feet. Ten participants (two males, eight females) were thus included for analysis (**Table 5-1**). Participants were not categorised according to foot posture.

The study conformed to the Declaration of Helsinki and was granted approval by the University of Salford Research, Innovation and Academic Engagement Ethical Approval Panel (HSCR 16-92, appendix).

Table 5-1. Participant characteristics (n=10)

	Mean	±SD	Median	Range
Age (years)	28	±4	29	(20-32)
Height (m)	1.69	±0.07	1.69	(1.60-1.83)
Mass (kg)	67.13	±9.22	64.25	(58.10-85.00)
Shoe size (UK)	7	±1	6	(6-9)
Duration of time between sessions (days)	5	±3	4	(1-11)

5.2.2 *Protocol*

Height and mass were measured using a stadiometer and scales respectively.

Participants walked at a self-selected speed for four conditions in two sessions, over two days (mean ±SD: 5 ±3 days apart). The conditions were barefoot, shoe with the manufacturer's inlay, shoe with a standard FO and shoe with a +6 mm high arch. The FO were from a previous PhD student study at the University of Salford and were fabricated from high density Ethylene vinyl acetate (EVA, Shore 65) with the aid of CADCAM to accurately define the orthosis geometry (**Figure 5-1**) (Sweeney, 2016). High density EVA is one of the most commonly used materials used to make pre-fabricated FO in clinical practice (Nester et al., 2017). The shoes were standardised trainers (Lonsdale Leyton, London) (**Figure 5-4**.). The order of conditions was randomised. Each participant would be given a few minutes to walk around to accommodate the FO and shoes before data was collected. Walks were recorded over ~6 m in a university gait laboratory, with data recorded in both directions. Six trials per condition during each session were recorded.

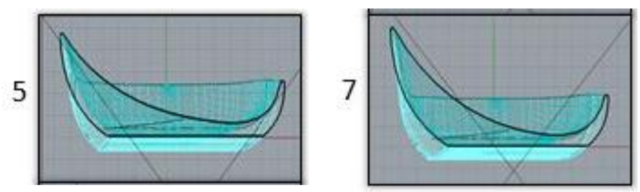


Figure 5-1. Orthotic conditions from (Sweeney, 2016) with 5) a standard arch and 7) a +6mm arch height

5.2.3 *EMG*

EMG signals were collected on PL, ABH, TA and MG of the right leg using Trigno™ wireless sensors (Delsys, Inc., Boston, USA). The leg chosen for the study was the right leg, as it is most often the dominant leg and without any reason to believe the effect of the FO would differ between legs in healthy participants, data from the left leg was considered redundant. Signals from TA and MG were recorded so that measures of PL and ABH EMG could be compared to those from more commonly recorded shank muscles that are generally considered reliable (Murley et al., 2010b, Winter and Yack, 1987). Standard Trigno™ LAB sensors (99.9% sliver contact material in single differential configuration, inter-electrode distance 10 mm, 4-bar formation, were used for the TA and MG and Trigno™ Mini sensors for PL and ABH (**Figure 4-1)**. The guidelines for Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000) were followed for placement of Trigno™ standard sensors on the TA and MG.

For PL firstly the approximate sensor location was found based on the SENIAM guidelines, which were as follows: with the participant supine on a physiotherapy plinth a mark was made using eyeliner pencil on the lateral malleolus and the head of the fibula, which was identified by palpation. A third mark was then made at 25% of the distance between these two marks, closest to the fibula head. Secondly the muscle belly of PL was scanned in the transverse plane at the level of the middle mark using ultrasound at 5-8 MHz (Linear 60 mm probe, Echo Blaster 128 CEXT, Telemed Medical Systems, Milan, Italy (**Figure 5-2**). The ultrasound probe was coated in water soluble gel for acoustic contact. A blunt metal stick was placed under the probe to gently apply pressure to the skin, so that deformation was visible in the ultrasound image in line with the two lateral borders of the muscle in turn. The ultrasound probe was removed and a mark was made by the stick with a surgical marker pen, which was able to write through ultrasound gel. The marks were typically within ~1 cm of the SENIAM placement in the longitudinal direction, but often more anterior/posterior of the line between the fibula head and malleolus depending on the individual. Marks from session one were not visible in session two. The skin was then cleaned of the gel and wiped with an alcohol wipe. The sensor head of a Trigno™ Mini was then affixed between the two markers orientated in the longitudinal direction of the shank and PL muscle fibres. The main sensor of the Trigno™ Mini was affixed over the tibia. Tape was fixed over the sensor head and cable to minimise potential movement artefact (**Figure 5-4**).

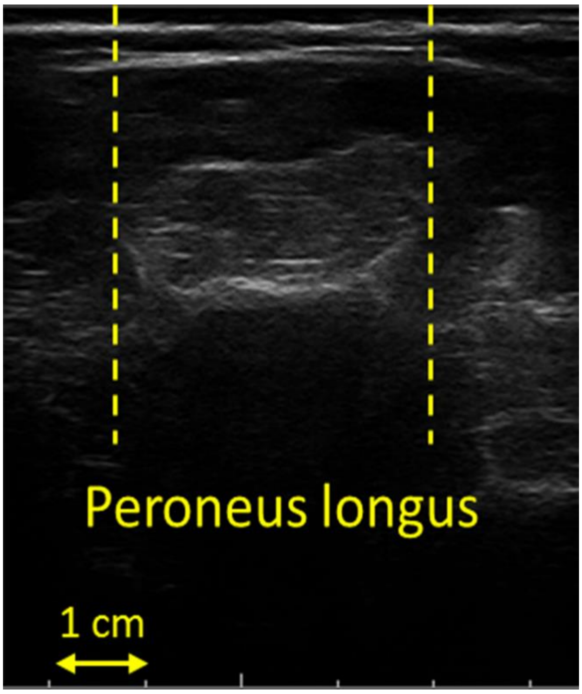


Figure 5-2. Ultrasound image of the peroneus longus in the transverse plane, dashed lines indicate the lateral borders of the muscle

Currently no established guidelines exist for surface EMG electrode placement for ABH. The foot was scanned using ultrasound with the probe orientated in the direction of the toes. The muscle belly was identified as approximately the thickest portion of the muscle in the ultrasound image, which was marked with the surgical marker (**Figure 5-3**). This typically resulted in an electrode placement approximately in line with the anterior aspect of the medial malleolus, which is in agreement to the protocol used in a previous study of ABH EMG during isolated contractions (Incel et al., 2003). The sensor head was orientated with the arrow pointing in the direction of the toes and the main sensor was attached to the distal shank, with the arrow orientated upwards. Tape was placed over the sensor head to maintain contact with the foot as firm as possible and over the cable to reduce its movement artefact.

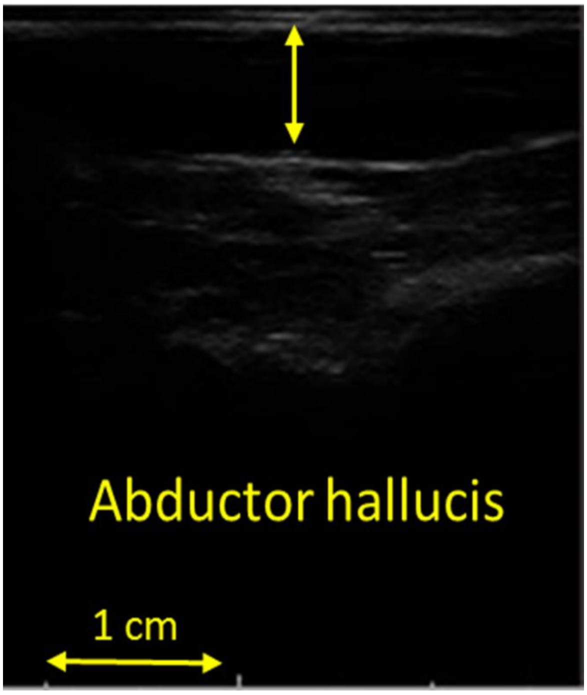


Figure 5-3. Ultrasound image of the abductor hallucis, vertical arrow represents the thickness of muscle

5.2.3.1 *Testing the signals*

Prior to collecting the EMG signal, each muscle was tested by asking the participant to perform some functional movements. If the signal appeared unsatisfactory then the sensor was removed, the location wiped again with an alcohol wipe and the sensor reapplied with a new adhesive sticker. The functional movements were dorsiflexion while standing on the left leg, plantar flexion by standing on their toes and eversion while standing on their left leg for TA, MG and PL respectively.

The functional movements for ABH were a toe-shortening movement (short foot) and abduction. For the short foot exercise participants were asked while standing on both feet to try to actively shorten their right foot by bringing their big toe towards their heel while keeping the foot on the ground (Jung et al., 2011). For hallux abduction participants were asked to move their big toe towards a pen placed medial to the medial border of their foot. The ability of participants to perform these movements varied considerably, some individuals were able to perform one task better than the other and some individuals could not perform either well.

Following the functional test, the participant was asked to stand on his/her left leg and raise the right leg to push down with the lesser toes against the hand of the investigator. The aim of this test was to see whether there was an absence of EMG activation so as to assure that no EMG signal was detected from the flexor digitorum brevis. The participant's heel was tapped to ascertain if movement artefact might be present during foot contact.

5.2.4 *Kinematics and Kinetics*

5.2.4.1 *Marker set up*

Individual markers (9 mm diameter) were placed bilaterally as per the Marker setup above in the Methods chapter (**Figure 5-4**).

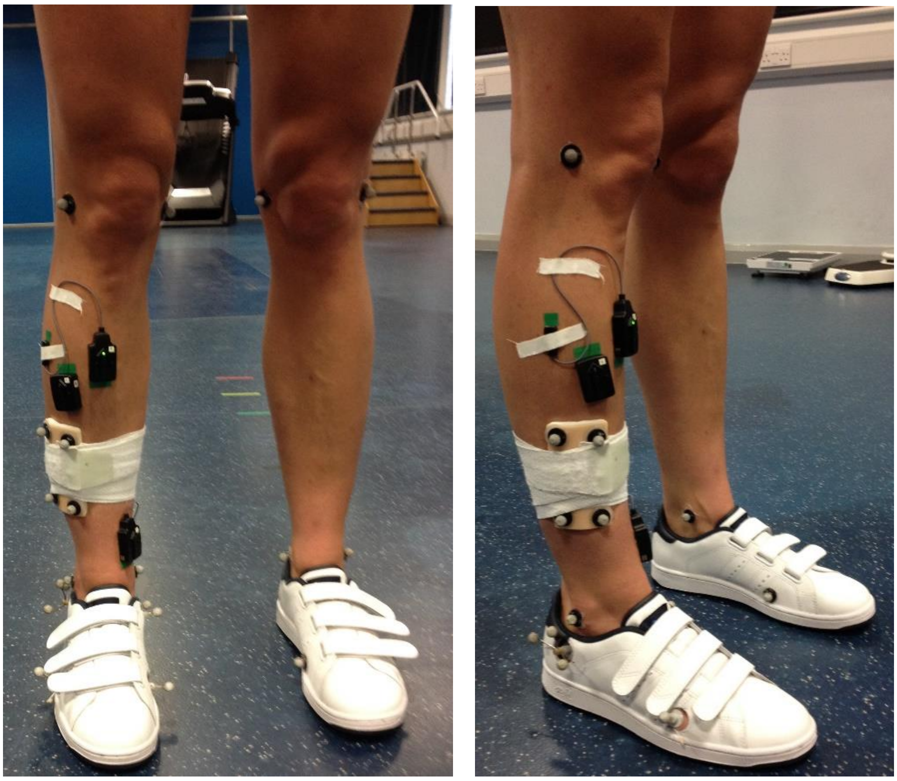


Figure 5-4. Location of the markers and EMG electrodes from the front and side

Custom made reflective markers on wands were attached to mounting plates through holes in the right shoe at MTP1, MTP5, the navicular tuberosity and medial aspect of the calcaneus. The Lonsdale Leyton trainers were deliberately chosen for this study because 1) the VELCRO® design enabled a quick change between conditions and 2) the firm heel cup meant that enough of the trainer's integrity could be maintained after cutting a hole in it.

5.2.4.2 *Medial longitudinal arch angle*

The medial longitudinal arch (MLA) angle was calculated in Visual3D as specified in Chapter 4 (**Figure 4-16**).

5.2.4.3 *Definition of kinematic and kinetic discrete variables*

Each discrete variable was calculated per trial and then a mean calculated for each participant. The abbreviations and definition of the kinematic and kinetic variables of interest are outlined in (**Table 5-2**). Variables were chosen if they were considered clinically meaningful and outcomes that a therapeutic FO intervention would seek to change (Telfer et al., 2013b). Previous work has shown that FO have the potential to reduce peak rearfoot eversion angle (Majumdar et al., 2013, Mills et al., 2009, Stacoff et al., 2007, Telfer et al., 2013b), rearfoot inversion/eversion ROM (Novick and Kelley, 1990, Zifchock and Davis, 2008) and peak ankle eversion moment (Stacoff et al., 2007, Telfer et al., 2013b).

Table 5-2. Definition of the discrete kinematic and kinetic variables

	Abbreviation	Definition	Calculation
Kinematics	MaxEv	Peak rearfoot eversion	Mean of the minimum calcaneus angle in frontal plane from each trial
	ROM	Eversion range of motion	Difference between maximum calcaneus angle during initial contact phase (first 5% of stance) and maximum eversion
	MaxES	Inversion at foot contact	Maximum calcaneus angle in the frontal plane during initial contact phase (first 5% of stance)
Kinetics	MaxMEv	Peak external eversion moment	Minimum ankle moment in frontal plane
	MaxMInv	Peak external inversion moment	Maximum ankle moment in frontal plane

5.2.5 *Statistics*

Inter-individual variability was assessed visually from the average EMG profiles. The coefficient of multiple correlation (CMC) was calculated to compare between sessions using a Microsoft Excel spreadsheet. Between day reliability was assessed using the standard error of measurement (SEM: SD$\Delta/\sqrt{2}$) and the intra-class correlation coefficient (ICC (3,1)) for the peak and timing of the peak from a freely downloadable Microsoft Excel spreadsheet for consecutive pairwise analysis (Hopkins, 2015). The SEM and ICC (3,1) were calculated for the inlay (shoe) and barefoot data. Negative ICC values are presented as 0.00 (Stanish and Taylor, 1983).. The SEM can be susceptible to the heterogeneity of the sample and it is sometimes more appropriate to express the error of measurement as a percentage of the mean, using a coefficient of variation (CV) (Hopkins, 2000). The CV was calculated as the (SEM/Grand mean)*100 as this has a similar accuracy to log-transforming the data and is easier to compute (Batterham and George, 2003, Hopkins, 2000). Within day reliability was assessed by visual inspection of plots of individual gait cycles of the inlay and barefoot conditions (**Figure 0-1**). Relative reliability was assessed visually by observing whether any potential effect of the condition was consistent between days.

5.3 Results

5.3.1 *Stride time*

The mean stride times for each condition on each day are presented in **Table 5-3** as an indication of walking speed.

Table 5-3. Mean ±SD stride times for the barefoot and shoe conditions on day 1 and day 2

	Barefoot		**Shoe**	
Day 1	1.04	±0.05	1.08	±0.04
Day 2	1.01	±0.05	1.04	±0.04

5.3.2 *EMG*

The group means ± SD EMG profiles in the shoe (inlay) condition and barefoot are presented in (**Figure 5-5**) and discrete variables in **Table 5-5** and **Table 5-6**. In the shoe condition both PL and ABH demonstrated repeatable EMG profiles and good CMCs (0.88 and 0.85 for PL and ABH respectively). For PL, reliability was moderate for the peak (SEM: 4% of peak, ICC: 0.44) and very good for the timing of the peak (SEM: 2% of gait cycle, ICC: 0.72). For ABH, reliability of peak amplitude was moderate considering the ICC value (0.63), but poor based on the SEM (14% of peak). Reliability of the time of peak ABH was poor (SEM: 8% of peak, ICC: -0.03).

There was no difference between conditions for PL average EMG, across both sessions. The difference between conditions was not maintained in all participants for the ABH. In some cases the inlay had greater ABH activation than the FO conditions and in other cases it was the same or less than the FO conditions, indicating the relative reliability of the mean ABH EMG profile is poor.

5.3.3 *Kinematics and kinetics*

Ankle kinematics and kinetics over stance in the sagittal and frontal planes are presented in **Figure 5-8** for the shoe condition. Discrete kinematic and kinetic variables for the frontal plane are presented in **Table 5-7** and in the shoe condition. Mean dorsi-plantar flexion angle across stance in all conditions is shown in **Figure 5-10** and mean external dorsi flexion moment across stance in all conditions is shown in **Figure 5-9**. There was less between subject variability in the sagittal plane than frontal plane.

5.3.4 *Medial longitudinal arch*

Due to the limited space and the difficulties in marker placement, marker data for MLA calculation on some subjects were not collected but the data collections from six participants were successful. The mean range of motion of the MLA in each condition is shown in (**Table 5-4**) and the mean MLA over stance for each condition is shown in (**Figure 5-11**). Although there was an offset between sessions in some conditions, the overall MLA trace over stance was consistent. The difference in ROM between sessions was generally small, except in participants P5 and P11 (difference of 10.8° in the Salfordinsole® and difference of 8.4° in shoe only condition respectively). The SEM and MDD of the ROM of the MLA was 2.8° and MDD 4.2° respectively (based on the shoe only data).

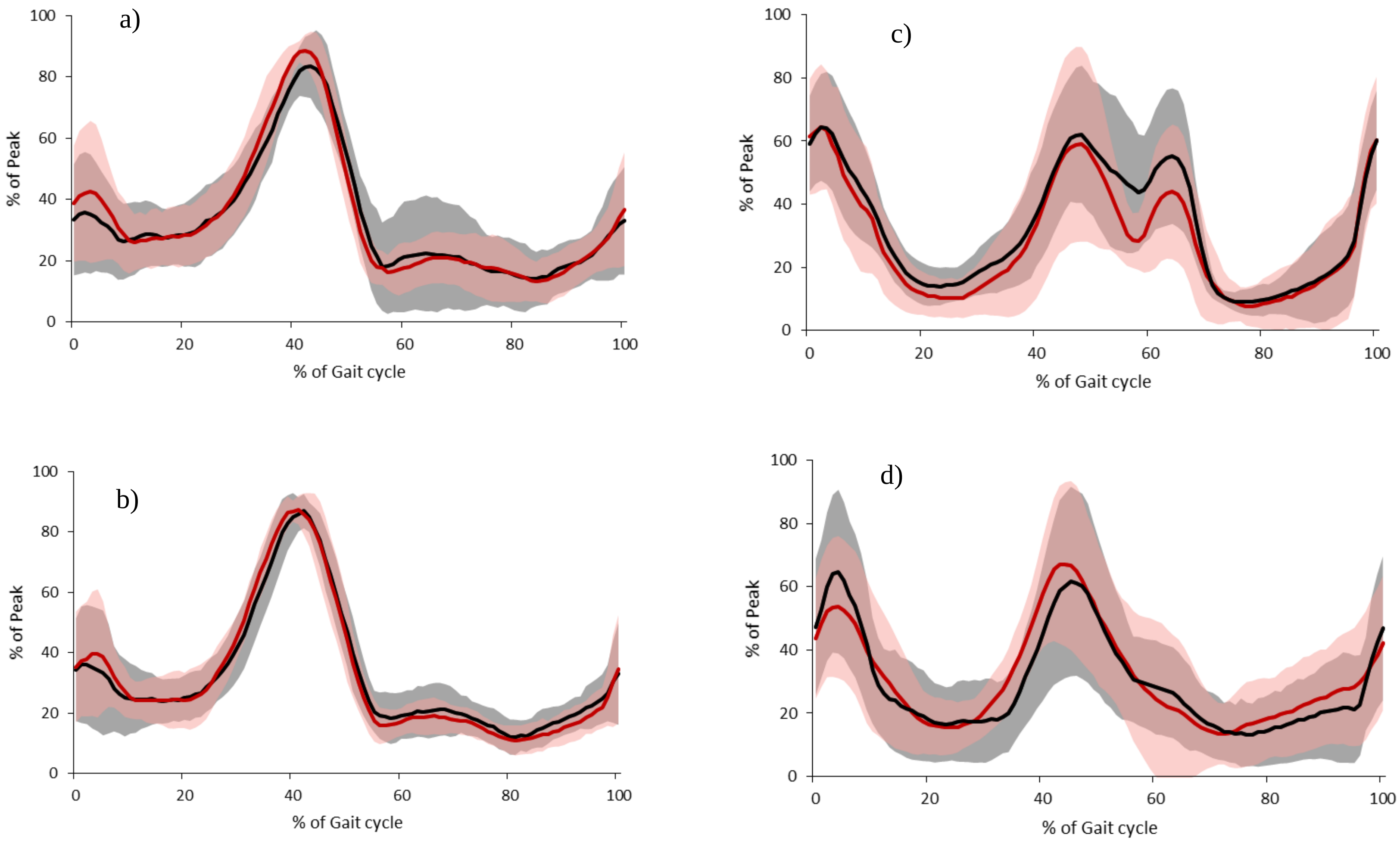


Figure 5-5. EMG ensemble averages ± SD over the gait cycle for peroneus longus in a) shoes and b) barefoot and abductor hallucis in a) shoes and b) barefoot. Day 1 and 2 (black and red lines respectively).

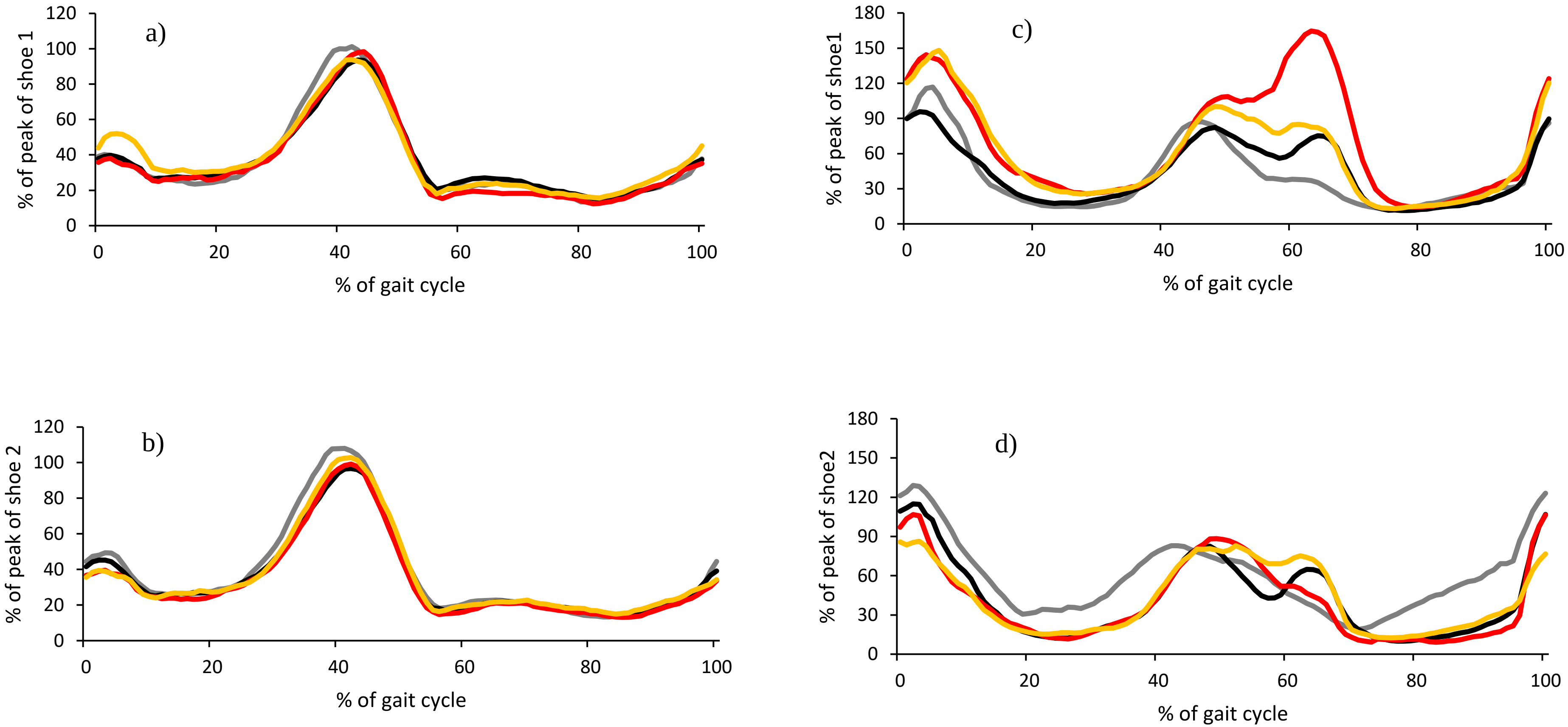


Figure 5-6. EMG ensemble averages from a) Peroneus longus in day 1 b) Peroneus longus on day 2, c) Abductor hallucis on day 1 and d) Abductor hallucis in day 2. Grey lines: barefoot; black lines: shoe only; red lines: high arch; yellow lines: Salfordinsole

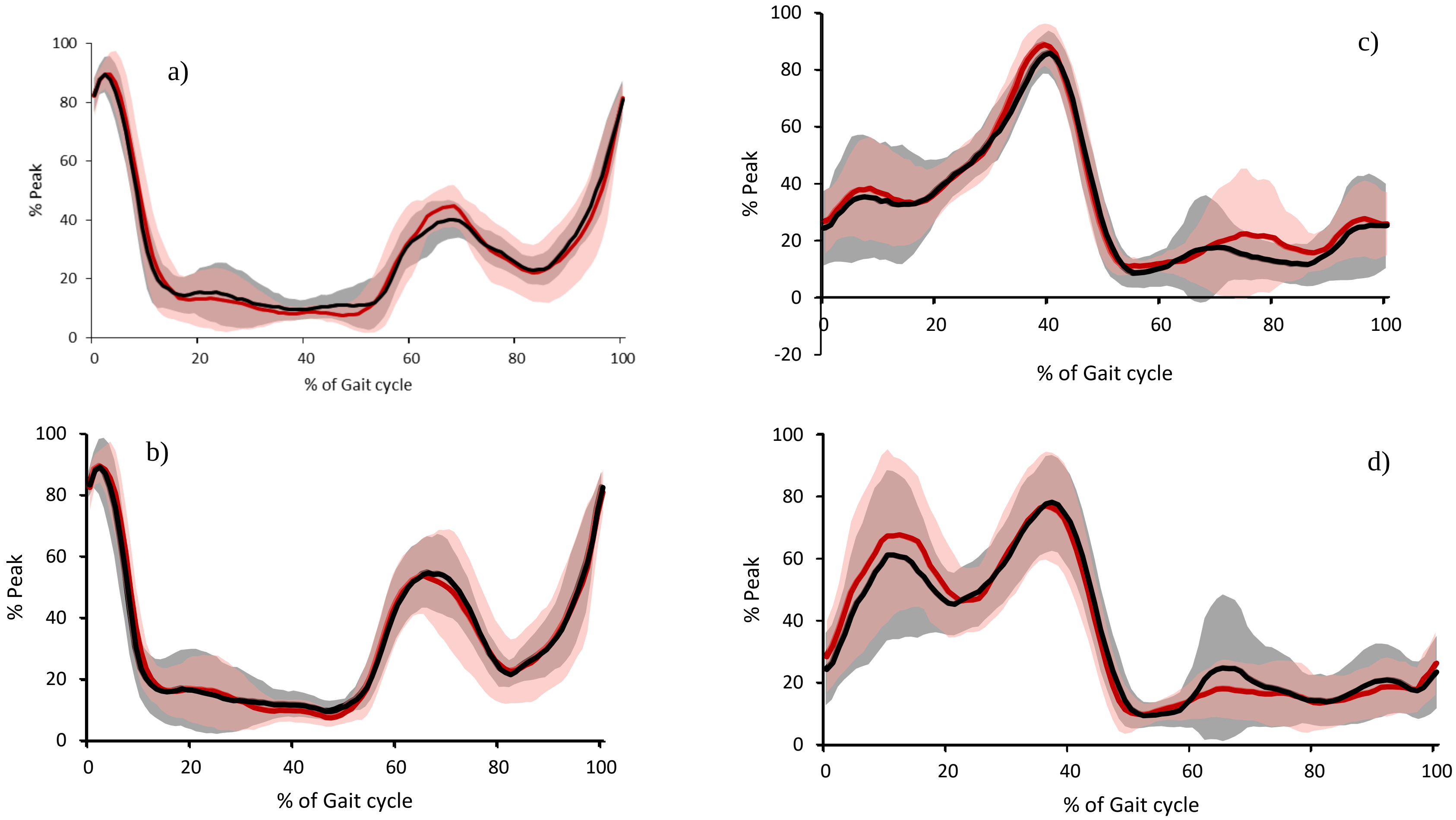


Figure 5-7. EMG ensemble averages ± SD over the gait cycle for tibialis anterior in a) shoes and b) barefoot and medial gastrocnemius in c) shoes and d) barefoot. Day 1 and 2 (black and red lines respectively)

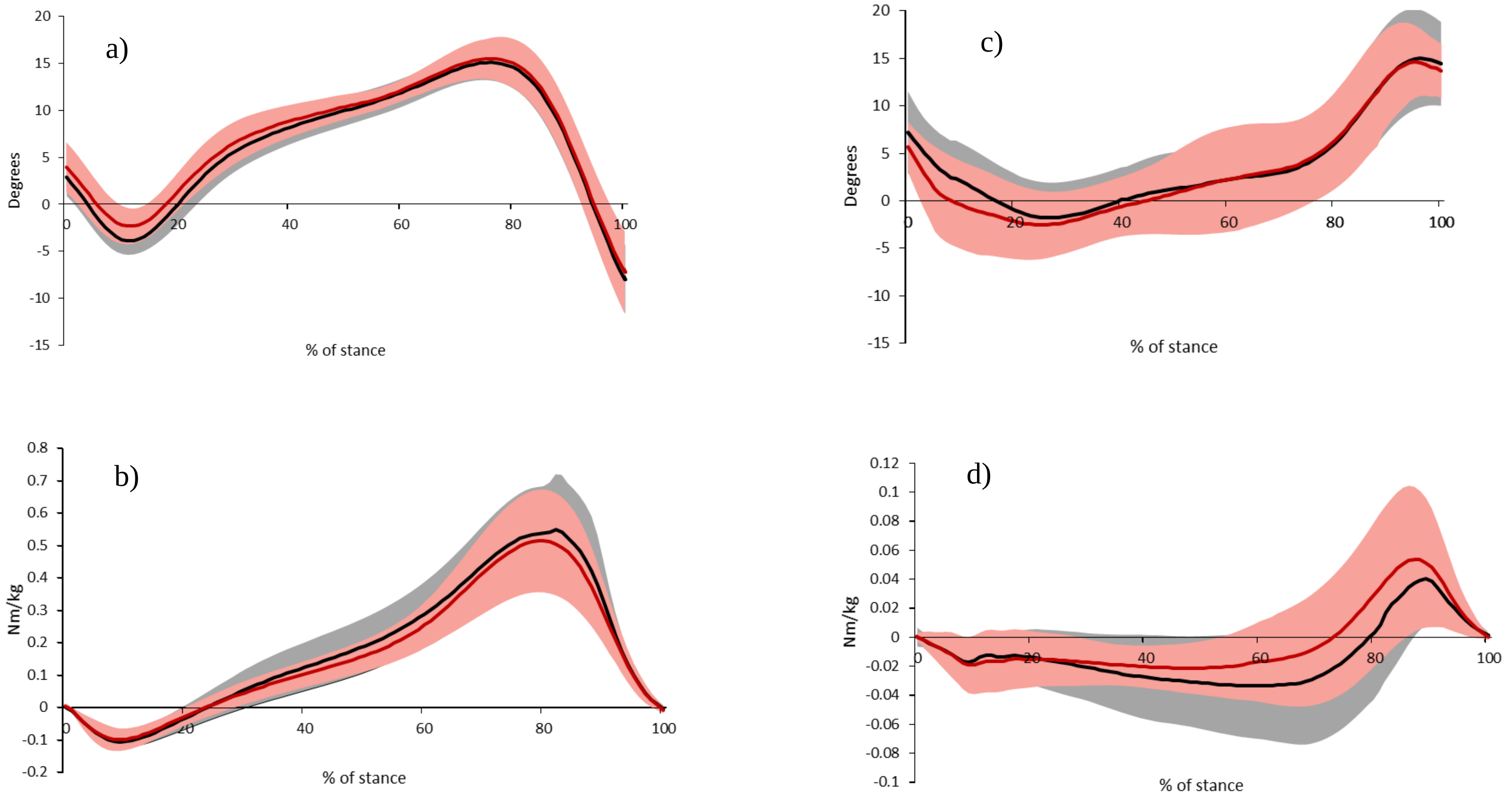


Figure 5-8. Mean ± SD kinematics and kinetics in shoes expressed over stance: a) Dorsi-plantar flexion angle (+ dorsiflexion/- plantar flexion), b) external dorsi-plantar flexion moment (+ dorsiflexion/- plantarflexion), c) rearfoot angle frontal plane (+inversion/- eversion), d) external inversion/eversion moment (+inversion/- eversion). Day 1 and 2 (black and red lines respectively)

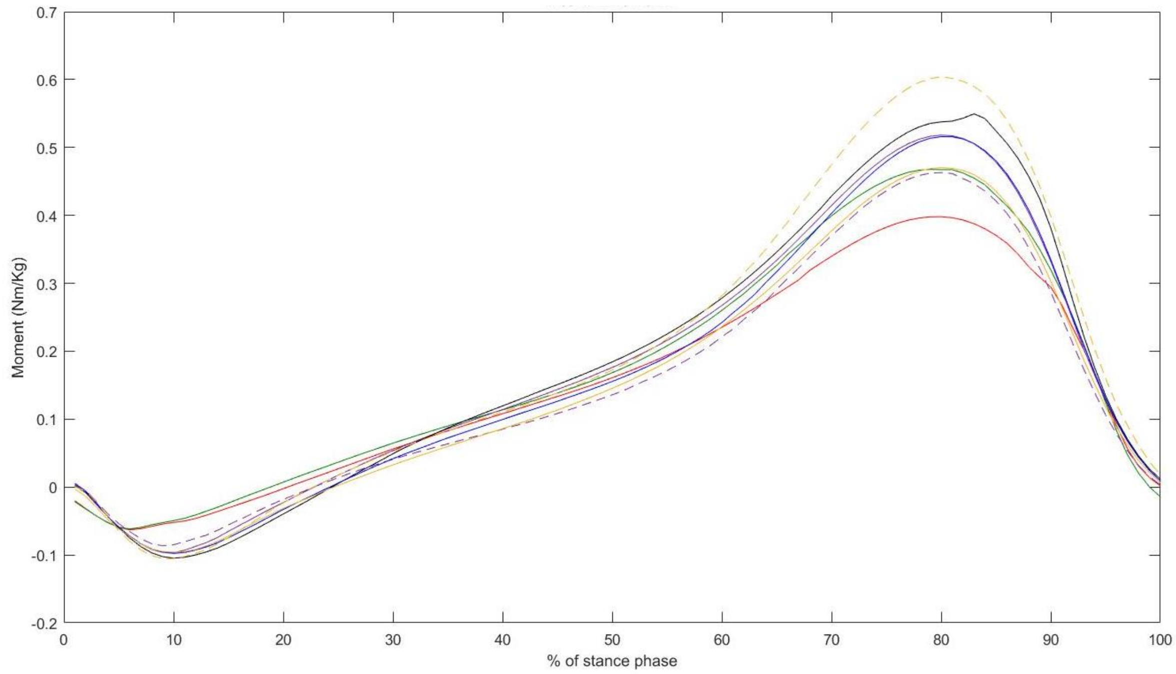


Figure 5-9. Mean external dorsi flexion moment (+ dorsiflexion/- plantar flexion), across stance for all conditions. Red and green lines represent barefoot on day 1 and day 2 respectively. Black and blue lines represent shoe on day 1 and 2 respectively. Yellow and purple lines represent Salfordinsole and high arch respectively (bold lines day 1 and dashed lines day 2).

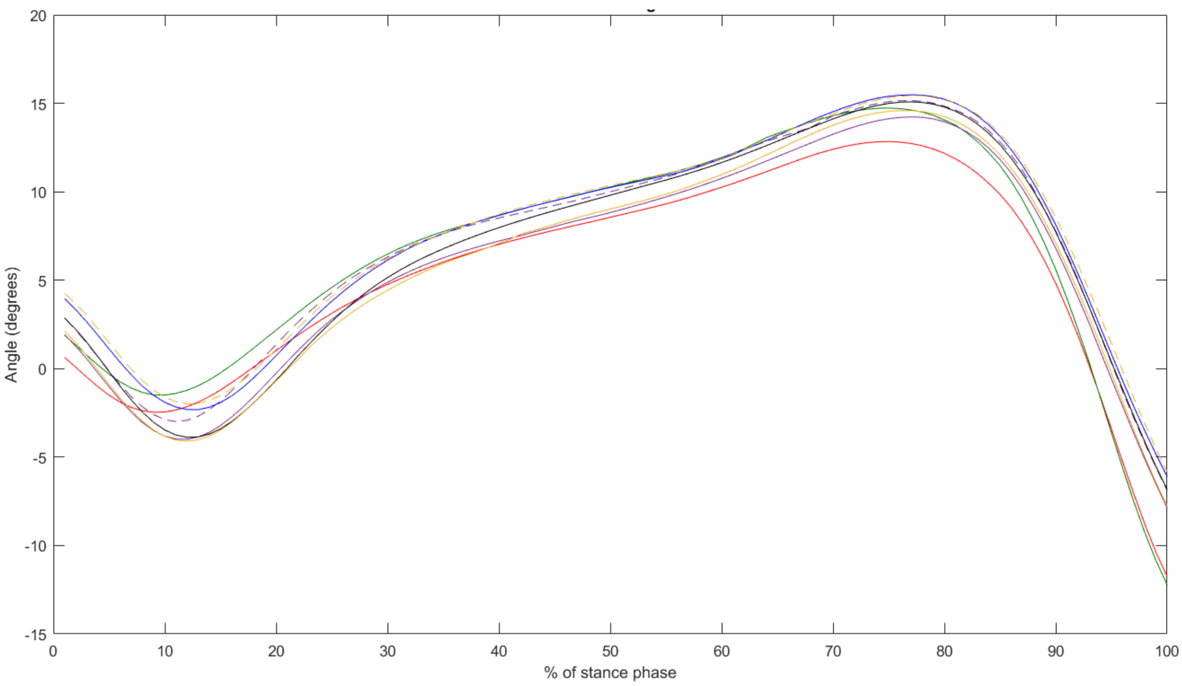


Figure 5-10. Mean dorsi-plantar flexion angle across stance for all conditions, (+ dorsiflexion/- plantar flexion). Red and green lines represent barefoot on day 1 and day 2 respectively. Black and blue lines represent shoe on day 1 and 2 respectively. Yellow and purple lines represent Salfordinsole and high arch respectively (bold lines day 1 and dashed lines day 2).

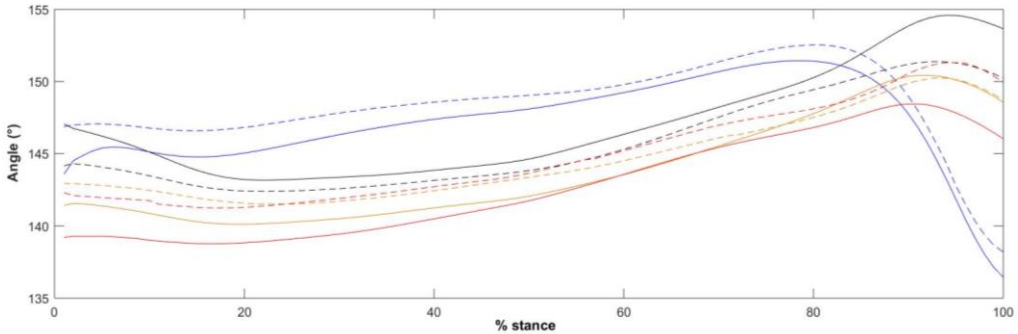


Figure 5-11. Mean medial longitudinal arch angle over stance. Blue lines: barefoot, black lines: shoe only, yellow lines: Salfordinsole, red lines: increased arch height. Bold lines: day 1, dashed lines: day 2 (for six participants)

Table 5-4. Mean range of motion of the medial longitudinal arch angle (°) over stance for six participants

Participant ID	**Session 1**				**Session 2**				**Difference**			
	Barefoot	Shoe	Salfordinsole	High arch	Barefoot	Shoe	Salfordinsole	High arch	Barefoot	Shoe	Salfordinsole	High arch
P5	13.6	15.9	19.4	16.8	12.9	14.9	8.7	11.6	-0.7	-1.1	-10.8	-5.1
P7	9.5	6.5	6.2	6.7	8.8	6.7	8.3	9.5	-0.7	0.2	2.1	2.8
P8	15.9	9.6	6.9	8.1	18.4	6.3	7.6	8.2	2.4	-3.3	0.6	0.1
P9	14.8	16.8	13.9	13.6	10.8	13.6	9.9	No data	-4.0	-3.1	-3.9	
P10	14.6	10.3	15.5	14.3	13.2	9.9	11.7	15.2	-1.4	-0.4	-3.7	0.9
P11	26.7	18.7	9.3	6.9	28.5	10.2	11.7	13.4	1.8	-8.4	2.4	6.5
Mean	15.9	13.0	11.9	11.1	15.4	10.3	9.6	11.6	-0.4	-2.7	-2.2	1.0
SD	5.8	4.8	5.2	4.4	7.1	3.5	1.8	2.8	2.3	3.2	5.0	4.2

Table 5-5 Discrete EMG variables for peroneus longus and abductor hallucis in shoes

Variable		Mean	± SD	SEM(%) (95% CI)	CV (%)	ICC	(95% CI)	CMC
Peroneus Longus								0.88
Peak (% of peak per gait cycle)	**Day 1**	87	7	4 (3-8)	5	0.44	(0-0.82)	
	Day 2	91	3					
Time of Peak (% of gait cycle)	**Day 1**	45	3	2 (1-3)	3	0.72	(0.20-0.92)	
	Day 2	44	2					
Abductor Hallucis								0.85
Peak (% of peak per gait cycle)	**Day 1**	74	20	14 (10-26)	20	0.63	(0.05-0.89)	
	Day 2	70	24					
Time of Peak (% of gait cycle)	**Day 1**	57	8	8 (5-14)	14	-0.03	(0-0.58)	
	Day 2	53	7					

Table 5-6 Discrete EMG variables for peroneus longus and abductor hallucis barefoot

Variable		Mean	± SD	SEM(%) (95% CI)	CV (%)	ICC	(95% CI)	CMC
Peroneus Longus								0.91
Peak (% of peak per gait cycle)	**Day 1**	90	4	3 (2-5)	3	0.23	(-0.33-0.67)	
	Day 2	92	3					
Time of Peak (% of gait cycle)	**Day 1**	44	2	1 (1-2)	2	0.85	(0.52-0.96)	
	Day 2	44	2					
Abductor Hallucis								0.79
Peak (% of peak per gait cycle)	**Day 1**	65	27	21 (15-38)	30	0.24	(-0.42-0.74)	
	Day 2	76	21					
Time of Peak (% of gait cycle)	**Day 1**	49	5	4 (3-7)	8	0.63	(0.04-0.89)	
	Day 2	47	7					

Table 5-7. Discrete kinematic and kinetic variables in shoes

		Day 1			Day 2						
	Variable	**Mean**	**± SD**		**Mean**	**± SD**		**SEM (95% CI)**	**CV (%)**	**ICC**	**(95% CI)**
n=9	MaxEv (°)	-3.6	2.4		-3.6	2.2		1.59 (1.14-2.71)	44	0.60	(0.07-0.86)
n=9	ROM (°)	11.1	3.3		9.6	2.7		2.98 (2.14-5.10)	29	0.03	(0-0.57)
n=9	MaxES (°)	7.4	4.2		5.9	2.6		3.12 (2.24-5.33)	47	0.23	(0-0.69)
n=10	MaxMEv (Nm/kg)	-0.12	0.06		-0.12	0.04		0.04 (0.03-0.06)	32	0.56	(0.05-0.84)
n=10	MaxMInv (Nm/kg)	0.13	0.10		0.18	0.12		0.06 (0.04-0.10)	39	0.75	(0.37-0.91)

5.4 Discussion

5.4.1 *Electromyography*

The first aim of this study was to determine the reliability of PL and ABH in walking. The second aim was to determine the between-day reliability of MLA angle using a foot-shoe model. The final aim was to establish reference values of the between-day reliability of TA and MG EMG and discrete kinematic and kinetic variables.

5.4.1.1 *Peroneus Longus*

It was found that activation of PL can be confidently studied in a repeated-measures study design like an intervention. The between-day reliability of discrete EMG variables in PL was more reliable in the present study than in the study by Barn et al. (2012). The SEM (95% CI) for peak amplitude in early stance was 19% (-53, 51) and 12% (-27, 40) for barefoot and shod respectively in the previous study. Low reliability in early contact might be expected in early stance if crosstalk with TA was present due to high TA activity in this phase. Considering the peak in midstance/propulsion, when PL is most active, the SEM for peak amplitude was 17% and 18% for barefoot and shod respectively in the previous study, whereas only 3% and 4% in the present study. The SEM of the time of peak in midstance/propulsion was 8% and 9% in the previous study and 1% and 2% in the present study for barefoot and shod respectively.

The use of ultrasound guidance and small EMG sensors means there was less potential for crosstalk in the present study than there would have been with larger sensors, such as the standard surface Delsys Trigno™ (Delsys, Inc., Boston, USA) used previously (Barn et al., 2012). The potential for crosstalk in surface PL EMG recordings has been demonstrated, with large variability of the PL surface EMG signal envelope with electrode location during walking (Campanini et al., 2007). A two-dimensional grid of 4 X 3 EMG electrodes over the PL was used to compare the bipolar signal from the centre of the grid to the signal 2 cm medially towards the tibialis anterior (TA) and 2 cm laterally towards the lateral gastrocnemius. Electrode location had on average only a moderate effect on peak PL activation (10% ± 3% relative to the central location), however the maximum relative change in the peak was 29% ± 13%. The majority of participants showed moderate variability in the timing of peak activation with a shift in recording electrode location towards TA, with a mean difference of 8% ± 4% of the gait cycle, ranging from an average of 1-12% per individual. In extreme cases a change in peak amplitude and timing of peak amplitude of PL could be explained by crosstalk with TA due to the presence of a high level of activation in early stance when PL is typically relatively inactive and TA is very active. Erroneous recording TA activity in early stance would inevitably lead to an incorrect estimation of onset of PL activity, as there would likely be greater amplitude over the baseline period. The area of the linear envelope after normalisation was most effected by electrode location, with a mean relative change of 29% ± 10% and a maximum change of 73% ± 40%. The crosstalk index used in this study was not designed to quantify crosstalk between muscles, but to provide a means to explain contributions to signal variability with electrode placement. The crosstalk index used was defined as "the ratio between the area of the normalized envelope outside the expected activation phase of the muscle and the area of the entire normalized envelope." The crosstalk index for PL increased from the lateral to the medial column of the electrode grid, towards TA.

The greater between-day reliability in midstance/propulsion in the present study compared to the study by Barn et al. (2012) may be due to differences in power, study populations, and/or differences in normalisation. Barn et al. (2012) acknowledged the limitation of a sample size of five, due to the challenge of recruiting individuals with RA. In addition, the sample in the earlier study was older than in this study (53 ±9 years vs. 28 ±4 years respectively), which could mean there were differences in muscle size and/or function due to ageing. Barn et al. (2012) also suggested that RA may have induced added between-day variability in both raw and normalised EMG signals. Fluctuations in inflammatory cytokines can influence muscle function and subsequent EMG recordings and variability in MVCs may have been an issue as differences in joint tenderness between days may affect the capacity to successfully perform an MVC (Barn et al., 2012).

For barefoot walking, reliability in the present study was better for some variables and worse for others compared to the study by Murley et al. (2010b). The ICC (3,1) for peak PL amplitude in barefoot walking in the present study 0.23 (-0.33-0.67) was worse than the ICC (2,1) of peak PL amplitude in midstance during barefoot walking in the previous study: 0.53 (0.17–0.77) and 0.52 (0.15–0.76) for MVC and sub-maximum normalisation respectively (Murley et al., 2010b). However, in the study by Murley et al. (2010b) there was a systematic bias (% mean difference) of 7.7% between sessions for midstance peak amplitude when using sub-maximum normalisation. The difference in mean of peak amplitude in the present study in the barefoot condition was only 2% and the SEM 3% (2-5) and CV (3%) were low despite poor reliability according to the ICC. The ICC (3,1) of 0.85 (0.52-0.96) in the present study for time of peak amplitude barefoot was superior to time of peak in midstance in the earlier study ICC (2,1): 0.58 (0.27–0.78). Differences between the present study and that by Murley et al. (2010b) may be due to the differences between fine-wire and surface EMG. Fine-wire EMG is sensitive to the positioning of the recording tip of the electrodes with respect to active motor units (Kadaba et al., 1985). With slight variation in electrode position, the number of active motor units recorded from that contribute to the EMG signal could vary between sessions, impacting on between-day reliability. Nevertheless fine-wire electrodes are able to collect from a substantial number of motor units and repeatability is similar in soleus, TA and medial gastrocnemius for surface and fine-wire EMG sensors (Bogey et al., 2000, Onmanee, 2016). However it is unknown if these results apply to PL because motor units could be distributed differently in this muscle and if a different number of active motor units are recorded from between sessions then this could impact on between-day reliability (Onmanee, 2016). Nonetheless the present protocol has the advantage of being less invasive than fine-wire EMG.

In conclusion, the PL EMG profile was highly repeatable between sessions during barefoot and shod walking. Good reliability was also demonstrated for the majority of statistical measures of reliability for peak PL amplitude and time of peak amplitude. Therefore activation of PL can be confidently studied in a repeated-measures study design like an intervention using this protocol.

5.4.1.2 *Abductor Hallucis*

The use of discrete variables for ABH requires caution depending upon the precise nature of the research question. Study of ABH activation over time requires more consideration than that of PL, especially as there was large variability in the profile of ABH EMG between participants, which was similar to previous work in straight line walking (Reeser et al., 1983). The overall pattern of ABH activation was more consistent between days in the shoe only and barefoot conditions, so it may be more appropriate to study the effect of FO on ABH qualitatively by visual inspection of the RMS signal over the gait cycle than with discrete variables.

There has been little study of the activation pattern of ABH during walking since early work that indicated that ABH is active from mid-stance to toe-off (Reeser et al., 1983). However more recently ABH has been shown to be active after initial contact in walking (Kelly et al., 2015). In the present study peaks of ABH activity was seen in late swing prior to contact into early stance and again in late stance prior to toe-off. Previous studies on ABH EMG have used fine-wire rather than surface electrodes. As surface EMG electrodes can be more prone to movement artefact than fine-wire electrodes, the high amplitude in early stance in this study could have been interpreted as an artefact. The impact at heel strike could have led to an artefact, which if this was the case could explain why activation in early stance was not seen in some individuals while barefoot, if they contacted the ground with a flatter foot to protect their heel. However, the signal was still present after experimenting with a more severe filter, so it seems less likely that the signal was due to artefact from contact of the sensor with the shoe. The pattern of ABH activation in late swing into early stance is similar to the pattern of TA activation. It could be that as ABH can flex the hallux that the muscle can act eccentrically at foot contact to limit flexion of the hallux when the TA is dorsiflexing the foot to enable toe clearance.

It is suspected that the bump in the mean EMG ABH profile seen in some cases after 60% of stance (after toe-off) in shod conditions is in fact an artefact, due to contact with the shoe as it was not seen barefoot. For one participant who had narrow feet it was noted that the shoe had to be pulled particularly tight for them and they showed the clearest example of this artefact. The relative reliability of ABH EMG was poor, when considering how the difference between FO conditions and shoe only varied between sessions. With more material in the arch with the FO conditions it is likely there was more interference with the sensor than in the shoe condition and especially the barefoot condition. A potential solution to reduce the artefact after toe-off would be to cut a hole in the shoe for the mini sensor head. However, as a number of holes were already made in the shoe for kinematic markers to calculate MLA, the side of the shoe might lose its integrity if further holes were made it in and the shoe would be less supportive. However, knowing the limitation of using a surface sensor for ABH EMG in shod walking, future work could only consider the peak prior to toe-off and neglect the second bump in early swing. The response of ABH EMG to FO was still recorded in the subsequent immediate effects study with this consideration of shoe/FO interference with the sensor in mind.

5.4.1.3 *Medial gastrocnemius*

The double bump pattern in MG EMG profiles in the barefoot conditions was a surprise considering normative data for MG EMG (Onmanee, 2016). The pattern was seen more in the barefoot condition than shod. Normative or control data for MG EMG has often been obtained from walking in shoes (Forghany et al., 2014, Hof et al., 2002, Nymark et al., 2005, Scott et al., 2012, Warren et al., 2004, Winter and Yack, 1987), which may partly explain why the double pattern shape is not considered typical. In studies that have recorded EMG of MG when walking barefoot (Murley et al., 2014a, Murley et al., 2009a, Scott et al., 2012), it might be that the double bump shape has been masked by averaging data across the sample. The double pattern shape was not observed in all participants in this study. If fewer individuals exhibited the pattern in the earlier studies then the pattern could have been lost when averaging across all participants, resulting in the standard rising ramp seen in shod walking.

5.4.2 *Kinematics and kinetics*

The second aim of this study was to determine the between-day reliability of MLA angle using a foot-shoe model.

5.4.2.1 *Medial longitudinal arch*

The pattern of MLA angle was consistent across stance and the ROM was consistent in most participants for whom there was MLA data, with a MDD of 4.2°. The angle was greater in barefoot than shod in both sessions. The barefoot condition followed a different pattern to the shoe and orthoses conditions in that the angle dropped steeply around 80% of stance, prior to toe off. This could reflect a greater stiffness in the shod conditions than the barefoot and so the foot maintains a flatter position when shod. The drop in MLA angle in late stance barefoot is similar to the pattern in an earlier study of multi-segment foot kinematics barefoot (Leardini et al., 2007). The mean MLA angle was around 170° in the earlier study as supposed to between 140-150° in the present study (**Figure 5-11**). Leardini and colleagues placed the first metatarsal marker on top of the metatarsal head, rather than on the side of the metatarsal head as in this study, which may explain why the angle was more flat, closer to 180° in the earlier study. The mean MLA angle during walking obtained from fluoroscopy was found to be 131.5° ±11.8° across foot types (n=15) (Balsdon et al., 2016). In the study by Balsdon et al there was a significantly greater dynamic MLA angle in pes planus feet vs. normal feet (142.2° ±4.7° vs. 133.2° ±7.7°, $p<0.05$), although there were only five participants in each group. Foot posture was not recorded for the purposes of this study, but in light of the variability of MLA angle with foot posture reported by Balsdon et al. the data in this study appear realistic.

5.4.2.2 *Kinematics and kinetics results*

Comparing the width of the standard deviation in the plots and the differences in the mean traces for day one and day, it can be seen that there was greater variability between and within session in the rearfoot angle in the frontal plane than the plantar flexion angle. The variability in kinematics observed was to be expected given that sagittal plane kinematics are typically more reliable than frontal plane kinematics (McGinley et al., 2009). Moment data is also typically more variable than kinematics (Pinzone et al., 2016). Although some discrete variables had poor reliability statistics, the difference in the mean between sessions was generally small and likely not meaningful (at most a difference of around 1.5°). Standard deviations were large, this could be improved upon with a larger sample.

5.4.2.3 *Limitations*

The walking speed was controlled only within session. The study design could have been improved by keeping the speed constant between sessions. However, good reliability of PL EMG was achieved without this consideration and this reflects the pragmatic nature of studies on any specific day and the natural variation between days that should be expected. In addition, the difference in stride time between days was within one standard deviation of the mean of each day, suggesting that overall walking speed did not differ greatly between the two days.

5.5 Conclusion

The PL EMG profile and discrete variables were repeatable between sessions, supporting the use of this protocol for subsequent studies involving repeated measures, including investigating the immediate effects of FO. The reliability of the protocol for recording PL EMG developed in this chapter was accepted for publication (Reeves et al., 2019). In Chapter 6 the effect of FO on PL is studied by recording from PL EMG using the Delsys Trigno™ Mini sensor and prior ultrasound guidance to aid sensor placement. The general activation pattern of ABH was reliable, however individual variability was greater than in PL. In the immediate effects study, the effect of FO on ABH EMG will also be studied using the Delsys Trigno™ Mini sensor and ultrasound guidance, however only the response of the activation profile will be studied and not discrete variables, along with ROM of the MLA.

Chapter 6 IMMEDIATE EFFECT OF FOOT ORTHOSES ON EMG AND FOOT AND ANKLE BIOMECHANICS

6.1 Introduction

Foot orthoses alter external joint moments (Nester et al., 2003b, Sweeney, 2016, Telfer et al., 2013b), but less is known about the effect of FO on internal joint moments and further research into the effect of FO on neuromuscular factors is particularly warranted (Mills et al., 2009). This study explored the simultaneous effects of FO on muscle activity (using EMG) and foot and ankle biomechanics.

6.1.1 *Rationale*

There are several important reasons that justify this study. Firstly, studies on the effects of FO on EMG included in the literature review (Chapter 3) generally reported limited details on the FO materials and designs used. Therefore, there is a need to investigate if specific aspects of medial FO geometry, like heel wedging and arch height, have the potential to affect muscle activity differently. For instance, it is possible that as an increased medial arch height offloads the heel earlier, in early stance when the heel is on the ground, the onset and offset times may change and/or the duration of TP activity in early stance may decrease. Greater external support of MLA could mean less activation, in mid-late stance, is required of TP to actively support the arch. Conversely with a medial wedge, if the external eversion moment decreased in early stance and less force was borne by the TP tendon and less elastic energy was stored in the tendon, there would be less potential for energy return during active supination, while TP EMG activity could remain the same.

Secondly, previous literature frequently did not simultaneously collect kinematic and kinetic data when analysing the effect of FO on EMG. This means any change in the EMG data is difficult to explain with respect to the external forces applied to the joints that the muscles cross, nor the kinematic outcomes of all the forces being applied. Furthermore, if walking speed is kept constant, a change in EMG would not be expected if the FO were not successful in changing how load was applied to the sole of the foot, and this change in load is rarely reported.

Thirdly, few high quality studies have investigated TP EMG (Akuzawa et al., 2016, Maharaj et al., 2018, Murley et al., 2010a), despite it being one of the muscles most likely to respond to FO, as in theory a reduction in external ankle moment with FO would reduce the requirement of TP to generate a counter internal inversion moment. Indwelling EMG is necessary for investigating the activity of deep muscles like TP, but using fine-wire electrodes can be challenging, which limits the use of the technique in research (O'Connor et al., 2006, Semple et al., 2009, Stacoff et al., 2007). Neither kinematics nor kinetics were reported in two of the studies of the effect of FO on EMG which included TP (Akuzawa et al., 2016, Murley et al., 2010a) . In the further study by Maharaj et al. the custom made FO was compared to barefoot and footwear, but the insole material of the shoe likely had different material properties to the FO, which could confound the effect of changes in FO geometry (Maharaj et al., 2018). Three studies that used fine-wire EMG of the TP identified in the literature search were not included in the review because either the control condition was barefoot (Barn et al., 2014) or the FO conditions were worn in a sandal and not a shoe (Garbalosa et al., 2015, Stacoff et al., 2007). Therefore, there is a need to conduct a study to understand the specific effects of FO geometry on TP activity.

A further issue is that no study has investigated the effect of FO on ABH activity, despite its fundamental role in medial arch control. The lack of investigation can again be attributed to measurement difficulties, as standard surface electrodes would be too big to be positioned correctly to minimise crosstalk with other muscles and could be uncomfortable and interfere with walking. According to the reliability study, the small head of the Delsys Trigno™ Mini allows study of the overall profile of ABH in walking. Activity of ABH might decrease with FO with increased arch height due to increased external support of the arch, so less active support is needed, or because of increased compression of the muscle, making it less able to generate force.

Previous work in the department found a threshold effect for increased arch height with FO and changes in plantar pressure and tissue thickness (Sweeney, 2016). With respect to the standard Salfordinsole, medial midfoot peak pressure increased by 41.1 % ($p < 0.001$), 49.5 % ($p < 0.001$) and 44.8 % ($p < 0.001$) for standard -6mm, standard and standard +6mm arch heights respectively. Additionally, altering arch geometry with a semi-custom FO was shown to not change kinematics, but did reduce plantar fascia strain (Ferber and Benson, 2011). From the reliability study PL does not appear to be active in early stance, so if the FO was expected to act in early stance when the heel is on the ground, as with a medial heel wedge, a change in PL EMG would not be expected. However, FO with medial arch support may have effects on PL EMG in mid-late stance. Consequently there is a need to explore the effects of different FO geometry on PL EMG.

6.1.2 *Research question*

What are the immediate effects of variations in FO geometry on EMG of select muscles of the lower limb and foot and ankle biomechanics?

6.1.3 *Aim*

To establish if medial wedging and increased medial arch height have effects on EMG of TP and other muscles of the lower limb, and foot and ankle moment/motion.

6.1.4 *Hypotheses*

A number of hypotheses were devised relating to changes in FO medial heel wedging and arch height. Arch height was increased from the flat inlay, to a standard Salfordinsole (20 mm arch height), to a further 6 mm increase in arch height with the Arch FO (total arch height 26 mm), and Arch & Wedge FO (8° extrinsic wedge combined with a 6 mm increase in arch height, total arch height 26 mm). Previous work found a ceiling effect of increasing arch height on plantar pressures, in which there was a large increase in peak pressure in the medial arch from flat control to a standard Salfordinsole, however a further increase in arch height had little additional effect on plantar pressure (Sweeney, 2016). As it is expected that changes in kinematics, kinetics and EMG will be driven by changes in load applied to the foot, it was expected that the changes in biomechanical variables as a result of increased arch height in Hypothesis 2- arch geometry will not be a systematic, dose response.

6.1.4.1 *Hypothesis 1- medial heel wedging*

It was hypothesised that increases in medial heel wedging would lead to the following changes relative to the flat inlay:

1.1 Reduced TP EMG peak amplitude in early stance.

1.2 No change in EMG of MG, PL or TA.

1.3 Decreased EMG of ABH for Arch & Wedge in mid-late stance.

1.4 Reduced external eversion moment across stance (more inversion).

1.5 Reduced peak external eversion moment (MaxMEv).

1.6 Increased peak external inversion moment (MaxMInv).

1.7 Increased inversion at foot contact (MaxES), reduced peak rearfoot eversion angle (MaxEv) and reduced rearfoot eversion range of motion (ROMEv).

6.1.4.2 *Hypothesis 2- arch geometry*

With a standard Salfordinsole FO and further increase in arch height the following *non-systematic* changes were hypothesised relative to the flat inlay condition:

2.1 Earlier onset of TP activity in early stance.

2.2 Reduced TP EMG peak amplitude in late stance.

2.3 No change in muscle activity of MG, PL or TA.

2.4 Decreased muscle activity in mid-late stance of ABH.

2.5 Reduced external eversion moment across stance (more inversion).

2.6 Reduced peak external eversion moment (MaxMEv).

2.7 For the Salfordinsole and FO with only a 6 mm increase in arch height (Arch), but not the Arch & Wedge: no effect on rearfoot kinematics.

2.8 Decreased range of motion of MLA.

6.2 Methods

6.2.1 *Participants*

Previous work by Murley et al. (2010a) found an effect size (ES) of 0.59 for the difference in TP peak EMG amplitude shod vs. shod with prefabricated FO during early stance and an ES of 0.56 for the difference in PL peak EMG amplitude shod vs. shod with prefabricated FO during midstance. A power analysis was performed using G*Power3.1 (Faul et al., 2007). Using a dependent t-test, with >80% power and a 0.05 two−sided significance level, a total sample size of 25- 28 was calculated for an ES of 0.59 and 0.56 respectively. Participants were recruited between 18-60 years of age.

6.2.2 *Foot posture index (FPI)*

Participants were screened for foot type classification using the FPI. Participants were excluded if they had an extreme cavus foot type, as it was thought that the FO would not likely be in contact with the medial arch in this foot type. Additionally, individuals with a cavus foot are less likely to be prescribed medial FO in clinical practice.

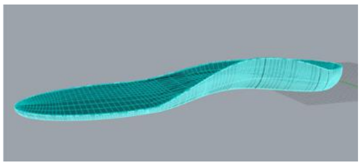
6.2.3 *Design and manufacture of FO*

Previous research within the department used both intrinsic medial heel wedges (measured in millimetres inside the heel cup) and extrinsic medial wedges (measured in degrees under the heel cup). The change in inversion moments at the ankle were similar between the two types of wedges, with large individual variation (Sweeney, 2016). Consequently, extrinsic wedges only were included in this study as they are more commonly used in clinical practice than intrinsic wedges. Rather than two magnitudes of arch height and two magnitudes of medial heel wedging as per the work by Sweeney (2016), a single extreme increase in arch height (6 mm above standard) and extreme increase in medial wedging (8°) were chosen as it has previously been shown that systematic changes in FO geometry result in systematic changes in joint kinematics, joint moments and plantar pressure, without an accompanied change in EMG (Telfer et al., 2013a, 2013b). Extreme changes were considered appropriate because if no change in EMG is observed with an extreme change in arch height or wedging, then it is unlikely that any change would be seen with a more minor change.

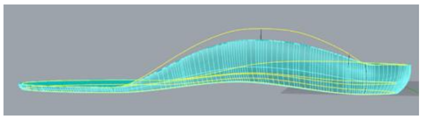
The FO were fabricated at the University of Salford from high density Ethylene-vinyl acetate (EVA, 50 Shore A). A computer-aided design/computer-aided manufacturing (CAD/CAM) system was used to design and manufacture all the insoles. The software used to design the FO was Inescop PAN, a plugin for Rhinoceros (Robert McNeel & Associates, Seattle, USA). This software works with a library of digital insole shapes based on the Salfordinsole (Salfordinsole Health Care Ltd, UK) morphology.

The FO designs used the standard Salfordinsole (**Figure 6-1.1, Figure 6-2**) as an initial template. Each condition is listed below:

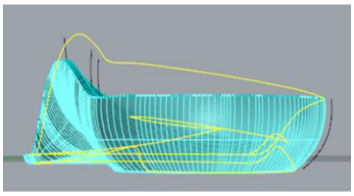
1. The standard Salfordinsole (20 mm arch height)



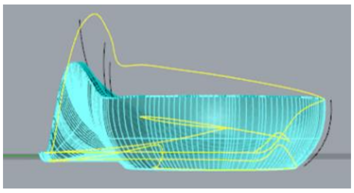
2. Salfordinsole with a 6 mm increase in arch height (26 mm arch height in total, Arch)



3. Salfordinsole with an additional 8° medial heel wedging (standard 20 mm arch height, Wedge)



4. Salfordinsole with both a 6 mm increase in arch height (26 mm arch height in total) and 8° medial heel wedging (Arch & Wedge)



5. A flat inlay (3 mm insole made from EVA, which was the same material as the FO conditions)

Figure 6-1. Experimental conditions (1-5). The yellow lines represent the border of the foot orthosis above the standard Salfordinsole.

The flat inlay had no heel or arch geometry and was used as the control condition to which the FO conditions were compared. The purpose of the flat inlay was to occupy around the same volume in the shoe under the forefoot as the FO conditions to separate the heel and arch modifications as independent variables.

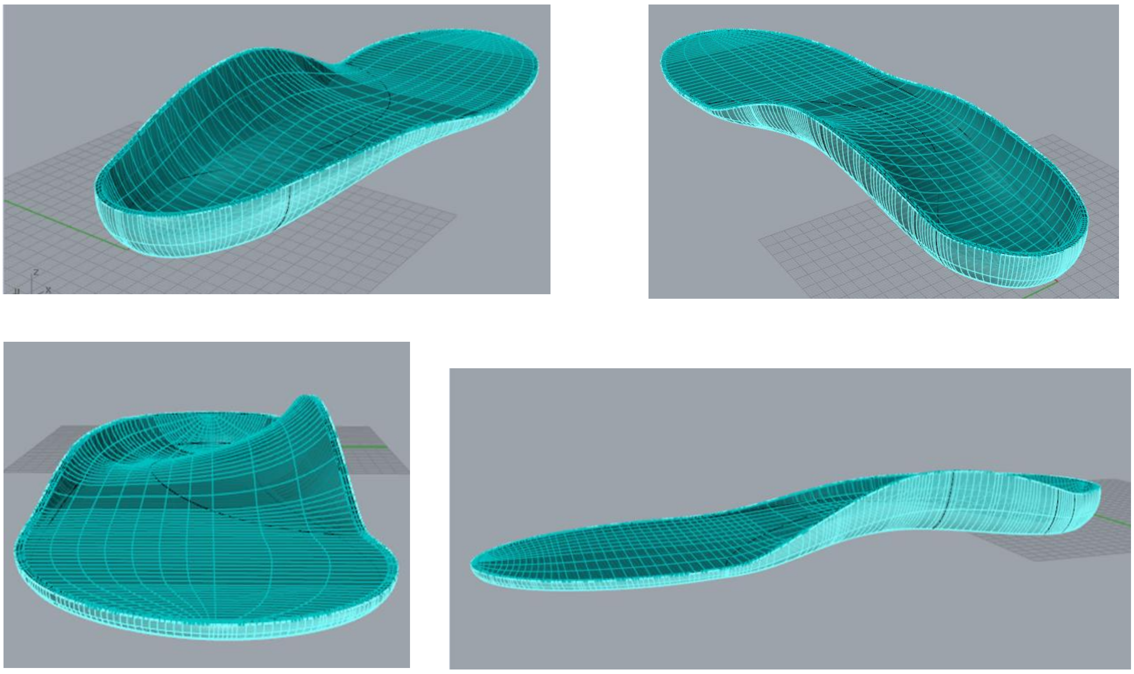


Figure 6-2. The standard Salfordinsole® viewed in Rhinoceros software

The FO files designed in the Inescop PAN software using the library template were imported into another Rhinoceros plugin (Machining PAN) that produced the files with the pathways for a milling machine to mill the FO out of blocks of EVA. Two files were produced by this software, one for the bottom of the FO and another one for the top.

6.2.4 *Protocol*

6.2.4.1 *Initial procedure*

Height, body mass, shoe size and FPI were recorded prior to data collection.

Initially the protocol began with the fine-wire insertion of TP as in Chapter 4, followed by EMG and marker set up, and data collection as per the reliability study in Chapter 5. There was however one difference compared to the marker set up in Chapter 5, in that the position of the fine-wire sensor meant it was not possible to bandage over a shank cluster, so the calibration markers (ankle and knee) were used for tracking of the shank. (**Figure 6-3**). The leg chosen for the study was the right leg, as it is most often the dominant leg and without any reason to believe the effect of the FO would differ between legs in healthy participants. However, markers were also placed on the left tibia and foot for the convenience of foot event recognition.

Figure 6-3. Experimental setup

Once set up was completed participants were presented with the first of the FO conditions, which occurred in a randomised order. To help the participants get used to the FO condition they were allowed a few minutes to walk around in the shoes and FO before data was collected. Walks were recorded over a 6-meter walkway in a university gait laboratory, with data recorded in both walking directions. A minimum of six good gait cycles per condition were used in the data analysis.

6.2.4.2 *Problem with initial data*

After initial processing of TP EMG data for six participants (in Visual3D), it became apparent that there was a degradation in TP amplitude over time, irrespective of condition, thus interpreted as measurement error. The data for these participants is presented in the appendix (**Figure 0-2** and **Figure 0-3**). A problem of signal degradation was noted despite the FO order being randomised, and typically the first FO condition had the highest amplitude (in four of the six cases).

It was suspected that the conductivity of the electrode tips altered with exposure to biological processes within the muscle, some localised oedema around the electrode for example. It appears that with these prefabricated electrodes there was approximately 30-40 minutes of walking time from which to obtain quality data. It was unclear how much the signal degradation problem is influenced by muscle contraction and how much the recording properties would be influenced simply by time in the body.

After the signal attenuation problem was identified, data collection for the next participant was conducted with time efficiency in mind. The insertion of the fine-wire electrode was performed after as much other set up was complete and less walking trials were performed. However, the signal degradation problem was still evident, as in **Figure 0-4** of the appendix, where a clear decrease in amplitude can be seen with each condition in the order they were worn (data from condition five, the Wedge, was discarded as the signal had clearly been lost). The participant was identified as an outlier and excluded from further analysis.

Based on these observations, further work sought to characterise the phenomenon of signal degradation with fine-wire electrodes over time. **Figure 0-5** of the appendix (legend in **Table 0-1**) shows TP EMG of one participant walking for an hour in a single shod condition with a clear progressive drop in the signal. An additional project aimed to characterise the phenomenon of signal degradation with fine-wire electrodes over time in comparison to surface EMG in TA. Initial results are presented in the form of a conference abstract in the appendix (**Tibialis anterior EMG over time with fine-wire and surface electrodes**). Future work could make comparisons regarding the amount of uninsulated recording tip exposed and contraction type. For the purpose of the study in this chapter on the immediate effects of FO on EMG, the protocol was altered to accommodate the signal attenuation issue.

6.2.4.3 *Revised protocol*

As multiple gait cycles can be recorded during a single walking trial, the protocol was revised so that only three walking trials were recorded per condition for TP EMG. A single walk in the flat inlay condition was performed before the randomised conditions, so that there would always be some control data if there was a problem with the signal in later trials. However, this first flat inlay trial was not used in the flat inlay average, because otherwise the conditions could not be considered truly randomised. The average time from start to finish of the trials used for EMG analysis was 25 minutes and no longer than 30 minutes.

Within each walking trial there was one foot contact with the force plate, however additional gait cycles could be defined using gait events from the kinematics. If foot contact with the force plate was partially or entirely missed, the participant was not asked to repeat the trial, in the interest of obtaining the TP EMG data as quickly as possible. When it was not possible to obtain gait events from the kinematics/kinetics, the filtered acceleration signal from the ABH sensor (closest to the ground) was used to define right foot contact (**Figure 6-4**). A bi-directional Butterworth filter with a 10 Hz frequency cut-off was used to filter accelerometer data (X) in Visual3D based on recommendations in a pipeline in EMG works (Delsys, Inc.). The threshold for foot contact with the acceleration data was based on the magnitude of the acceleration from another successfully defined foot contact from another gait cycle and/or trial.

Once EMG data was obtained from all FO conditions, the fine-wire electrode was removed. The participant then performed a minimum of a further three walking trials in each FO condition to obtain a sufficient number of force plate contacts for the kinetic calculations.

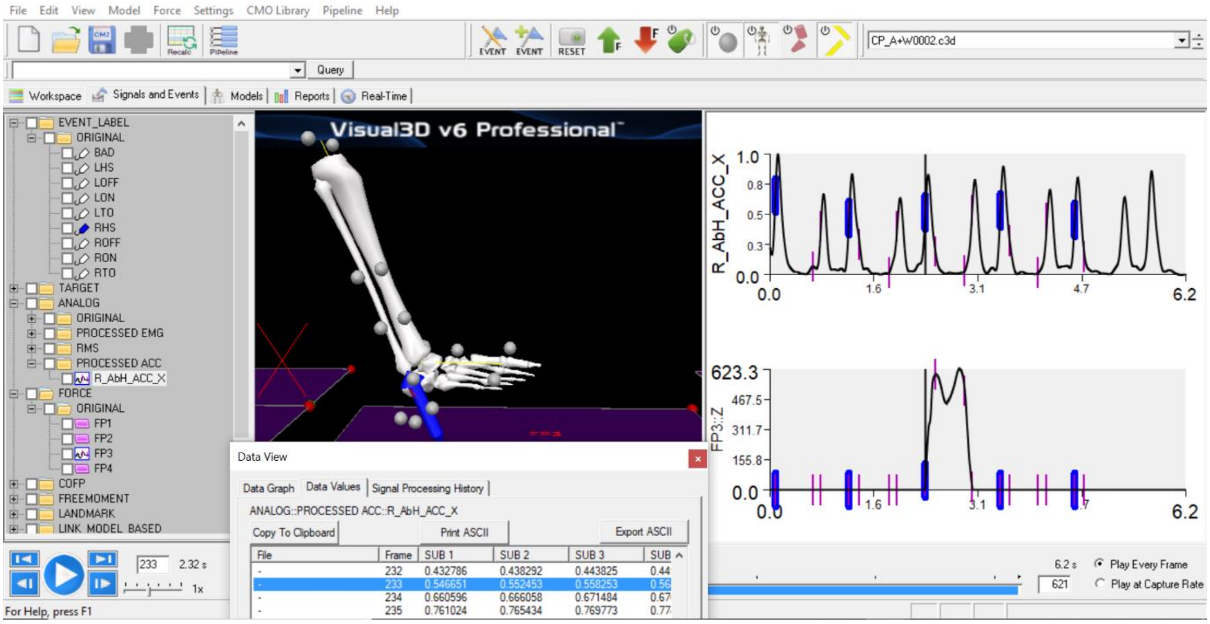


Figure 6-4. Establishing a threshold for foot contact from filtered acceleration data. Showing acceleration data, force data, foot and shank segment and acceleration values.

6.2.5 *Analysis*

6.2.5.1 *Onset of muscle activity*

Onset of the first peak of TP activity was determined for each gait cycle and then a median taken per participant for group analysis. A custom MATLAB function was written by Dr. Derek P. Zwambag for determination of onset of EMG activity using a threshold method. A baseline period was visually determined and was approximately 10-20% of the gait cycle in length in late swing, when TP is typically inactive (**Figure 6-5**). The threshold value was defined as the mean of the baseline values + (2* baseline standard deviation). Onset values were then made relative to foot contact so that zero represents foot contact and negative values indicate onset prior to foot contact.

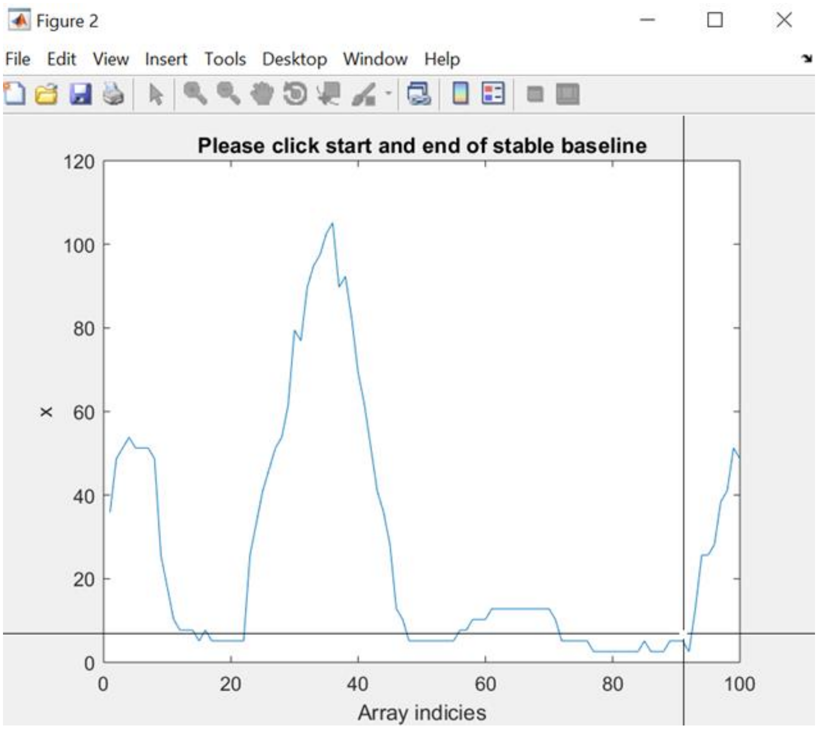


Figure 6-5. Defining the baseline period of tibialis posterior EMG in late swing using the graphical user interface of the MATLAB function for determining EMG onset time (Zwambag 2018)

6.2.5.2 *Statistics*

Statistical analysis was performed with SPSS (IBM SPSS Statistics 25) and Excel (Microsoft Office Excel 2013). A visual inspection of the histograms and box plots of data was used to establish normality and identify outliers. Outliers were determined based on the interquartile range (IQR). If a value was beyond the first quartile -3*IQR or the third quartile +3*IQR it was determined to be an outlier and the participant was excluded for that variable.

A one-way repeated measures ANOVA was performed on discrete variables unless there was a severe departure from normality, in which case non-parametric alternatives were considered. The alpha significance level was set at 0.05 (two tailed). A Bonferroni post hoc analysis and inspection of the means was carried out in the event a significant main effect was identified. A two-way paired ANOVA was performed on surface EMG peak amplitude, with factors FO condition and fine-wire (presence and absence of the electrode).

6.3 Results

6.3.1 *Participant characteristics*

The lower quartile (Q1), upper quartile (Q3) and interquartile range (IQR) were used to define outliers. Four participants were identified as mild outliers (Q1 - 1.5 * IQR or Q3 + 1.5 * IQR) based on the peak of their TP EMG and excluded from all analysis TP EMG analysis and TP EMG was not recorded from one participant. Data for all variables was not available for every participants due to measurement issues, the number of participants for each study variable, after outliers were removed, is presented in **Table 6-1**. Characteristics of all 24 participants are presented in **Table 6-2**, characteristics of participants for which TP data was collected and after outliers removed are presented in **Table 6-3**.

Table 6-1. Total number of participants (n) for each study variable after outliers were removed

Variable	n
EMG	
TP	19
TP no discomfort	15
TP discomfort	4
MG, PL, TA	19
ABH	9
Kinematics	
MaxEv, ROMEv, MaxES	16
Plantar flexion angle	17
MLA	9
Kinetics	17
MaxMEv, MaxMInv	17
Plantar flexion moment	17

TP= tibialis posterior, MG= medial gastrocnemius, PL= peroneus longus, TA= tibialis anterior, ABH= abductor hallucis, MaxEv=peak rearfoot eversion, ROMEV=eversion range of motion, MaxES= inversion at foot contact MLA= medial longitudinal arch, MaxMEv= peak external eversion moment, MaxMInv= peak external inversion moment. No discomfort and discomfort refers to participant perception of the presence of the fine-wire electrode in TP.

Table 6-2. Participant characteristics for all participants (n=24)

n=24	Mean	±SD	Median	Range
Age (years)	32	9	30	(18-56)
Height (m)	1.71	0.08	1.69	(1.58-1.86)
Mass (kg)	72	12	72	(58-105)
Shoe size (UK)	8	2	8	(5-11)
FPI (average of two feet)	2	2	2	(-3- 6)

Table 6-3. Participant characteristics for included participants, outliers removed (n=19)

n=19	Mean	±SD	Median	Range
Age (years)	31	7	30	(18-43)
Height (m)	1.71	0.08	1.69	(1.58-1.86)
Mass (kg)	74	12	72	(58-105)
Shoe size (UK)	8	2	8	(5-11)
FPI (average of two feet)	2	2	3	(-3- 5.5)

6.3.2 *Walking speed*

Mean ±SD walking speed for trials with and without the fine-wire electrode present are presented in (**Table 6-4**). Walking speed was significantly slower in the trials without the fine-wire electrode than with the fine-wire electrode ($p<0.05$).

6.3.3 <u>*EMG*</u>

6.3.3.1 *Fine-wire EMG*

Tibialis posterior EMG is presented in **Figure 6-7a**, and TP EMG separated into participants who experienced no discomfort from the fine-wire electrode in **Figure 6-7c** and those who found the electrode uncomfortable in **Figure 6-7d**. In **Table 6-5** peak TP EMG is presented as means ± SD for early stance and median ± IQR for late stance, because peak in late stance was not normally distributed.

There was reduced peak TP EMG in early stance with all FO, with small to very large effect sizes (0.31-1.1.5, **Table 6-5**). With respect to the flat inlay, peak TP EMG in early stance reduced by 17% (p=0.001) for the Wedge and by 14% (p=0.047) for the Arch & Wedge. Although there were some small to moderate reduction in the peak in late stance, there was large variability in the data and there was no significant effect of FO (p=0.164). As only four participants experienced discomfort with the fine-wire electrode there were not large enough numbers in each group to compare peak TP EMG with respect to comfort. There was no effect of FO on onset of the peak of TP activation in early stance (p=0.277, ηp^2=0.068).

6.3.3.2 *Surface EMG*

Surface EMG for PL, MG and TA are presented in **Figure 6-6**. There appeared to be an increase in peak PL EMG with FO, however the response appeared different with the presence of the fine-wire electrode in TP, see **Figure 6-6a** compared to without the electrode in **Figure 6-6b**. In the trials with fine-wire PL EMG activity was greater than in the trials without fine-wire (p=0.034), which could have been due to the faster walking speed in the trial with fine-wire (**Table 6-4**). The mean differences between fine-wire and no fine-wire trials were mean ±SD: 9% ±27%, 10% ±20%, 16% ±18% and 6% ±26% for the Salfordinsole, Arch, Wedge and Arch & Wedge conditions respectively. Although the variability in the difference between fine-wire and no fine-wire trials was considerable, for most FO conditions the difference between fine-wire and no fine-wire trials was greater than 8%, which was considered the minimal detectable difference, as it is twice the SEM of 4% in the reliability study in Chapter 5 (Hopkins, 2000). However in comparing the effect of condition (FO) and the presence or absence of the fine-wire electrode, there was <u>no main effect of condition</u> (p=0.191, ηp^2=0.099) and no interaction effect (p=0.229, ηp^2=0.088) for peak PL EMG, so the response of PL activity to FO did not depend on the presence of the fine-wire electrode.

For the timing of peak PL EMG, there was a trend towards earlier activation in the fine-wire trials than no fine-wire trials, which approached significance (p=0.054, ηp^2=0.226) and was around 1%, approximately 43% in fine-wire trials and 44% in no fine-wire trials. A difference of 1% is less than the SEM of 2% found in the reliability study in Chapter 5. There was no main effect of condition on timing of peak PL EMG (p=0.791, ηp^2=0.021), nor an interaction effect between the presence of fine-wire and condition (p=0.515, ηp^2=0.029).

For peak MG EMG activity there was no effect of condition (p=0.234, ηp^2=0.134), presence of the fine-wire electrode (p=0.614, ηp^2=0.026), nor an interaction effect (p=0.355, ηp^2=0.102). Likewise, for the timing of peak MG EMG activity there was no effect of condition (p=0.248, ηp^2=0.131), presence of the fine-wire electrode (p=0.852, ηp^2=0.004), nor an interaction effect (p=0.590, ηp^2=0.041). For peak TA EMG activity there was no effect of condition (p=0.153, ηp^2=0.115), presence of the fine-wire electrode (p=0.410, ηp^2=0.046), nor an interaction effect (p=0.094, ηp^2=0.141). Likewise, for the timing of peak TA EMG activity there was no effect of condition (p=0.421, ηp^2=0.059), presence of the fine-wire electrode (p=0.743, ηp^2=0.007), nor an interaction effect (p=0.639, ηp^2=0.035).

There was limited quality date for ABH EMG, likely due to contact of the sensor head with FO and losing contact with the skin over multiple trials. In **Figure 6-7b** ABH EMG data is presented that includes some participants that were either excluded from other analysis based on peak TP EMG or for which TP EMG was not collected, for the sake of increasing the amount of available data. Discrete variables were not calculated for ABH EMG as they were found not to be reliable in Chapter 5.

In one participant excluded as an outlier based on their peak TP EMG activity, there was a trend for a reduction in TA activity with FO, seen in **Figure 6-8a** for fine-wire trials and **Figure 6-8b** for no fine-wire trials. In **Figure 6-8b**, TP data is presented for the same excluded participant and in **Figure 6-8d** it can be seen that the trend for the response in ankle inversion/eversion moment for the excluded participant follows the same pattern as the mean response (**Figure 6-12**).

Table 6-4. Mean ± SD walking speed (m/s) per condition for trials with and without the fine-wire electrode present

Speed (m/s)	Flat inlay	Salfordinsole	Arch	Wedge	Arch & Wedge
Fine-wire					
Mean ± SD	0.99 ±0.16	1.01 ±0.14	1.00 ±0.15	0.99 ±0.17	1.01 ±0.14
No fine-wire					
Mean ± SD	0.92 ±0.12	0.85 ±0.14	0.88 ±0.16	0.87 ±0.17	0.86 ±0.15

Table 6-5. Mean ±SD (ES) of peak TP EMG in early stance and median ±IQR (ES) of peak TP EMG in late stance across conditions, expressed as a percentage of the peak of the flat condition average

Normalised EMG (%)	Flat inlay	Salfordinsole	Arch	Wedge	Arch & Wedge
Mean ±SD (ES)	89 ±12	83 ±25 (0.31)	78 ±23 (0.61)	72 ±17* (1.15)	75 ±21* (0.85)
Median ±IQR (ES)	100 ±26	77 ±40 (0.20)	65 ±56 (0.44)	75 ±52 (0.17)	63 ±57 (0.27)

(* p<0.05 with respect to flat inlay), ES= effect size with respect to flat inlay

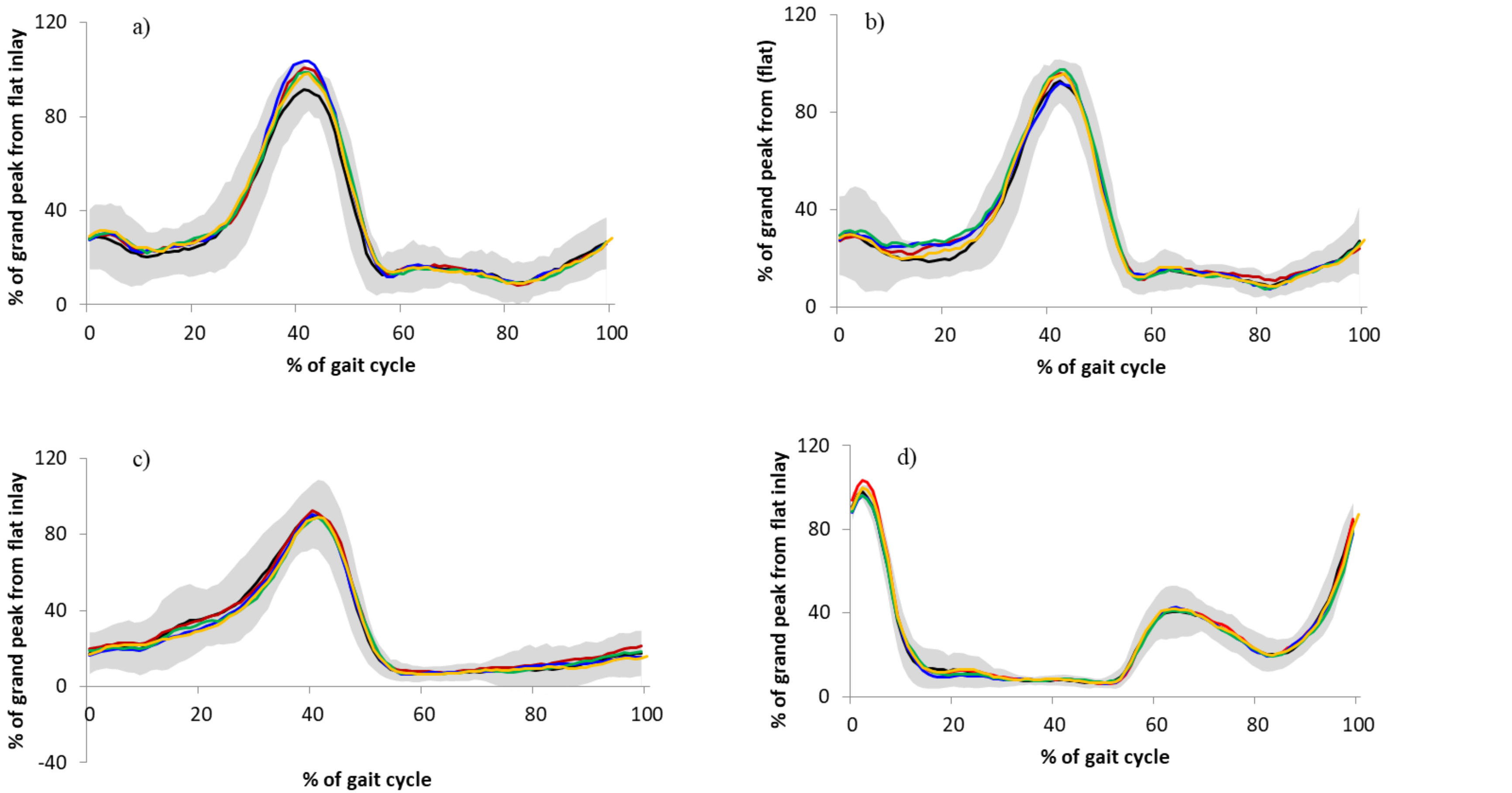


Figure 6-6. Peroneus longus EMG for fine-wire trials (n=19); b) Peroneus longus EMG for trials without fine-wire; c) Medial gastrocnemius EMG; d) Tibialis anterior EMG fine-wire trials (n=19). Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging. The grey shaded area represents the standard deviation of the flat inlay.

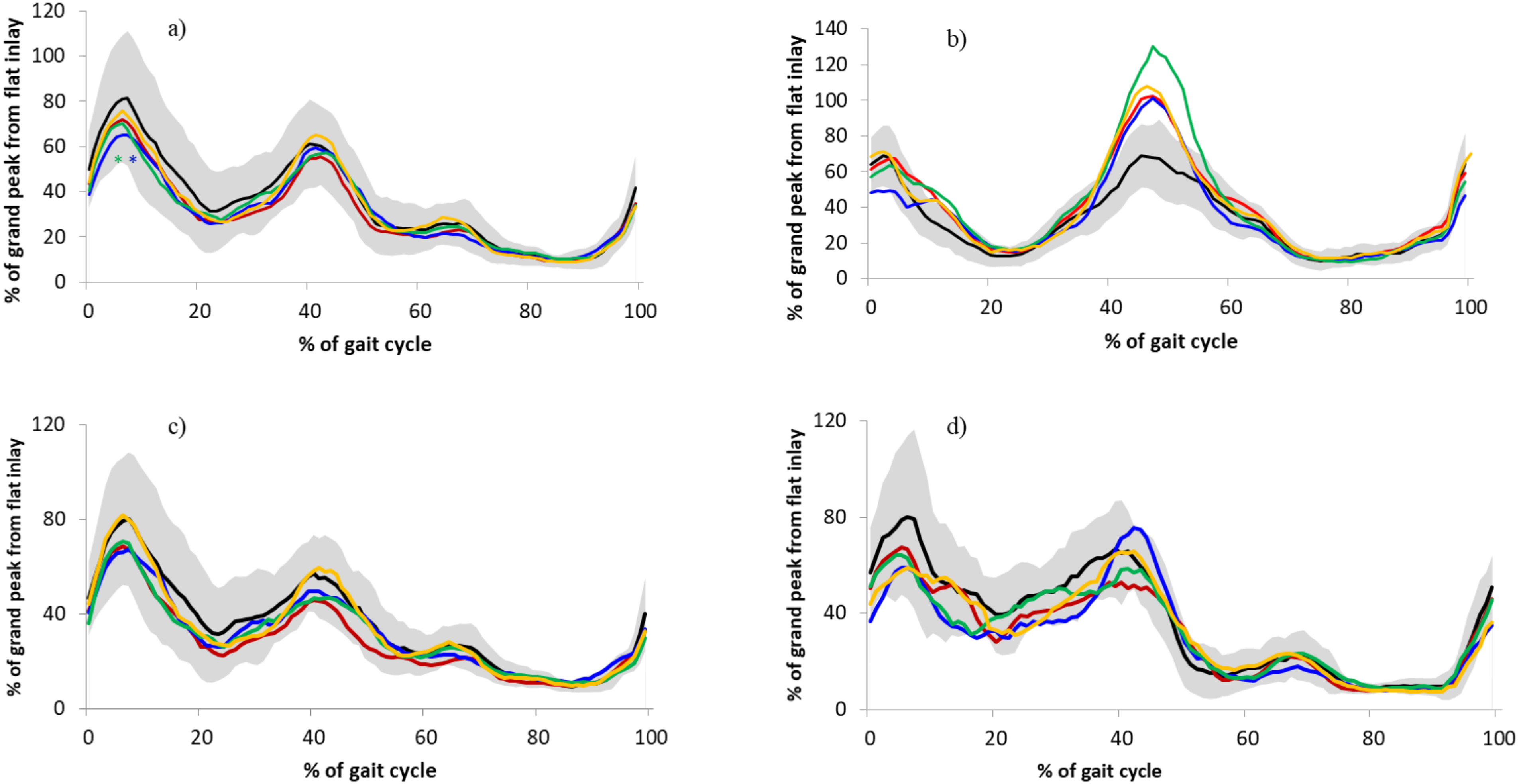


Figure 6-7. a) Tibialis posterior EMG (n=19); b) Abductor Hallucis EMG (n=9); c) Tibialis posterior EMG for participants who did not experience discomfort from fine-wire electrodes (n=15); d) Tibialis posterior EMG for participants who experienced discomfort from fine-wire electrodes (n=4). Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines and asterisk: 8 mm increase in medial wedging; green lines and asterisk: 6 mm increase in arch height combined with 8 mm increase in medial wedging. *p<0.05. The grey shaded area represents the standard deviation of the flat inlay.

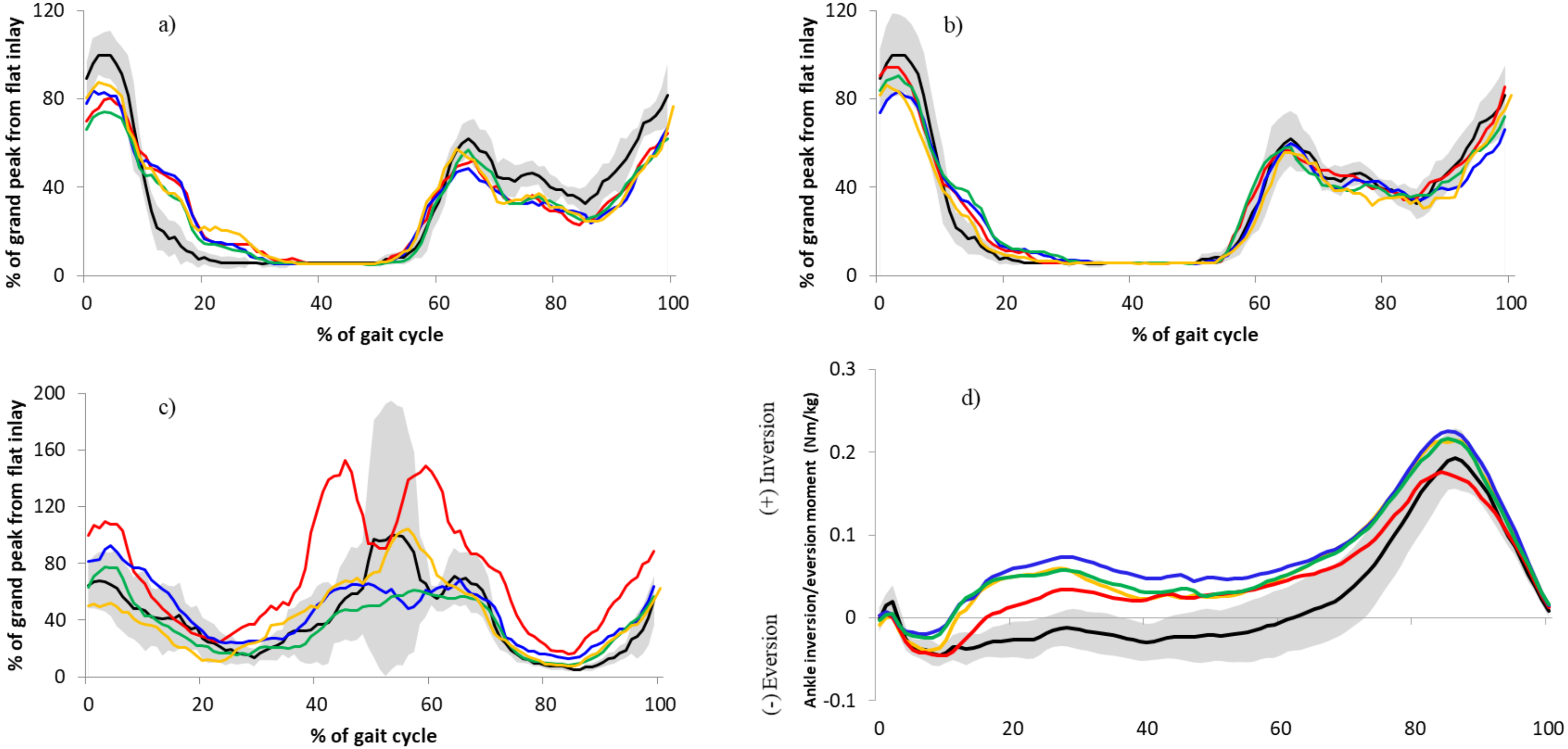


Figure 6-8. Data for excluded participant AL170118. a) Tibialis anterior EMG (fine-wire trials); b) Tibialis anterior EMG (fine-wire removed trials); c) Tibialis posterior EMG; d) ankle inversion/eversion moment. Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging. The grey shaded area represents the standard deviation of the flat inlay.

6.3.4 *Kinematics*

In **Figure 6-9** it can be seen that in early stance when the rearfoot was in inversion, the rearfoot appeared to be further inverted with FO with respect to the flat inlay, particularly with the Wedge and Arch & Wedge conditions. There was not a significant effect of condition on MaxEv ($p=0.077$, $\eta p^2=0.124$). An outlier was removed from the MaxEv data (MaxEv2), however there was still no main effect of condition ($p=0.133$, $\eta p^2=0.124$). There was a significant main effect of FO on ROMEv ($p=0.011$, $\eta p^2=0.233$). The ROMEv of the Wedge (7.5° ±2.7°) was almost significantly less than the flat (9.6° ±3.4°, p= 0.051).

As maximum calcaneus angle in early stance (MaxES) was not normally distributed, values in **Table 6-6** are presented as means and medians. However as the departure from normality was not extreme, an ANOVA was run and there was no significant effect of condition ($p=0.474$, $\eta p^2=0.056$). From **Figure 6-9b-c** the increase in inversion with the Wedge and Arch & Wedge appears to be more apparent in the no fine-wire trials than the fine-wire trials. However there was no significant interaction effect for the effect of condition and fine-wire ($p=0.375$, $\eta p^2=0.064$).

Data for MLA angle was available from 10 participants, presented in **Figure 6-11**. A further participant was identified as an extreme outlier from the MLA ROM data and excluded (**Table 6-7**). There was a trend towards decreased MLA ROM with all FO conditions, and this was significant for the Wedge condition (p=0.038).

From **Figure 6-10b** there was a reduction in dorsi/plantar flexion angle (i.e. more plantar flexed) with FO, particularly for the Wedge and the Arch & Wedge combined conditions.

Table 6-6. Discrete kinematic and kinetic variables of interest (n=17, all trials combined)

	Flat inlay	(±)	Salfordinsole	(±)	Arch	(±)	Wedge	(±)	Arch & wedge	(±)
Kinematics										
MaxEv (°)	-2.24	6.04	-0.80	7.55	-1.47	6.58	1.64	8.74	0.39	5.81
MaxEv2 (°)	-3.08	5.12	-1.40	7.37	-2.61	4.74	0.36	7.22	-0.61	4.25
ROMEv (°)	9.56	3.38	8.96	2.96	8.64	3.42	7.51	2.74	7.81	2.85
MaxES (°)	7.08	5.47	7.31	5.88	7.11	6.54	8.76	7.72	8.51	5.43
MaxES med (°)	5.65	5.35	6.04	7.95	5.79	6.99	8.06	8.34	7.47	4.85
Kinetics										
MaxMEv (Nm/kg)	-0.11	0.04	-0.09	0.04	-0.09	0.05	-0.08*	0.04	-0.07*	0.04
Decrease vs. flat inlay			13%		13%		30%		38%	
MaxMInv (Nm/kg)	0.18	0.07	0.19*	0.07	0.19	0.08	0.21*	0.08	0.22*	0.08
Increase vs. flat inlay			7%		8%		15%		19%	

(* $p<0.05$ with respect to flat inlay), med= median)

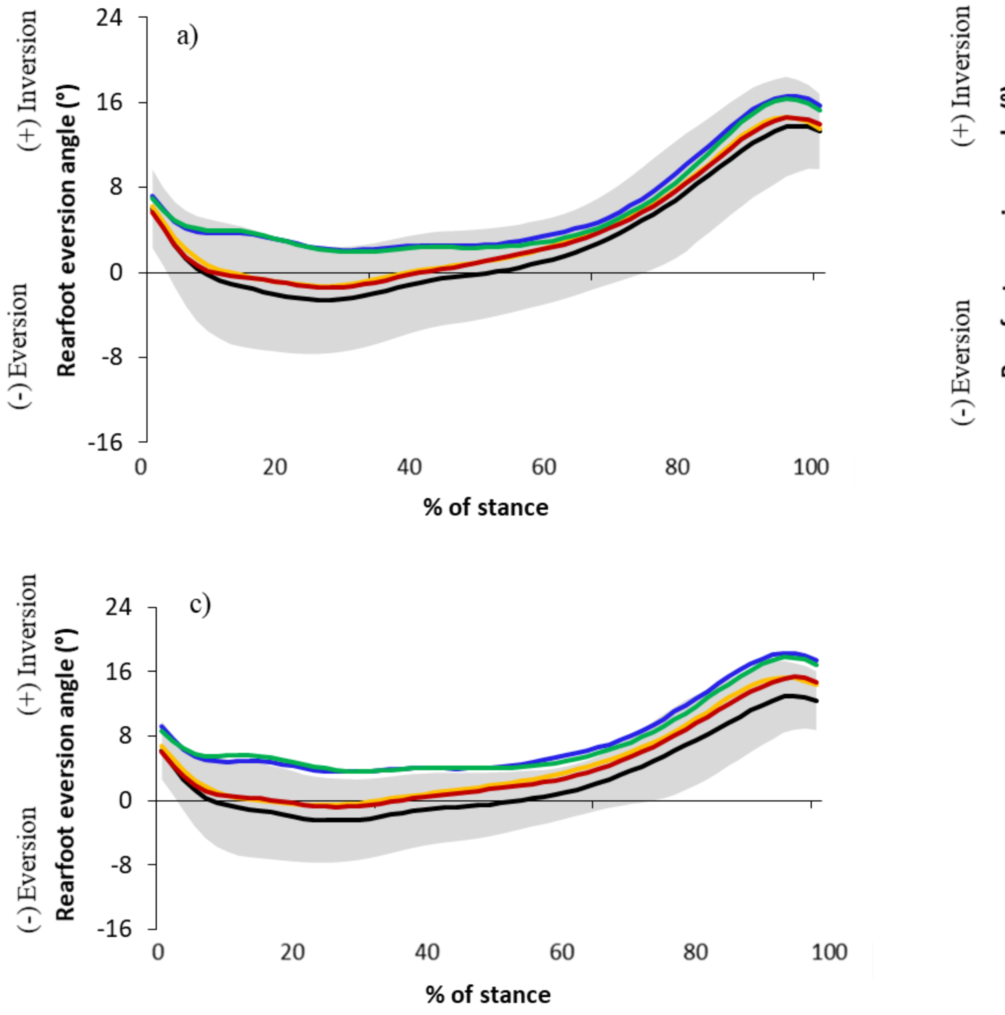


Figure 6-9. Rearfoot angle (one participant an extreme outlier excluded): a) All trials combined (n=16) b) Fine-wire trials (n=16)); c) No fine-wire trials (n=16). Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging. The grey shaded area represents the standard deviation of the flat inlay.

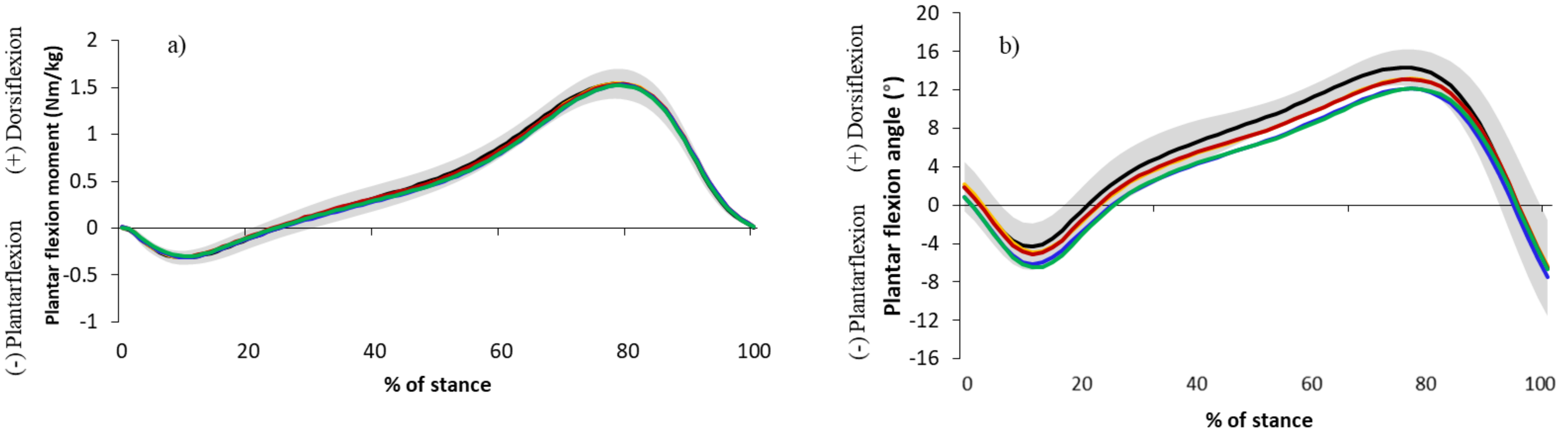


Figure 6-10. Sagittal plane kinetics and kinematics (n=17). a) Plantar flexion moment; b) plantar flexion angle. Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging. The grey shaded area represents the standard deviation of the flat inlay.

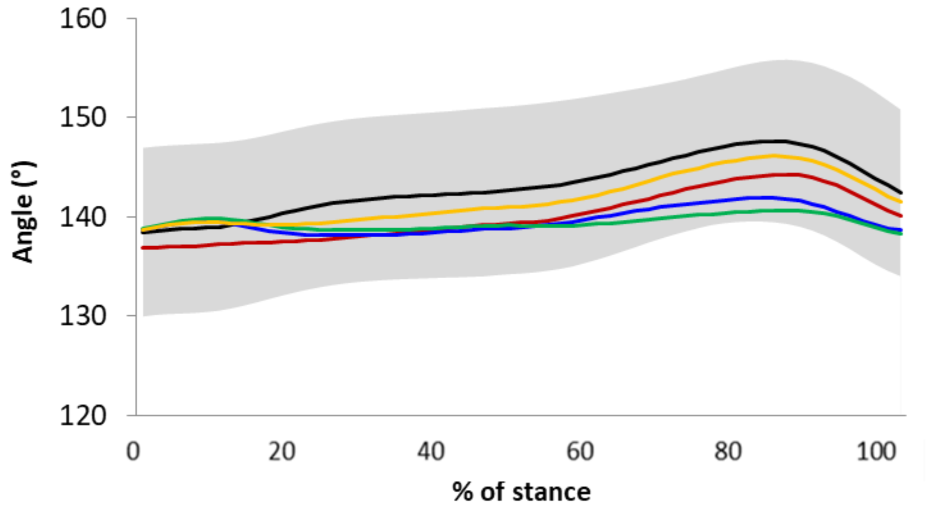


Figure 6-11. Medial longitudinal arch angle over stance. Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging. The grey shaded area represents the standard deviation of the flat inlay.

Table 6-7. Medial longitudinal arch angle range of motion over stance, all trial (n=9)

	Flat inlay	(±)	Salfordinsole	(±)	Arch	(±)	Wedge	(±)	Arch & Wedge	(±)
Mean (± SD)	11.0°	1.3°	9.4°	1.9°	9.5°	2.4°	6.8°*	2.4°	8.3°	2.5°
Median (± IQR)	10.7°	1.7°	8.9°	2.2°	9.5°	2.0°	6.7°	1.9°	8.6°	3.7°

*p=0.038 vs. Flat inlay

6.3.6 *Kinetics*

From **Figure 6-12** the external ankle inversion/eversion moment shifted more into an inversion moment with FO. There was decreased MaxMEV (i.e. more inverted), in all conditions from a 30% decrease to a 38% decrease (**Table 6-6**), with a significant effect of condition ($p<0.001$, $\eta p^2=0.530$). Decreased MaxMEv was significant for the Wedge ($p=0.001$) and Arch & Wedge ($p<0.001$) with respect to the flat inlay. There was an increase in MaxMInv with all FO, from 15% increase to 19% increase (**Table 6-6**), with a significant effect of condition ($p<0.001$, $p^2=0.540$). Increased MaxMInv was significant for the Salfordinsole ($p=0.035$), Wedge ($p=0.001$) and Arch & Wedge ($p<0.001$) and not significant with the Arch ($p=0.073$) with respect to the flat inlay.

From **Figure 6-10a** there was no difference in plantar flexion moment across conditions.

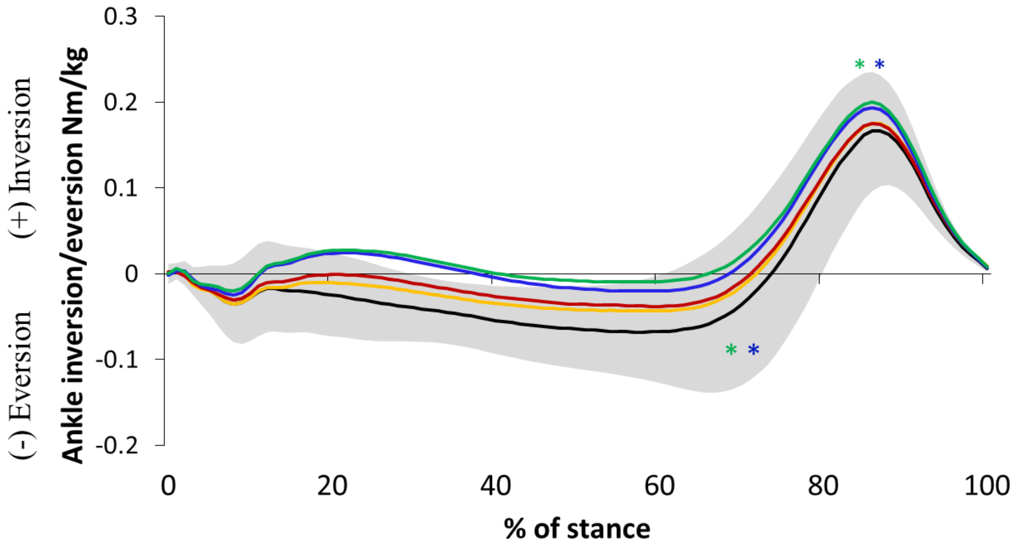


Figure 6-12. External ankle inversion/eversion moment (n=17) Black lines: flat inlay; yellow lines: Salfordinsole; red lines: 6 mm increase in arch height; blue lines and asterisks: 8 mm increase in medial wedging; green lines and asterisks: 6 mm increase in arch height combined with 8 mm increase in medial wedging. *p<0.05. The grey shaded area represents the standard deviation of the flat inlay

6.4 Discussion

6.4.1 *Recap of hypotheses (accepted/rejected)*

It was hypothesised that increases in medial heel wedging would lead to the following changes relative to the flat inlay:

6.4.1.1 *Hypothesis 1- medial heel wedging*

1.8 Reduced TP EMG peak amplitude in early stance (accepted).

1.9 No change in EMG of MG, PL or TA (accepted).

1.10 Decreased EMG of ABH for Arch & Wedge in mid-late stance (rejected).

1.11 Reduced external eversion moment across stance (more inversion, accepted).

1.12 Reduced peak external eversion moment (MaxMEv, accepted).

1.13 Increased peak external inversion moment (MaxMInv, accepted).

1.14 Increased inversion at foot contact (MaxES), reduced peak rearfoot eversion angle (MaxEv) and reduced rearfoot eversion range of motion (ROMEv), (rejected)

6.4.1.2 *Hypothesis 2- arch geometry*

With a standard Salfordinsole FO and further increase in arch height the following *non-systematic* changes were hypothesised relative to the flat inlay condition:

2.1 Earlier onset of TP activity in early stance (rejected).

2.2 Reduced TP EMG peak amplitude in late stance (rejected).

2.3 No change in muscle activity of MG, PL or TA (accepted).

2.4 Decreased muscle activity in mid-late stance of ABH (rejected).

2.5 Reduced external eversion moment across stance (more inversion, accepted).

2.6 Reduced peak external eversion moment (MaxMEv, rejected).

2.7 For the Salfordinsole and FO with only a 6 mm increase in arch height (Arch), but not the Arch & Wedge: no effect on rearfoot kinematics (accepted).

2.8 Decreased range of motion of MLA (rejected).

6.4.2 *EMG*

Hypothesis 1.1 for reduced TP EMG peak amplitude in early stance with increased medial heel wedging can be accepted. Whilst there was a trend towards reduced TP EMG peak amplitude in early stance with all FO, is was not significant for the Salfordinsole condition nor with the addition of increased arch height.

The Salfordinsole FO have minimal additional material in the rearfoot area, so it is unsurprising that its effect in early stance when only the rearfoot is in contact with the ground, is subtle. A true systematic effect of wedging may have been found with different magnitudes of wedging rather than the addition of extreme increased arch height to extreme wedging. Systematically increasing the degree of medial heel wedging has been shown to linearly reduce external ankle eversion moment (Telfer et al., 2013b), so it is possible that there could be a corresponding linear decrease in peak TP EMG amplitude. However extreme designs were used in this study to explore what changes in TP EMG were possible with specific changes in FO wedging.

Hypothesis 2.1 for earlier onset of TP EMG peak amplitude in early stance and hypothesis 2.2 for reduced TP EMG peak amplitude with increased arch height can be rejected, as there was no effect of FO on timing of TP and although there was a reduction of peak TP EMG amplitude in late stance the variability in TP activation was high.

The reduction in TP EMG amplitude (14-17%) in early stance with FO was of a similar magnitude to that seen with custom and pre-fabricated FO relative to shoes by Murley et al. (12-19%) (Murley et al., 2010a). The difference in the relative size of the two peaks of activation between the previous and present study is likely due to differences in foot posture between the two study populations. Murley et al. (2010a) recruited flat footed individuals, based on clinical and radiographical measurements, whereas in this study participants with a normal foot and a flat foot were included (average FPI= 2). The profile of TP activation in the present study, with a greater peak in early stance than late stance, resembles that of the normal arch group in a previous paper by Murley and colleagues (2009c) comparing normal and flat foot posture. Based on the results presented here, it appears that the potential for FO to reduce TP activity in early stance is not specific to those with a flat foot. In another study customised FO with a 4-mm medial skive at 15° and a 5° extrinsic rear foot post reduced TP EMG activity with respect to barefoot, but not compared to the shoe only condition (Maharaj et al., 2018). It could be that in the latter study the effect of the FO geometry was too subtle to be detected as a significant difference within the variable TP EMG pattern. Alternatively it could be that the stiffer insole material (semi rigid 4-mm polypropylene) compared to the EVA shoe liner, was less shock absorbent, which may have counteracted any potential effect of the FO geometry (Maharaj et al., 2018), or affected a different aspect of muscle function than activation.

The reduction in TP activity with medial wedged FO could be beneficial in treating tibialis posterior tendon dysfunction, as reduced muscle activity would mean less force going through the TP muscle tendon unit, which could facilitate tendon healing. Generating negative work through eccentric muscle contractions has the potential to lead to overuse injury (Maharaj et al., 2017a). In this study FO with a medial heel wedge reduced TP activity in early stance, the period when TP is acting eccentrically to resist the external eversion moment.

Four participants were identified as outliers based on the peak TP EMG amplitude and excluded from analysis. For the first outlier, the protocol was likely still too long because there was still evidence of signal degradation. In another two outliers there was no clear reason why the signals were outside the norm. A fourth outlier excluded (**Figure 6-8**) was the oldest in the group (54 years). From **Figure 6-8c**) it can be seen that the peaks of TP activation for this participant were a little flatter than the group average (**Figure 6-7**a). However, in **Figure 6-8d**) it can be seen that the trend of the ankle inversion/eversion moment response to FO followed the same pattern as the group average (**Figure 6-12**). Unlike the group average, in which there was no effect of FO on TA EMG activity, in this excluded participant there was a trend for reduced TA activity in early stance with FO conditions in both trials with the fine-wire present (**Figure 6-8a**) and without the fine-wire present (**Figure 6-8b**). It could be that this participant had reduced normal functioning of their TP, perhaps as a consequence of ageing and their TA may be compensating and have a greater than average role in inversion. Potential functional decline in TP with age may limit the transferability of the results of this study to a patient population, particularly as patients with rheumatoid arthritis, tibialis posterior tenosynovitis and acquired flat foot are often middle-aged or older (Barn et al., 2013, Jordan et al., 2014).

Hypotheses 1.2 and 2.3 for a lack of change in MG, PL or TA with FO can be accepted. There was no effect of condition on MG, PL or TA. There was however an effect of the fine-wire electrode in TP on peak PL amplitude in that there was a main effect for greater amplitude in the fine-wire trials than the trials without fine-wire. It is possible that increased PL amplitude was a means of compensating for restricted ROM or activity in TP. There was a slightly greater initial rearfoot eversion in the no fine-wire trials than fine-wire trials, although not significant. If PL successfully compensated for reduced function in TP, then we might not expect a difference in the kinematics between fine-wire and no fine-wire trials. Nonetheless a lack of an effect of condition or interaction effect between electrode presence and condition indicates that, importantly with respect to the research question, there was no effect of FO on PL EMG. One previous study that found increased activity in PL with FO used fine-wire electrodes in both the PL and TP (Murley et al., 2010a). In earlier work by the same group, Maximum PL amplitude did increase in walking by 19% for pronated individuals when wearing a 15° inverted orthosis compared with shoes alone, but without a linear dose-response to the degree of wedging (Murley and Bird, 2006). Other studies using surface electrodes have not found an effect of FO on PL EMG in walking (Barn et al., 2013, Telfer et al., 2013a). As PL is the antagonist of TP, if FO reduced TP EMG activity this would possibly be accompanied by increased PL EMG activity. With decreased TP EMG activity with FO one might expect increased PL EMG activity, however muscle activity from TP and PL do not necessarily represent equal opposing inversion and eversion moments respectively, due to additional muscle tendon parameters (Murley et al., 2009a) and different moment arms. As evident in the graphs **Figure 6-7a, c** and **d** there are typically two peaks of TP EMG activity, in early and late stance, but one peak of PL EMG activity in mid-late stance (**Figure 6-6a** and **b**). The lack of effect of FO on MG and TA is in line with what was expected given previous work in walking (Murley and Bird, 2006, Telfer et al., 2013a, Murley et al., 2010a).

The results give some grounds to reject the hypotheses 1.3 and 2.4 that FO would reduce ABH EMG in mid-late stance. For ABH EMG there was a trend towards *increased* activation in late stance for all FO conditions, although data was from a limited number of participants. It is possible that better ABH EMG data could have been obtained with another hole in the shoes, however several holes had already been made for the kinematic markers to calculate MLA angle, so another hole may have compromised the integrity of the shoe. The trend for increased activity with FO is against the belief held by some that FO would reduce the activity of the intrinsic foot muscles. Reduced integrated EMG of ABH was found with FO during step descent (Bonifacio et al., 2018). Further work with a greater sample size, less trials per condition (to facilitate maintained contact of the sensor) and perhaps different holes in the shoes would be necessary to better understand the effect of FO on ABH in walking.

6.4.3 *Kinetics*

Hypotheses 1.4, 1.5 and 1.6 can be accepted, as there was reduced external eversion moment across stance, reduced peak external eversion moment and increased peak external inversion moment mid-late stance with the Wedge and the Arch & Wedge. The reduction in TP EMG amplitude with the Wedge and Arch & Wedge FO was of a lesser magnitude (17% and 14% respectively) than the reduction MaxMEv (reduced maximum eversion of -30% and -38%

respectively). There was reduced external moment across stance (**Figure 6-12)** for all FO conditions, so hypothesis 2.5 can be accepted. Although there were changes in the discrete kinetic variables with the Arch & Wedge, increasing arch height alone with Salfordinsole and the Arch condition, did not significantly alter joint moments, so hypotheses 2.6 can be rejected. Nevertheless, the 13-38% reduction in peak ankle eversion moment across FO conditions in this study is greater than the 1.1±1.1% (p= 0.003) per 2° of medial posting reported previously (Telfer et al., 2013b). The reduction in external ankle eversion moment demonstrates the FO had the underlying effect on external forces expected and necessary to elicit a change in the internal forces in response.

A different magnitude of change between EMG and joint moment with medial heel wedging is not surprising given that firstly there is not a linear relationship between force and EMG for dynamic contractions. Secondly, the axes of rotation around which the TP acts is not the same as where the external ankle inversion/eversion moment was calculated. Finally, the FO could have influenced the energy storage of TP tendon (Maharaj et al., 2016, Maharaj et al., 2017b), which would affect the joint moment, but not be reflected in the external ankle inversion/eversion moment. Conversely energy storage and release in the TP tendon in late stance may have contributed to the increase in greater inversion moment in late stance in the Wedge and Arch & Wedge conditions, despite the fact that the rearfoot, where the wedge acts, is not in contact with the ground at this stage. Sweeney (2016) found that the FO exerted their effect on the ankle moment in mid-late stance. Nevertheless, the present study has demonstrated that with specific changes in rearfoot posting of FO, it is possible to change both joint moment and muscle activity.

6.4.4 *Kinematics*

Hypothesis 1.7 can be rejected. In **Figure 6-9** it can be seen that there was a shift into a more inverted foot position with the Wedge and the Arch & Wedge conditions, however the change in discrete kinematic variables was not significant. This is likely due to the large standard deviations in the data set as there was over a 3° change in the mean MaxEv2 from -3.08° ±5.12° eversion with the flat inlay to 0.36° ±7.22° and -0.16° ±4.25° with the Wedge and Arch & Wedge conditions. High individual variation in the kinematic response to FO has been noted previously (Donoghue et al., 2008, Mills et al., 2009), with Donoghue et al. (2008) reporting a 2-9° variation in discrete kinematic measures. Although the reduction in peak rearfoot eversion with FO in this study was not statistically significant, it was greater than the 2.08° to 2.35°

reduction ($p \leq 0.004$) reported in a meta-analysis of the effect of external devices on kinematics (Cheung et al., 2011). The 1.68° reduction in peak rearfoot eversion angle (MaxEv2) with the Salfordinsole in this study was less than the significant 3.8° (2.7°-5.0°, 95% confidence interval, $p < 0.001$) reduction reported previously (Majumdar et al., 2013). The different magnitude of change in rearfoot eversion angle in the previous work versus the present study could again be attributed to the high individual variation in response to FO and/or the larger sample size in the previous work (n=27) versus the present study (n=16 for kinematics).

No significant change was found in rearfoot kinematics with increased arch height as expected, so hypothesis 2.7 can be accepted. Hypothesis 2.8 that increased *arch height* would decrease MLA ROM can be rejected. There was no effect of arch height on MLA ROM, however there was a reduction from 11.0° ±1.3° to 6.8° ±2.4° with the Wedge condition, (which is a reduction of the same magnitude as the MDD in Chapter 5). As a marker on the calcaneus is used to calculate the MLA, a more elevated rearfoot with the wedging may have led to this reduction in MLA ROM.

6.4.5 *Limitations*

A limitation to this study is that it is possible that even with the revised protocol there could have still been some reduction in signal amplitude recorded with the fine-wire electrode over time, independent of the FO conditions. The maximum time from which EMG can be accurately recorded with the electrodes used without a drop in amplitude was not established. However, from data collected from one participant walking over an hour in the same condition in the appendix (**Figure 0-5**) there was a drop in amplitude by the fourth recording. Each recording consisted of six trials of the same distance (6 m) as in the main study, so the first three recordings represents 18 trials. It could be concluded that approximately 20 walking trials can be collected before there is a drop in amplitude. With only three trials per condition in the main study and five conditions, the total number of trials equates to only 15. Additionally the conditions were randomised in the main study, so it is conceivable that any slight degradation in amplitude over time would have been washed out by participants wearing the FO conditions in different orders, so a main effect of medial heel wedging was still detectable.

It has been argued that the presence of a fine-wire electrode in the muscle could affect walking pattern and thus data recorded with a fine-wire electrode present may not adequately reflect normal gait (Semple et al., 2009). However, it does not appear that the presence of fine-wire impacted the main findings of this study. Firstly, one might expect gait to be altered if the presence of the electrode was uncomfortable. Four of the nineteen participants included in the final analysis complained that they found walking with the electrode in place uncomfortable. However the group trend for reduced TP amplitude in early stance for all FO conditions, but particularly the wedge conditions (**Figure 6-7a**), can be seen in both the participants who did not complain of discomfort (**Figure 6-7c**) and those who did **Figure 6-7d**).

The need to repeat the walking trials without recording fine-wire EMG, in order to obtain more moment data, meant that kinematics can be compared with and without fine-wire in TP. Walking speed was significantly slower in the trials without the fine-wire electrode compared to with the fine-wire electrode (**Table 6-4**). Effort was made to keep walking speed within ±5% of participants' pre-established self-selected walking speed during trials. However, the participant was not asked to repeat a trial outside the self-selected range in the fine-wire trials, in the interest of obtaining the TP EMG data as quickly as possible, with regards the signal attenuation problem identified. Nevertheless, it can be seen that the rearfoot eversion angle from all trials (**Figure 6-9b**) is very similar to rearfoot eversion angle in both **Figure 6-9c** with the fine-wire present and **Figure 6-9c**) without fine-wire. In all trials the patterns of rearfoot eversion angle is similar and the rearfoot was clearly more inverted for the first 40% of the gait cycle in the Wedge and the combined Arch & Wedge conditions compared to the other conditions irrespective of the presence of fine-wire. If the fine-wire electrode impaired walking, we might expect walking speed to be slower in trials in which it was present than when in fact the opposite was the case. It can thus be concluded that the presence of the fine-wire electrode did not impede walking, so does not reduce the validity of the finding of reduced TP amplitude in early stance with the Wedge and Arch & Wedge combined FO.

The nil effect observed for some variables is unlikely to be due to a lack of power, as effect sizes were typically less than small (<0.1) for non-significant effects. Nineteen participants were included in the EMG data in this study, which was less than the required sample size determined in the a priori power calculation of 25-28 participants. However the power calculation was based on an effect size of 0.59 and 0.56 for the effect of FO on TP and PL peak amplitude respectively in a previous study (Murley et al., 2010a). The effect size for the effect of FO condition on PL peak amplitude was 0.099 in this study, which indicates a genuine lack of effect.

6.5 Conclusion

The concurrent reduction in external eversion moment and peak TP EMG amplitude in early stance with Wedge and Arch & Wedge demonstrates the potential for specific FO geometry to alter TP activation. If the intention of a treatment was to reduce force going through the TP muscle tendon unit, which could facilitate tendon healing in the case of tibialis posterior tendon dysfunction, then a medial wedge FO could be an effective design to achieve this.

Chapter 7 EFFECTS OVER TIME OF FOOT ORTHOSES ON SOFT TISSUE, PLANTAR PRESSURE AND SKIN SENSITIVITY

7.1 Introduction

This study took a novel approach to FO research in measuring soft tissue morphology, plantar pressure and skin sensitivity before and after wearing FO for three months.

7.1.1 *Rationale*

The mechanisms behind the effect of FO has not been studied with respect to the potential for altered loading under the foot over time leading to adaptations in internal foot structures over time. As discussed, FO alter the pressure applied to the foot (Farzadi et al., 2015, Hodgson et al., 2006, McCormick et al., 2013, Nester et al., 2003b, Sweeney, 2016, Telfer et al., 2013a), consequently altering the distribution of work required of internal structures, including muscles and connective tissue. It is well established that muscles and tendons of the lower limb can change their size and structure as a result of a change in the load they experience, due to training for example (Folland and Williams, 2007, Magnusson et al., 2008a, Reeves et al., 2004). Changes in the size and structural properties of these soft tissues as a result of an increase or decrease in loading would affect the force generating capacity of muscle tendon units (Folland and Williams, 2007, Magnusson et al., 2008a, Reeves et al., 2004, Lieber and Friden, 2000). It has also been shown that intrinsic foot muscles can change size in response to modified loading of the foot with a change in footwear (Bruggemann et al., 2005, Johnson et al., 2016). It is therefore possible that redistributing the external loading of the foot with FO has potential to alter muscle and connective tissue morphology.

Changing the position at which internal structures within the foot operate with FO could lead to morphological changes over time due to the plasticity of these structures. The soft tissues that could respond to such changes in load and position, like the plantar fascia, also contribute to foot posture. Greater cross sectional area and thickness of extrinsic foot muscles, less cross sectional area and thickness of intrinsic foot muscles and a thinner plantar fascia, have been shown in those with a pronated foot posture compared to a neutral foot posture (Angin et al., 2014, Angin et al., 2018). Pes planus has also been associated with a thicker tibialis anterior tendon, a thicker peroneus longus muscle and a thinner Achilles tendon (Murley et al., 2014a). Thus, foot posture and foot type appear to be some of the factors influencing soft tissue morphology. It follows that if the distribution of foot loading and foot position was changed with a FO, then soft tissue morphology might change over time.

Foot orthoses could increase the contact area at specific regions of the foot, such as the medial arch (Williams and Nester, 2010c). Changes in pressure with FO incorporating a metatarsal bar have been shown to lead to changes in skin sensitivity in the forefoot, in that there was increased ability to detect mechanical stimuli from a bespoke loading device (Vie et al., 2015). Increased sensitivity could result from stimulation of mechanoreceptors in the sole of the foot that would otherwise not have been activated, or been activated in different ways. Cortical plasticity allows for the potential for increased skin sensitivity through increasing the relevant area in the primary somatosensory cortex (Björkman et al., 2009). Increased pressure in the medial arch could increase sensitivity due to increases in the relative weighting given to receptors from that region, or skin sensitivity could decrease if the receptors become saturated (2.2.10.1). Changes in the magnitude of loading in specific regions with a FO could also lead to changes in the sensitivity of other mechanoreceptors in adjacent areas (Vie et al., 2015). Altered stimulation of mechanoreceptors would modulate afferent feedback to the CNS, which has the potential to influence muscle activity and subsequent movement pattern (Bent and Lowrey, 2012, Fallon et al., 2005, Howe et al., 2015, Nurse and Nigg, 2001, Perry et al., 2008). Any long term effect on skin sensitivity at sites other than the metatarsal area has not been investigated. The response of the medial arch to increases in pressure may be different to the metatarsal area because the medial arch is the most sensitive region of the foot, despite having potentially fewer mechanoreceptors than other regions (2.2.10.1).

Previous work has shown the immediate effect of the standard Salfordinsole FO is to reduce peak pressure in the medial and lateral heel and increase peak pressure in the medial arch (Sweeney, 2016). The relative change in plantar pressure between control and FO with a medial arch support has been shown to be similar at initial assessment and following 4-6 weeks of use (Hodgson et al., 2006, McCormick et al., 2013, Farzadi et al., 2015). Muscle and or cutaneous sensitivity may well take longer than six weeks to adapt to loading. Thus, it remains to be seen whether neuromuscular adaptation to alterations in plantar loading occurs with long-term FO use and whether this affects foot function.

The effect of FO incorporating a medial arch support on soft tissue size and structure and skin sensation around the foot and ankle has not been studied. The purpose of this study was to enhance our understanding of the mechanisms behind the effect of FO by investigating whether skin sensitivity and muscle and connective tissue morphology were altered when plantar loading was altered by use of a FO.

7.1.2 *Approach*

Research question: **What are the longitudinal effects of foot orthoses on soft tissue morphology, plantar pressure and skin sensitivity?**

Aim: To investigate the effect of changes in peak plantar pressure with foot orthoses worn for three months on soft tissue morphology, plantar pressure and skin sensitivity

7.1.3 *Hypotheses*

7.1.3.1 *Hypotheses- soft tissue structures*

It was hypothesised that from pre to three months post use of a FO, there would be increased CSA and thickness of ABH, FHB and Achilles tendon, and increased thickness of the plantar fascia, in the FO group.

7.1.3.2 *Hypotheses- skin sensitivity*

From pre to three months post use of a FO, the following changes in the FO group were hypothesised:

- Skin sensitivity would increase (i.e. monofilament threshold and neurothesiometer amplitude would decrease) in the medial arch.
- Skin sensitivity would decrease (i.e. monofilament threshold and neurothesiometer amplitude would increase) in the medial heel and lateral heel following decreased loading.
- Skin sensitivity would remain the same in the other regions of the foot (lateral arch and 1st metatarsal head and dorsum).

7.1.3.3 *Hypotheses- Plantar pressure in the standard shoe*

It was hypothesised that there would be no significant difference between baseline and three-month follow up in both groups for the percent differences in peak pressure in the medial arch and the heel between the flat inlay and FO in the standard shoe.

7.2 Methods

7.2.1 *Participants*

Fifty-three healthy participants (Females= 36, mean ±SD age: 29 ±9 years; height: 1.67 ±0.07 m; mass: 68.6 ±12.9 kg) wore pre-fabricated EVA or thermoplastic FO (Salfordinsole) for three months (n=27) or no insert (n=26). The study recruited healthy participants to take a purely mechanistic study of the response to FO and because there is not substantial evidence that individuals with the pathologies most commonly treated with FO demonstrate different kinematics from each other or from healthy controls (Dowling et al., 2014, Neal et al., 2014, Pohl et al., 2009, Powers et al., 2002, Ribeiro et al., 2011). Participants were not recruited based on FPI score as FO can be prescribed for individuals with a range of foot postures, not only pes planus.

Participants were excluded if they self-declared any of the following:

1) Symptomatic lower limb injury or foot or ankle pathology/deformity in the last three months.
2) Any cardiovascular, musculoskeletal or neurological conditions or disease that affect foot sensation or muscle structure (e.g. diabetes, peripheral vascular disease).
3) Walk with an aid.
4) Have been wearing foot orthoses during the last six months.
5) Footwear worn during the day is unsuitable for foot orthoses.

Participants at the University of Salford (n=39) were randomised using Microsoft Excel into either the intervention or the control group. Participants at the University of Guelph (n=14) were allocated into either the intervention or control group based on the shoe size due to the limited availability of FO sizes and the time restrictions of the intervention. Participants under 18 years of age were not recruited as there was little data currently on baseline skin sensitivity measures in children.

7.2.2 *Protocol*

Height, body mass, shoe size, FPI and leg dominance were recorded at the first assessment. Leg dominance was established with the question: “which leg would you kick a ball with?”; the answer of which was considered to be the dominant leg. Ultrasound measurements, skin sensitivity testing, plantar pressure recordings during walking in the standard shoes with and without the FO and an assessment of physical activity levels were conducted at baseline and after three months of either FO use or continuation of normal footwear use (they continued to wear their own shoes with their existing inserts). Physical activity levels were assessed using the International Physical Activity Questionnaire (IPAQ), Short Form. This had seven questions relating to the duration and intensity of physical activities and sitting over the last seven days and it would take approximately five minutes to complete. Participants were considered “active” if their self-reported physical activity levels met the World Health

Organization guidelines (2011), otherwise they were considered "inactive". In brief, an active individual does a minimum of 150 minutes moderate-intensity activity (including walking), or 75 minutes of vigorous intensity activity per week in bouts of 10 minutes or more.

7.2.3 *Foot orthoses*

Participants were asked to wear a pair of thermoplastic or EVA FO for a minimum of four hours a day, following a protocol from a previous randomised control trial (RCT) of insoles for knee pain (Hossain et al., 2011). Each week the participants received emails checking on the condition of the FO, whether they were comfortable to use, whether they had been wearing them and to record any change in self-reported physical activity levels, (if they dramatically increased or decreased how often they are active in the week, e.g. starting a new training program).

Participants were issued FO that elicit at least an 8% increase in peak pressure in the medial arch and an 8% decrease in peak pressure in the medial heel compared to walking in their normal shoes alone. Pressure measurements were taken using shoe insoles with 99 capacitive sensors (Pedar Mobile, Novel Electronics Inc., GmbH Munich, Germany). The Pedar insoles were calibrated using the Novel bladder system in advance of data collection, which is recommended by Novel and has independently been shown to improve the accuracy of the insoles (Hsiao et al., 2002). Plantar pressure values were recorded after two minutes of habituation, allowing adequate time to perform the 166 steps recommended for familiarisation to footwear (Melvin et al., 2014). Walking speed was maintained within ±5% of the participants' self-selected walking speed. Walking speed was controlled using timing gates or a mobile app (GaitAnalysisPro, Developer: YTA, K.K., © 2014 Wataru, Yasuda).

Initially, participants wore either a medium density EVA or "flex" (orange) thermoplastic Salfordinsole FO and pressure data were recorded and compared with that from no FO. If the minimum pressure change in the heel was not achieved, medial heel wedges were added to the medial heel (if the pressure change was insufficient at that location), and/or the same insole in high density EVA or "firm" (blue) thermoplastic was worn. If the initial Salfordinsole FO were uncomfortable in the medial arch area for the participant, then they were given a low density EVA Salfordinsole FO.

Variations in material density and medial heel wedge use were used until an 8% increase in peak pressure in the medial arch and an 8% decrease in peak pressure in the medial heel, compared to walking in their normal shoes alone, was achieved. Twenty participants achieved this using medium density EVA or flex orange thermoplastic Salfordinsole FO alone, one participant received the flex thermoplastic Salfordinsole FO with additional medial heel wedges, one participant received firm thermoplastic Salfordinsole FO and four participants received low density EVA Salfordinsole FO. The predominant FO of the intervention were made of medium density EVA or of thermoplastic that is comparable to high density EVA (Majumdar et al., 2013). The most common FO in this study therefore reflects clinical practice, as the majority of prescribed prefabricated and customised FO are made of either medium or high density EVA or rigid plastic (Nester et al., 2017).

7.2.4 *Pressure changes with the FO used in the intervention (measured in their own shoes pre intervention, n=15)*

Mean peak pressure at the medial and lateral heel was significantly less in the FO vs. the flat inlay (p= 0.012 and p= 0.004 respectively, **Figure 7-1**). Mean peak pressure at the medial arch was significantly greater in the FO vs. the flat inlay (p= 0.005), but not significantly different at the lateral arch (p= 0.106, **Figure 7-1**).

The mean ± SD percent decrease in peak pressure was 21% ±14% at the medial heel and 17% ±14% at the lateral heel. The mean percent increase in peak pressure at the medial arch was 15% ±19% and the mean percent difference at the lateral arch was 7% ± 17%.

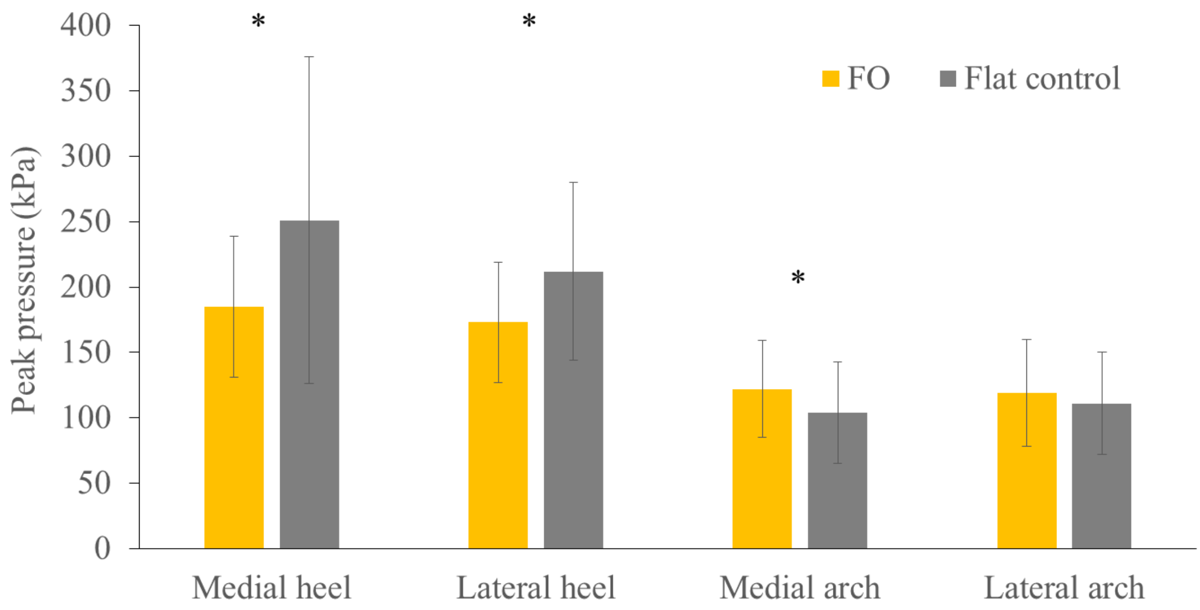


Figure 7-1. Mean ± SD of peak pressure (n=15 due to missing data) at the medial heel, lateral heel, medial arch and lateral arch with foot orthoses (FO) and the flat inlay in the foot orthoses group in their own shoes. * denotes significant difference (p<0.05)

7.2.5 *Plantar pressure measurements in standard shoes*

In order to test the hypothesis that the effect of the FO on regional peak pressures would be consistent over time, both the intervention and control groups walked in standard shoes with a flat EVA insert and FO at baseline and after three months. The FO used for the plantar pressure measurements in the intervention group was the same FO they wore during the three month intervention. The standard shoes were the same as those used in the previous two experimental studies in this thesis (Lonsdale Leyton, London). The use of the standard shoes for these plantar pressure measurements was made so that the effect of the FO could be compared across participants without the confounding effect of different material properties of the participants' own shoes.

7.2.6 *Measures of soft tissue thickness and cross-sectional areas*

Ultrasound images were recorded with one of three US machines: MyLab 70 Xvision ultrasound machine with a 13 MHz linear array transducer (Type, LA523, Esoate Europe, UK), a portable Venue 40 (GE Healthcare, UK) and a portable M-turbo musculoskeletal ultrasound system (Sonosite, Bothell, WA, USA) with a 6 cm linear array probe (HFL50x, 15-6 MHz wideband). Images for each individual participant were recorded with the same machine pre and post three-month FO use. To optimise image quality the automatic gain settings were used and the focus and depth were adjusted manually for each structure.

Before the ultrasound scan, standardized landmarks were marked with eyeliner or water-soluble pen on the participants' lower legs and feet. Measurements of structures focused on those that previous studies showed to have an association with foot posture, play a role in controlling pronation and supporting the medial longitudinal arch of the foot (Angin et al., 2014, Kelly et al., 2014, Murley et al., 2014a, Semple et al., 2009).

The scanning protocol for measuring the thickness of the abductor hallucis (ABH), flexor digitorum brevis (FDB), flexor hallucis brevis (FHB), proximal plantar fascia (PFINS) and mid-portion plantar fascia (PFMID) was as outlined previously (Angin et al., 2014, Crofts et al., 2014) and shown in **Figure 7-2**. The cross-sectional area (CSA) was also measured for ABH and FDB. Two images were taken per structure.

Images of the Achilles tendon were firstly taken at the insertional site on the calcaneus (ATINS). Rather than recording the images at an absolute distance from a bony landmark, the mid-portion cross-sectional area and thickness of the Achilles tendon were measured at the level where the soleus muscle was visible below the tendon (ATMID). These positions reflect the fact that Achilles tendinopathy typically occurs at the insertion of the tendon on the calcaneus, or in the mid-portion (2-6 cm from the calcaneus) (Irwin, 2010). The cross-sectional area is typically the smallest in the mid-portion of the Achilles tendon (Magnusson and Kjaer, 2003). In the appendix an additional project is presented as a **Conference paper on the relationship between toe grip strength and muscle size**.

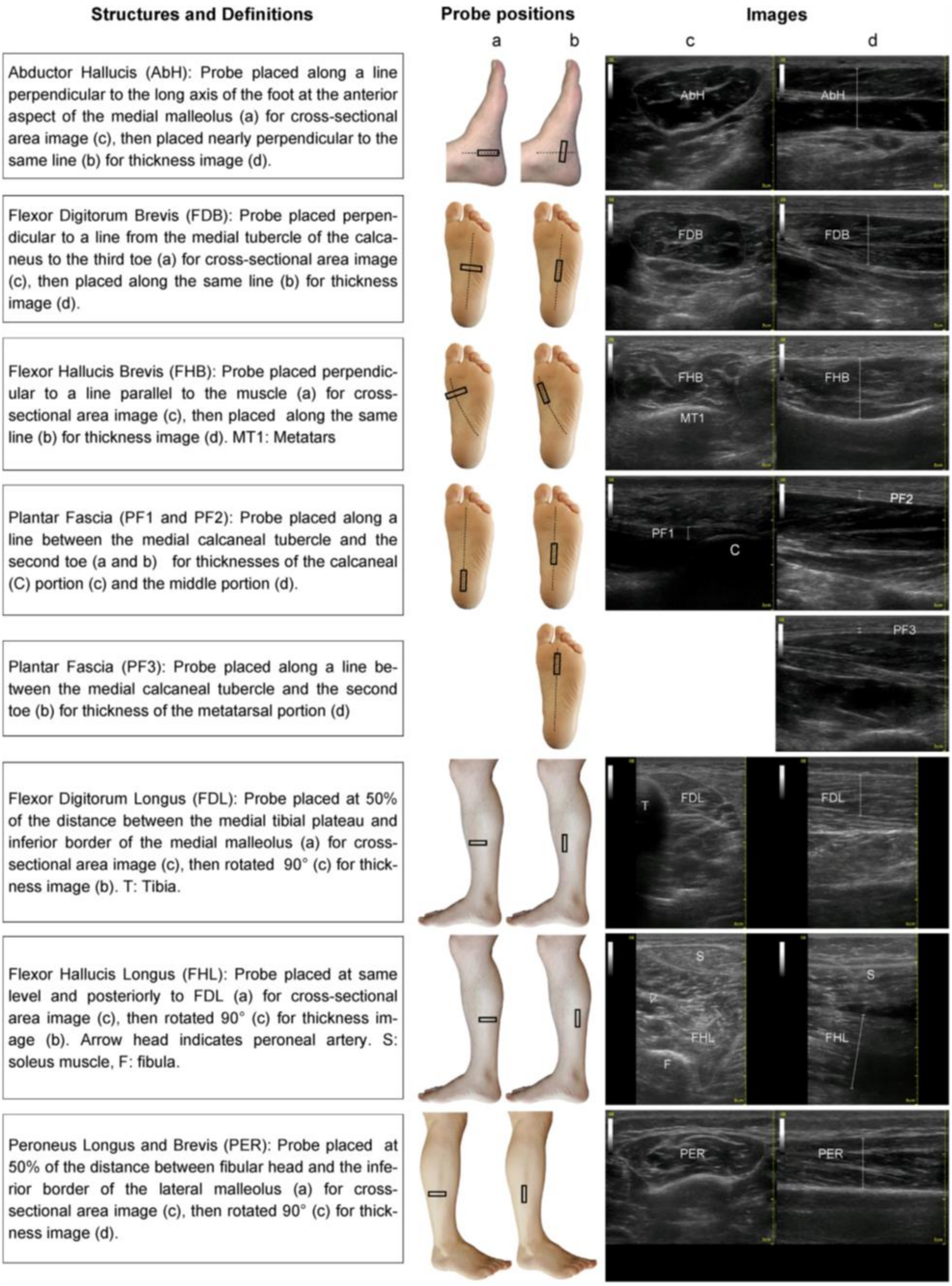


Figure 7-2. The scanned structures and scanning protocol definitions with probe position, and corresponding sample images (Angin et al., 2014).

7.2.7 <u>*Skin sensitivity*</u>

Skin sensitivity was tested whilst participants were prone and their feet were in a relaxed and neutral position (ankle at approximately 90°). The location of skin touch sensitivity is shown in **Figure 7-3** with an additional site on the dorsum, at approximately the level of the distal interphalangeal joint between the hallux and second toe. The afferent firing response of cutaneous mechanoreceptors in the foot sole has shown to be very variable below 20°C (Lowrey, 2012). Skin temperature was therefore measured at each site using a handheld infrared thermometer (Brannan Thermometers, Cumbria, UK) or (Thermoworks, USA.). If skin temperature fell below 20°C the participant was asked to put their sock back on to rewarm the foot.

7.2.7.1 *Monofilament testing*

Perceptual threshold was measured using Semmes-Weinstein monofilaments. They are filaments of different diameters that were pressed against the skin until they bend, with the diameter related precisely to the force being applied (across a range of 0.008 g to 300 g). A 3-2-1 countdown was given before the monofilament touched the skin. Each monofilament was applied for 1.5 seconds. The participant responded with a “yes” or “no” to indicate perception of the stimuli or not. The staircase, or stepping, method was used to determine sensitivity threshold (Dyck et al., 1993), whereby the 0.6 g monofilament was initially applied and then the force would be increased if not perceived, or decreased if perceived, by two steps on the scale. Once the stimulus could be detected (increasing force) or not detected (decreasing force), then the direction on the scale would be reversed and the force would be increased or decreased by one step. The sensitivity threshold at each site was the smallest diameter monofilament which could be detected with an accuracy of 75% (correct on ¾ occasions). Anticipation bias was minimised using a couple of random catch trials, in which a monofilament would not be applied to the skin.

Figure 7-3. The foot sole with blue dots depicting the approximate locations of skin sensitivity testing. An additional site was on the dorsum at approximately the level of the distal interphalangeal joint between the hallux and second toe.

7.2.7.2 *Vibration perception*

Vibration perception was measured using a Horwell Neurothesiometer (Scientific Laboratory Supplies Ltd., Nottingham, UK) operating at 50 Hz. The amplitude of the vibration was increased until the participant indicated they could just barely feel it. The test was performed three times on each site, with a few seconds between application (**Figure 7-3**) and the median value used for analysis. The median was taken as the measure of central tendency as unlike the mean, the median is not sensitive to extreme values, which could have occurred while becoming accustom to the task or through lapses in concentration. As the stimulus of the neurothesiometer continuously increases until perceived, this method can be categorised as a “method of adjustment”, one of the classical psychophysical methods of determining an absolute threshold (Pelli and Farrell, 1995).

7.2.8 <u>*Foot posture assessment*</u>

Participants’ static foot type was classified using the FPI at the first session. For data collected at the University of Salford, the FPI was assessed by an investigator with five year’s podiatry experience. At the University of Guelph photographs were taken of the participants’ feet of the relevant features that make up the FPI for later verification of the assessment by the same assessor at the University of Salford.

7.2.9 *Compliance with foot orthoses use*

In addition to the weekly emails checking the condition of the FO and if the participants were wearing them, the following questions were asked of participants after the three months of FO use:

- Out of 24 hours, how many hours a day did you typically wear the insoles in the first week?
- Out of 24 hours, how many hours a day did you typically wear the insoles after the first week?
- If there were days when you did not wear the insoles at all, how many days out of 7 might that have been?
- Did you wear the insoles in activities other than everyday walking? Please give examples.

7.2.10 *Data analysis*

7.2.10.1 *Ultrasound*

Ultrasound images were blinded, so that the single assessor performing the analysis was unaware as to whether the image came from a participant in the FO or control group. All images were measured using ImageJ software (NIH, USA). For thickness measurements, three measurements were taken per image between the deep and superficial aponeurosis in the central portion of the image (the muscle mid-belly for muscle measurements), as per a previous protocol (de Boer et al., 2008) (**Figure 7-4**) and for CSA two measurements were taken per image. With thickness it is possible to take three measurements from the one image. However, for CSA it was necessary to save over each raw image twice, so only two CSA measurements per original image were taken in the interest of time efficiency in analysis, especially as CSA from image one and two were highly correlated (e.g. r= 0.98 for ABH). Means were then calculated from the four measurements for CSA and six measurements for the thickness respectively. Median values of the differences within participants over time were reported as the median is less sensitive to extreme values. It was thought that measurement error in one session could lead to an extreme change which would affect the average value less if the median was used as the measure of central tendency than the mean.

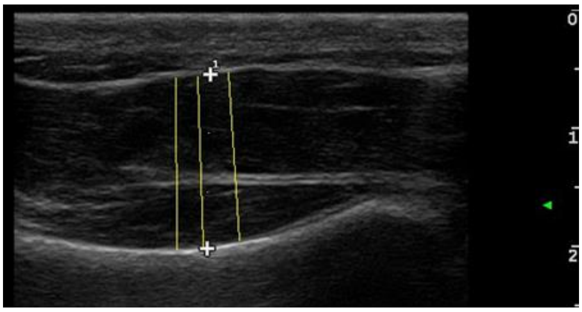


Figure 7-4. Ultrasound image of the flexor hallucis brevis with three thickness measurements (yellow lines)

7.2.10.2 *Plantar pressure*

Plantar pressure data were analysed in MATLAB using the InShoePressureAnalyser script © University of Salford 2010, Version 1.1 (16/012/10). Masks were created for the medial and lateral arch and medial and lateral heel (**Figure 7-5**). The arch region was taken to occupy the middle third of the footprint (Cavanagh and Rodgers, 1987). Peak plantar pressure was identified as the maximum pressure over the trial. Analysing multiple plantar pressure variables was deemed unnecessary given that typical plantar pressure outcomes like peak pressure, mean pressure and pressure-time integral were highly correlated (Keijsers et al., 2010). Peak pressure was chosen for analysis as it is the most commonly reported of plantar pressure parameters (Melvin et al., 2014) and it has been suggested that the most useful measure clinically is the highest pressure in each region of the foot during any point in stance (Cavanagh and Ulbrecht, 1994).

As there were no significant differences between the left and the right plantar pressure measurements ($p>0.05$), analysis was limited to the right foot, as per all other data in this thesis. The Pedar insoles used to record from the University of Guelph group were new, and as it was indicated that the accuracy of Pedar insoles could be reduced after a year of use (Hsiao et al., 2002), which could result in a difference in recording capacity of the systems at the two sites. Independent t-tests were run to compare the University of Salford and University of Guelph groups on the percent differences between the flat inlay and FO condition on the right foot and there were no significant differences ($p>0.05$). Consequently, the two samples were pooled together.

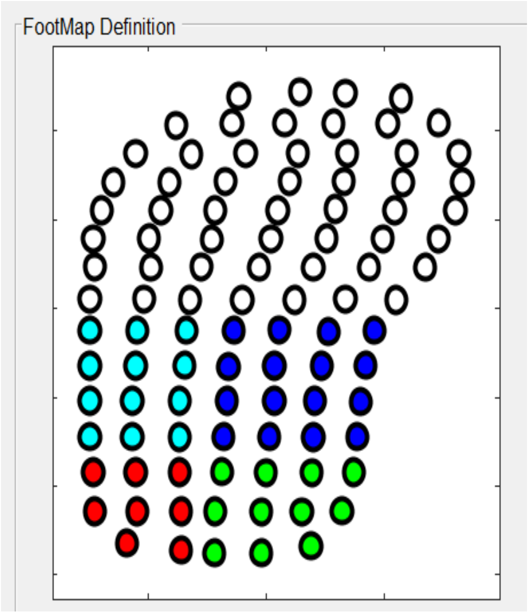


Figure 7-5. Mask definitions in MATLAB. Light blue: lateral arch; dark blue: medial arch; red: lateral heel; green: medial heel

7.2.10.3 *Statistics*

Statistical analysis was performed with SPSS (IBM SPSS Statistics 25) and Excel (Microsoft Office Excel 2013). A visual inspection of the histograms and box plots of data was used to establish normality and identify outliers. Outliers were determined based on the interquartile range (IQR). If a value was beyond the first quartile -3*IQR or the third quartile +3*IQR it was determined to be an outlier and the participant was excluded for that variable.

To test the hypotheses that there would be increases in the thickness and CSA of soft tissues and site-specific changes in skin sensitivity after three months of FO use, the following tests were performed. For mean soft tissue measurements, monofilament scores and neurothesiometer values two-way mixed model ANOVAs were used with a paired factor of time (pre and post three months) and an unpaired factor of group (FO and control). The monofilament scores were non-normally distributed, so the values were log-transformed (base 10) before performing the ANOVA. A Friedman's test was not performed because an assumption of a Friedman's analysis is that the groups have to be measured three or more times (Laerd, 2018). For all statistical tests, level of α was accepted as 0.05 to determine significance.

To test the hypothesis that the effect of FO on peak pressure in the medial arch and the medial heel in the standard shoe would be similar at baseline and the three-month follow up the percent differences between the flat inlay and the FO in the standard shoe were analysed with a two-way mixed model ANOVA with factors time (pre and post) and group (FO and control). A possible interaction effect between time and group was of primary interest as it would indicate that the difference between pre and post intervention measurements varied between the FO and control group.

7.3 Results

7.3.1 *Participants*

Twelve of the total 53 participants recruited dropped out, four of whom were in the FO group (one firm thermoplastic FO and three medium density FO) and eight of whom were in the control group. One participant in the FO group dropped out because she experienced hip pain which she attributed to the FO. Several participants self-reported experiencing some foot soreness when initially using the FO, but this did not persist beyond the first week and is typical when becoming accustom to FO (Woodburn et al., 2002).

Characteristics of participants (including those who dropped out) are presented in (**Table 7-1**). There were more female than male participants in both the FO and control groups.

Table 7-1. Mean participant (±SD) characteristics

		Age	(±SD)	Sex		Height (m)	(±SD)	Mass (kg)	(±SD)	Shoe size	(±SD)	Leg dominance		FPI L	(±SD)	FPI R	(±SD)
Pre	FO (n=26*)	27	7	F=16	M=10	1.680	0.078	68.1	13.2	7	2	L=0	R=24$	3	4	3	4
	CO (n=26)	30	11	F=19	M=7	1.662	0.069	69.2	12.7	7	2	L=2	R=24	2	5	2	4
	Total (n=52*)	29	9	F=35	M=17	1.671	0.074	68.6	12.9	7	2	L=2	R=48$	2	4	2	4
Post	FO (n=23)	28	10	F=15	M=8	1.651	0.070	67.8	13.3	7	2	L=0	R=23	3	4	3	3
	CO (n=18)	28	10	F=12	M=6	1.656	0.069	68.7	13.1	7	2	L=0	R=18	3	4	3	4
	Total (n=41)	28	9	F=27	M=14	1.673	0.074	68.1	12.2	7	2	L=0	R=41	3	4	3	4

F= female; M= male; L= left; R= right; SD= standard deviation; FPI= foot posture index; FO= foot orthoses group; CO= control group

*[*should be 27/53 but missing data on P40, $missing data n=2]*

7.3.2 *Compliance*

Sixteen of the 23 participants in the FO group completed all the compliance questions after the three-month intervention. The seven participants who did not complete the structured compliance questions had been responding positively to the follow up emails throughout the intervention that they were wearing the FO throughout the day. Except for one participant who reported only wearing the FO for two hours a day, all participants who completed the compliance questions wore the FO for the minimum of four hours a day requested and more (approximate mean ±SD: 8.5 ±3 hours).

7.3.3 *Physical activity levels*

In the FO group 21 of 23 (91%) were considered active pre-intervention and 22 of 23 (96%) were considered active post intervention. Within individuals, one participant was considered active who was previously considered inactive. In the control group 14 of the 18 (78%) were considered active both pre and post intervention.

7.3.4 *Ultrasound*

Ultrasound measurements pre and post intervention are presented in **Table 7-2** and **Figure 7-6** and **Table 7-3** and **Figure 7-7** for thickness and CSA respectively. The differences presented are the median of the differences over time within an individual. No average difference in US measurement was greater than the MDD (**Table 7-2** and **Table 7-3**). There was a trend for an effect of time for ATMID thickness ($p=0.056$, $\eta p^2=0.103$) and a main effect of group for ATMID CSA ($p=0.049$, $\eta p^2=0.127$). All other effect sizes were <0.1.

Table 7-2. Mean ultrasound thickness measurements at pre and post intervention

Thickness (mm)		Pre Mean	Pre SD	Post Mean	Post SD	Difference (median)	MDD
ABH	FO (n=19)	10.639	2.268	10.736	1.720	0.037	0.742
	CO (=16)	10.491	1.739	10.568	2.134	0.092	
ATINS	FO (n=19)	4.339	0.680	4.348	0.657	0.001	0.323
	CO (n=17)	4.245	0.581	4.259	0.517	0.076	
ATMID	FO (n=19)	3.577	0.781	3.773	0.831	0.289	0.451
	CO (n=17)	3.922	0.882	4.009	0.788	0.034	
FDB	FO (n=19)	8.916	1.437	9.304	1.575	0.581	2.138
	CO (n=17)	9.892	2.101	9.745	1.542	0.033	
FHB	FO (n=16)	11.989	2.602	11.682	2.608	0.151	1.032
	CO (n=12)	12.902	2.840	11.688	3.352	-0.031	
PFINS	FO (n=20)	2.178	0.457	2.286	0.380	0.070	0.495
	CO (n=17)	2.204	0.448	2.100	0.407	-0.018	
PFMID	FO (n=20)	1.652	0.193	1.667	0.260	-0.010	0.302
	CO (n=17)	1.600	0.151	1.629	0.183	0.036	

FO= foot orthoses group, CO=control group

Table 7-3. Mean ultrasound cross-sectional area measurements at pre and post intervention

CSA (mm^2)		Pre Mean	Pre SD	Post Mean	Post SD	Difference (median)	MDD
ABH	FO (n=17)	204.332	62.368	203.705	57.624	5.232	7.229
	CO (n=15)	198.601	53.211	197.935	48.367	1.062	
ATINS	FO (n=17)	68.460	13.890	69.971	12.591	1.808	11.963
	CO (n=16)	66.365	16.741	67.263	14.149	-2.716	
ATMID	FO (n=15)	53.495	13.717	55.527	9.472	1.829	14.169
	CO (n=16)	47.986	7.826	51.044	9.347	2.108	
FDB	FO (n=17)	200.401	44.741	204.332	46.820	2.086	40.360
	CO (n=14)	195.531	55.193	199.157	57.221	-0.790	

FO= foot orthoses group, CO=control group

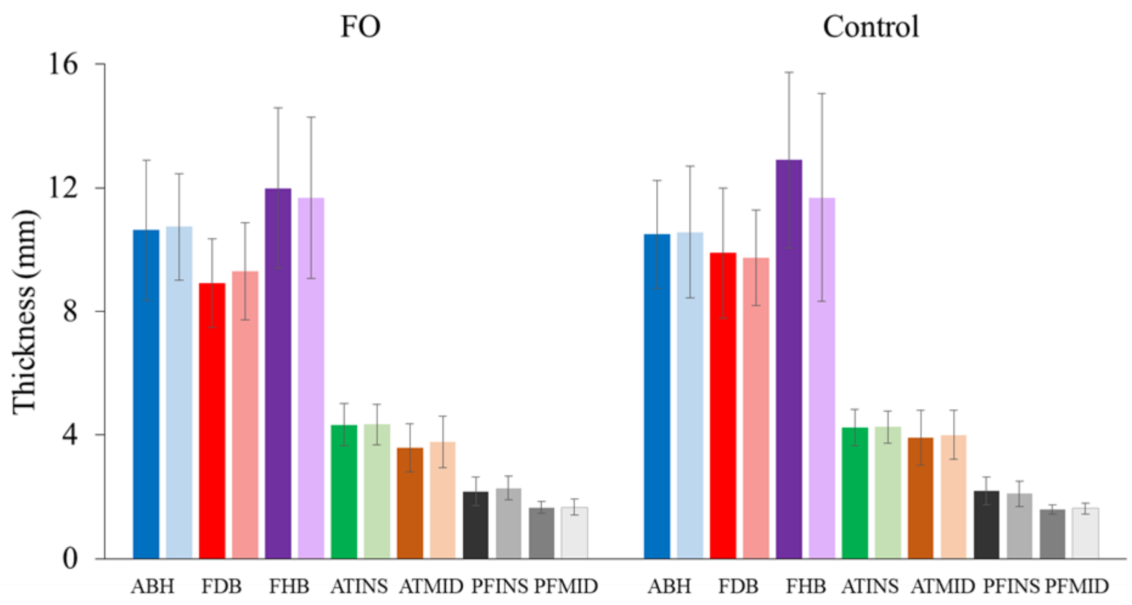


Figure 7-6. Mean ± SD of ultrasound thickness measurements for the foot orthosis group (FO) and the control group for the abductor hallucis (ABH) n=35, flexor digitorum brevis (FDB) n=36, flexor hallucis brevis (FHB) n=28, Achilles tendon at insertion (ATINS) n=36, Achilles tendon at mid-portion (ATMID) n=36, plantar fascia at insertion (PFINS) n=37 and mid plantar fascia (PFMID) n=37. Darker colour bars represent pre-intervention and lighter colour bars represent post intervention

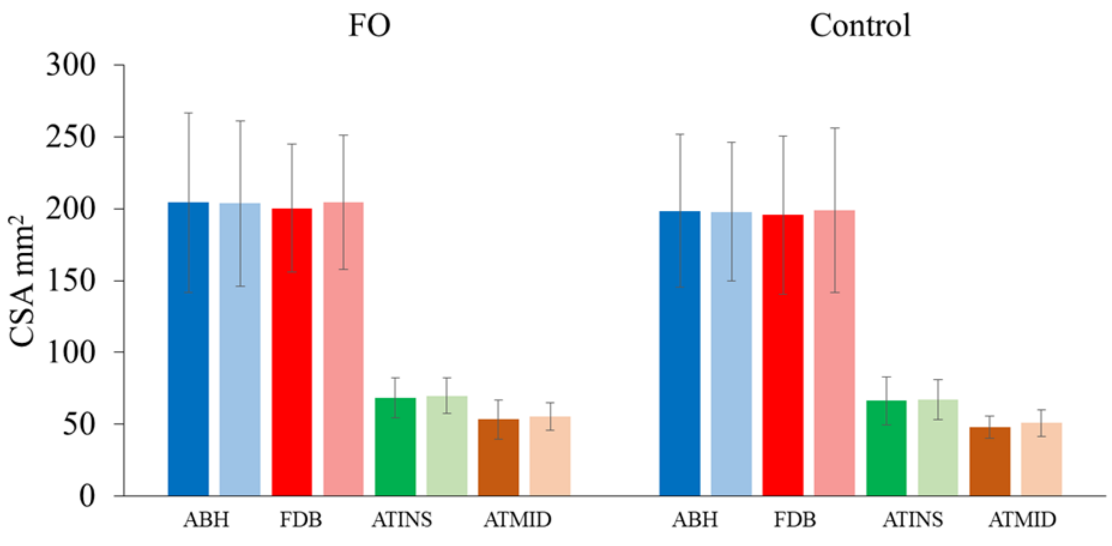


Figure 7-7. Mean ± SD of ultrasound cross sectional area (CSA) measurements for the abductor hallucis (ABH) n=32, flexor digitorum brevis (FDB) n=33, Achilles tendon at insertion (ATINS) n=31 and Achilles tendon at mid-point (ATMID) n=31. Darker colour bars represent pre-intervention and lighter colour bars represent post intervention

7.3.5 Skin sensitivity

Skin temperature was mean ± SD: 26.3°C ±1.6°C pre and 29.8°C ±1.3°C post intervention.

7.3.5.1 *Monofilaments*

The monofilament results are presented in **Figure 7-8**. There was a main effect of time for the 1st metatarsal head, in that the mean monofilament threshold increased with time for both control (0.34 ±0.53 to 0.60 ±0.55 g) and intervention groups (0.27 ±0.42 g to 0.55 ±0.62 g, $p=0.003$, $\eta p^2=0.211$). There were no significant effects at any of the other regions of the foot in either group with effect sizes <0.1. There was large variability across sites, which can be seen from the SD error bars.

7.3.5.2 *Neurothesiometer*

There was no change in vibration perception over time (**Figure 7-9**). There was a trend towards an interaction effect for the 1st metatarsal head ($p=0.092$, $\eta p^2=0.109$), this was due to a decrease in vibration perceptual threshold from pre to post for the control group (2.5 ±1.0 to 1.5 ±0.5) whereas there was no change for the intervention group (2.5 ±1.0 both pre and post).

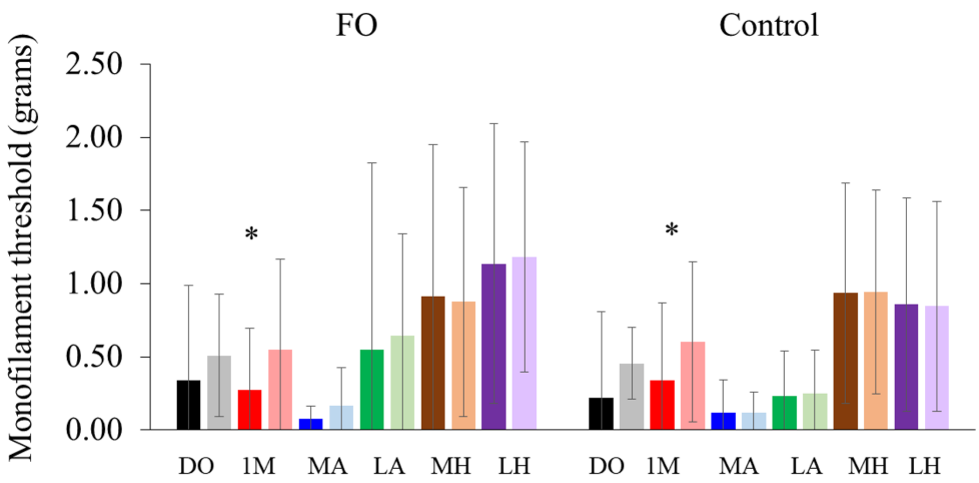


Figure 7-8. **Mean ± SD monofilament threshold (grams) for the foot orthoses group (FO, n=23) and control group (n=18). DO= dorsum, 1M= 1st metatarsal head, MA= medial arch, LA= lateral arch, MH= medial heel, LH= lateral heel. Darker colour bars represent pre-intervention and lighter colour bars represent post three-month intervention respectively, * denotes significant main effect (log10 transformed data, p<0.05)**

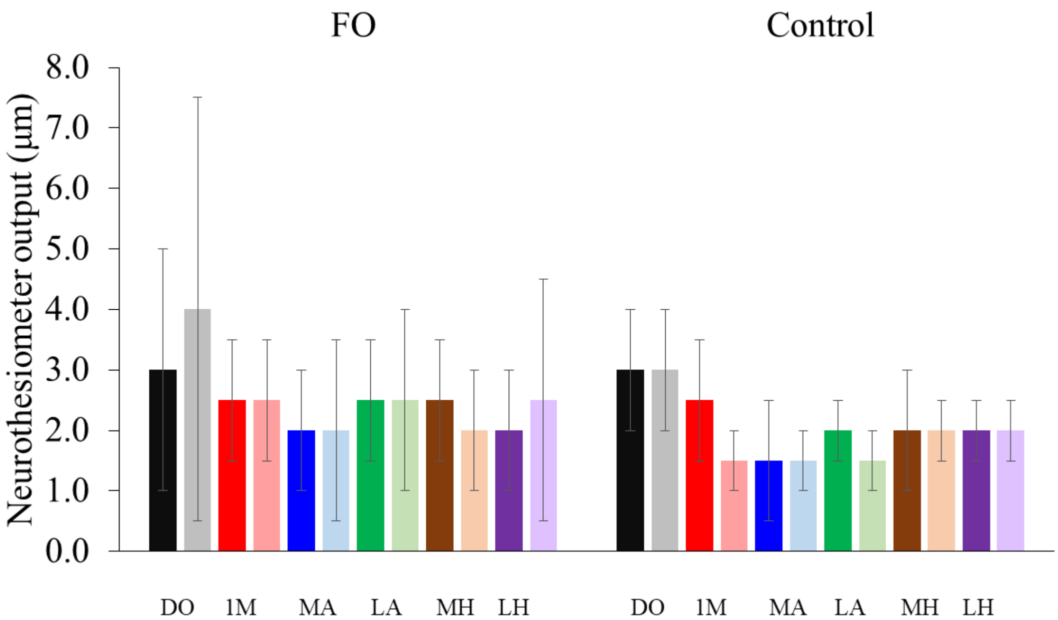


Figure 7-9. Mean ± SD neurothesiometer output (micrometre) for the foot orthoses group (FO, n=16) and control group (n=11). DO= dorsum, 1M= 1st metatarsal head, MA= medial arch, LA= lateral arch, MH= medial heel, LH= lateral heel. Darker colour bars represent pre-intervention and lighter colour bars represent post three-month intervention respectively.

7.3.6 *Peak plantar pressure (standard shoe)*

The percentage changes in peak plantar pressure between the flat inlay and the FO condition pre and post intervention are presented in **Figure 7-10**. There was a significant interaction effect for the medial arch (p=0.004). For the FO group the peak pressure of the medial arch increased from 99 ±33 kPa with the flat inlay to 115 ±25 kPa with the FO at pre-intervention, but there was virtually no change at post intervention (122 ±44 kPa with the flat inlay to 124 ±30 kPa with the FO). However, for the control group there was a greater increase in peak pressure of the medial arch at post intervention (113 ±28 kPa with the flat inlay to 121 ±26 kPa with the FO) than that at pre-intervention (114 ±33 kPa with the flat inlay to 133 ±32 kPa with the FO).

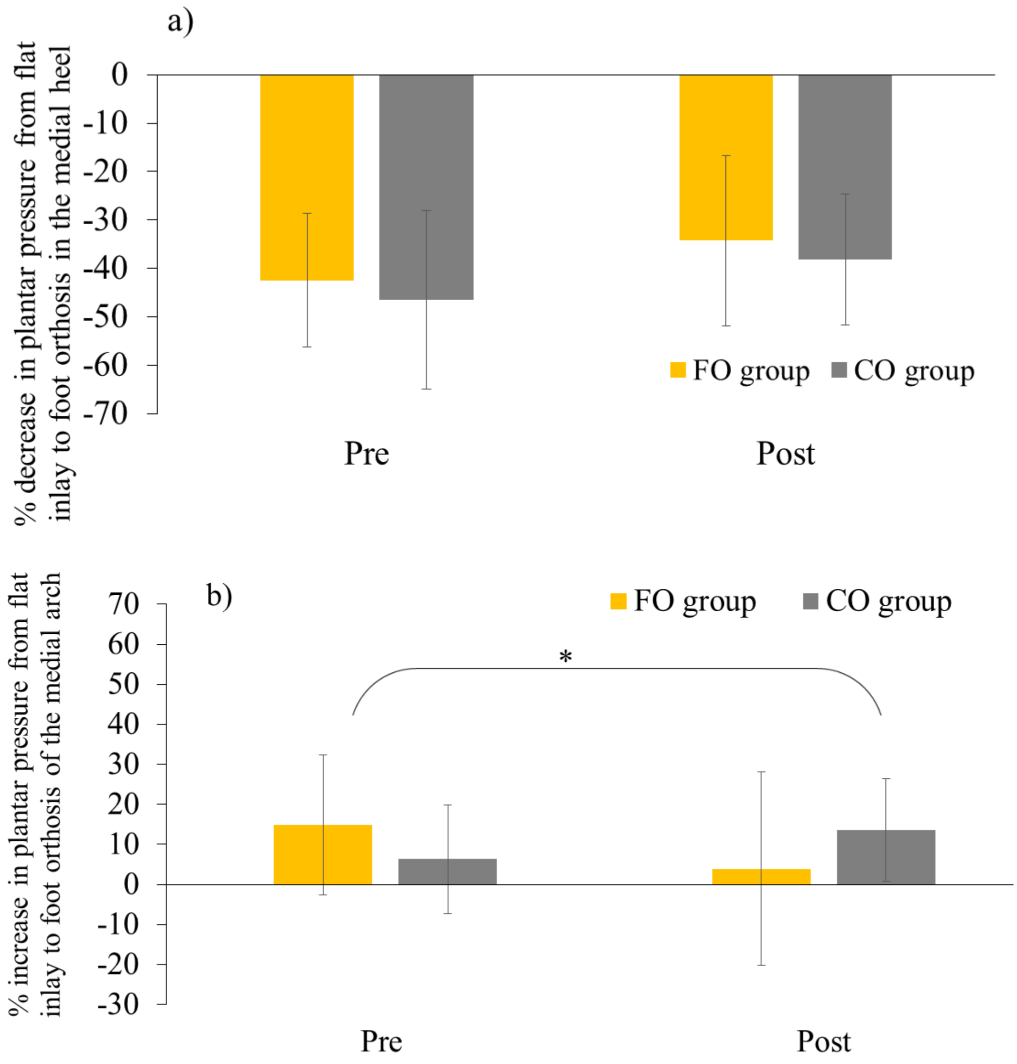


Figure 7-10. Mean ± SD of the percent differences in peak pressure of medial heel (a) and medial arch (b) between two conditions, i.e. flat inlay and FO in the standard shoes, at pre- and post-intervention (n=36). Yellow bars represent the foot orthosis group (FO) and grey bars represent the control group (CO). * represents a significant interaction effect of group and time (p<0.05)

7.4 Discussion

To explore the effect of FO over time on soft tissue in relation to joint loading, this study measured the thickness and CSA of muscle and connective tissue around the foot and ankle, skin sensitivity and peak plantar pressure.

7.4.1 *Recap of hypotheses*

Hypothesis 7.3.1 was that, following use of FO over three months, there would be increased CSA and thickness of ABH, FHB and Achilles tendon, and increased thickness of the plantar fascia, in the FO group.

Hypothesis 7.3.2.1-3 were that skin sensitivity would increase in the medial arch and decrease at the medial and lateral heel while other regions would remain unchanged following use of FO over three months.

Hypothesis 7.3.3 was that the increase in peak pressure in the medial arch and decrease in peak pressure in the medial heel when using FO would remain consistent from pre to post intervention.

7.4.2 *Soft tissue measures*

Hypothesis 7.3.1 can be rejected as no interaction effects were identified, which meant the FO intervention did not affect the thickness and CSA of ABH, ATINS, ATMID and FDB, and thickness of FHB, PFINS and PFMID. A lack of change in soft tissue morphology in the FO group could reflect a true lack of effect of the FO on intrinsic foot morphology and connective tissue, or simply that any change was below the level of measurement error. Alternatively, effects could have occurred in structures that could not be measured, such as deeper intrinsic foot muscles, or in some other property of a muscle, such as fibre characteristics or fibre recruitment patterns, that would not affect morphology. It might also be that the use of FO, i.e. hours per day and therefore the dose of mechanical change imposed, was insufficient to drive changes in morphology. However, the average reported wear time of the FO was 8.5 ±3 hours a day, which represented the length of the working day, which was arguably close to what could be expected of patients in clinical practice and comparable to previous research (McPoil et al., 2011, Munteanu et al., 2015). The changing demand on the neuromuscular and sensory systems imposed by altered loading with the FO might be within the normal scope of muscle and mechanoreceptor behaviour and so not enough to change soft tissue morphology or sensory function.

A lack of change in the FO group in this study conflicts with a previous intervention study (published at a conference) with customized FO, whereby CSA decreased in the FO group and not the control group. In that work, involving a small sample of five in the FO group and five in the control group, FDB reduced from 209 ±50 mm^2 to 189 ±46 mm^2 (-9.6%) and ABH from 138 ±51 mm^2 to 114 ±38 mm^2 (-17.4%) over 12 weeks of use (Protopapas and Perry, 2018). However, that study did not report MDDs and the reduction in FDB CSA was smaller than the MDD in this thesis (40.360 mm^2), which meant the reported change might be error of measurement. The reduction in ABH CSA with customized FO was however greater than the MDD in this thesis (7.229 mm^2). The results from this thesis and the reliability study by Crofts et al (2014) showed that the reliability of muscle thickness measures was better than CSA measurements. For example, with limits of agreement being 13% and 13.5% for ABH and FDB thickness respectively, and 16% and 17% for ABH and FDB CSA (Crofts et al., 2014). As such it could be argued, especially in the absence of reliability data for the work by Protopapas and Perry, that less confidence might be placed on the CSA measurements than the thickness measurements, which were collected in the earlier study but not reported.

As only a conference abstract, the study by Protopapas and Perry (2018) had additional limitations, as it failed to report details about how many ultrasound images were taken per session and how many measurements per image, the wear time of the FO, and whether physical activity levels of their participants changed over the 12 weeks. For the majority of participants in this study the physical activity levels of the participants were consistent from pre- to post-intervention, so it was unlikely that there was a substantial change in the internal loading of the soft tissues tested that would have confounded the results. The average FO wearing time in this study equated to ~59.5 hours per week, which were similar to the 57.5 ±25.8 hours per week reported in a previous study, which was a three-month intervention study with prescribed customised FO for Achilles tendinopathy (Munteanu et al., 2015) and longer than the 40.6 hours per week reported in another study that was a three-week follow-up study on the customised FO use for patellofemoral pain (McPoil et al., 2011). Several participants in this study reported not wearing the FO some days on weekends when they stayed indoors for most of the day and not wearing shoes. Not wearing the FO one day a week likely reflects clinical practice. In a 30-month RCT of custom FO for patients with rheumatoid arthritis, the average wearing time was 6.3 ±3.5 hours per day and 6.1 ±1.9 days per week (Woodburn et al., 2002).

It is possible that use of customised rather than prefabricated FO may explain the difference in outcomes between the work of Protopapas and Perry (2018) and this study. However, prefabricated and customised FO have been shown to have similar effects on peak plantar pressure and can be equally effective clinically in reducing pain symptoms (Almeida et al., 2009, Redmond et al., 2009). However, without knowledge of the change in plantar pressures or the material of the FO in the study by Protopapas and Perry (2018), we cannot compare the changes, or doses, in loading achieved in the two studies. In the study in this thesis the mean percent decreases in peak pressure with prefabricated FO at the medial and lateral heel (21% ±14% and 17% ±14% respectively) were greater than that reported previously with custom FO (13% and 7% for the medial and lateral heel respectively (McCormick et al., 2013)). However the percent increase in peak pressure at the medial arch in the study in this thesis was very similar to the previous work with a custom FO: 15% ±19% vs. 15% (McCormick et al., 2013). The custom FO in the study by McCormick et al. were designed to reflect common clinical practice. The comparable changes in peak plantar pressure in this thesis compared to previous work would suggest that the changes in regional distribution of plantar pressure were realistic.

7.4.3 *Skin sensitivity*

Mean skin temperature increased by approximately 3.5°C from pre to post intervention, possibly due to a change in seasons and warmer air temperature. However, a range of mean ±SD skin temperature between 26.3°C ±1.6°C pre and 29.8°C ±1.3°C post intervention respectively is similar to previously reported ranges of baseline skin temperature (Lowrey, 2012, Strzalkowski et al., 2015a, Sun et al., 2005). Therefore the increase in mean skin temperature from pre-post intervention was not considered meaningful and likely did not impact on results.

Hypotheses 7.3.2.1-3 can be rejected because there was no change in monofilament scores or neurothesiometer values. It would appear that the plantar pressure changes induced by the FO were insufficient to alter skin sensitivity. Alternatively, any changes were below the minimum detectable difference of the methods used to measure skin sensitivity and the large variability in the sample. For instance monofilament testing has demonstrated low reliability (ICC 0.46-0.61) in skin on the back (Ellaway and Catley, 2013). Additionally, while useful as a clinical research tool, the neurothesiometer has been shown to have poor reliability for research purposes (Schlee et al., 2012).

It is possible that as the participants were healthy, mostly active, young participants, free from foot and ankle pathology and neurological deficits that monofilament thresholds did not change because participants had little to gain in skin sensitivity. Baseline monofilament threshold in the medial arch was low (~0.1 g vs. ~1.0 g at the heel), so it may be that increased loading in this area was not able to increase sensitivity because it was already very sensitive. Individuals with deteriorated skin sensitivity on the other hand, with an increased stimulus threshold for activation of cutaneous mechanoreceptors, might benefit from increased plantar loading and increased repeated stimulation of receptors. The lack of change in skin sensitivity from this study was opposite to the study by Vie et al. (2015) that found skin sensitivity increased in the forefoot when wearing a hard metatarsal pad, but not with a soft pad. Differences between the studies could be attributed to the different methods of measuring skin sensitivity (a bespoke device that applied a mechanical load in the previous study versus monofilaments and a neurothesiometer in the present study). However, the stimulus was likely more extreme in the study by Vie et al. than the present study as most participants in the current study received a low or medium density FO, as supposed to the hard (presumably denser) material that had the effect on skin sensitivity in the earlier study. Also, the elevated pressure in the study by Vie et al. was placed close to the ball of the foot, an existing weight bearing area, and perhaps changes in skin sensitivity offer added value here but not elsewhere under the foot.

It is also conceivable that changes in other aspects of skin sensitivity changed with FO use that were not captured with the methods used in this study. Both the monofilament threshold and neurothesiometer testing would target FAIs (Johansson et al., 1982, Strzalkowski et al., 2015b), Vibration sensitivity has been measured before and after space flight (an extreme reduction in pressure due to microgravity) and no difference in sensitivity to 60 Hz vibration, designed to target FAIs, was detected (Lowrey et al., 2014). There was however decreased sensitivity to 3 Hz and 25 Hz at the great toe and increased sensitivity at the heel at 250 Hz in a subset of astronauts. No change in perceptual threshold measured with monofilaments was reported following space flight. The approximately 20% decrease in pressure at the medial heel in this study would presumably be more subtle than the reduction due to microgravity, therefore it would not be surprising that if monofilaments were not sensitive enough to detect changes following space flight that they were also not sensitive enough to detect changes in skin sensitivity following a FO intervention.

Any effect of the FO on other cutaneous mechanoreceptors and corresponding skin sensitivity is unknown. For instance the SAI receptors detect texture, pressure, edges and curvature (Johnson, 2001). It is possible that the reduced pressure in the medial heel, the increased pressure in the medial arch and the curvature of the arch support of the FO used in this study could have changed the sensitivity of SAI receptors from pre to post wearing the FO for three months. A measure of spatial acuity, which would theoretically target SAI receptors, was developed using plastic domes with gratings (Van Boven and Johnson, 1994), named JVP domes after their inventors Johnson, Van Boven and Phillips (Tremblay and Marie-A. Wassef, 2000). The individual indicates whether the grating was applied to their skin in the vertical or horizontal orientation (Tremblay and Marie-A. Wassef, 2000). Although preliminary work using JVP domes on the foot exists (Apollinaro et al., 2018, Smith et al., 2018), JVP domes have yet to be validated for use on the foot. Participants reported "getting used to" the FO after around a week of use and that they then noticed the difference when wearing footwear without the FO. Anecdotally it would seem that some neurological adaptation to the redistribution of plantar pressure with the FO occurred, but this was not captured in monofilament and neurothesiometer testing.

7.4.4 *Plantar pressure (standard shoe)*

Hypothesis 7.3.3 can be partially accepted. For the medial heel the difference in peak pressure between flat inlay and FO was similar pre and post intervention in both groups. However, the percent difference in peak pressure in the medial arch between the flat inlay and FO was higher at pre-intervention than at post-intervention in the FO group, and higher at post intervention than at pre-intervention in the control group.

The greater effect of FO on peak pressure in the medial arch in the control group after three months was unexpected. As the control group continued to wear their regular footwear over the three months, we would not expect their underlying foot function and immediate response to FO to change. Given the large variability, the differing immediate effects of the FO on peak pressure in the medial arch between the FO group and control group post-intervention could be noise in the measurement.

The finding that relative changes in peak plantar pressure between flat insert and FO groups were generally consistent over time is in agreement with previous work that found that the effect of FO on regional peak pressures were similar before an intervention and after wearing FO for 4-6 weeks (Farzadi et al., 2015, Hodgson et al., 2006, McCormick et al., 2013). The use of the FO over three months did not fundamentally change foot function regarding the relative difference in plantar pressure distribution between a flat inlay and FO. It would also appear that the intended change in loading of the foot, designed to establish if changes in soft tissue morphology and skin sensitivity, was sustained.

7.4.5 *Limitations*

For ultrasound measurements the MDD was relatively high, especially for CSA, which could have impacted on the ability to detect an effect of FO. Less measurement error could be achieved with MRI, however this was not accessible or financially feasible.

The measurement protocol could have contributed to the variability in skin sensitivity results. Skin touch sensitivity monofilament testing has demonstrated low reliability (ICC 0.46-0.61) on the back (Ellaway and Catley, 2013). Additionally the monofilament testing sites were not as precisely defined as other used previously (Strzalkowski et al., 2015a). It is possible that different mechanoreceptors were targeted in the testing sessions before and after the intervention, especially as the receptive field sizes of FAIs which monofilament threshold and neurothesiometer testing would target are relatively small, approximately 50-60 mm^2 on average (Strzalkowski et al., 2015b). For the same reasons, different testing sites at baseline and after three months could have added additional variability to the data. However, the method we used was designed to give a practical measure of skin sensitivity of the functional regions of the foot, and similar to those used in clinical practice. Likewise the neurothesiometer test was performed as a quick and clinically relevant test of skin sensitivity. However, the test is subjective because of observer’s understanding of the criteria for detection of stimuli (Pelli and Farrell, 1995). The more complex method used by Vie et al. (2015) involved a bespoke loading apparatus with a fixed foot position was perhaps more objective and controlled than the methods in this study, however the apparatus was only validated for detecting stimulation of the fingers not feet (Balzamo et al., 1995). Additionally, there was still presumably still some degree of subjectivity in when the participant verbally indicated they perceived the stimulus, which was not described.

It is possible that nil effects were due to sample size, characteristics or group allocation. The sample recruited was pseudorandomised in that the sample at the University of Salford was randomised into FO or control group whereas the sample at the University of Guelph were allocated based on convenience, particularly based on sizes of the insoles available. Although entirely random allocation would have been preferred there were no statistical differences in participant characteristics between the FO and control groups for the sample collected at the University of Guelph which does not support the idea that a lack of randomisation contributed to a nil effect. As in Chapter 6, the nil effects observed are unlikely to be due to a lack of power, as effect sizes were typically less than small (≤0.1) for non-significant effects.. As muscle-tendon size and structure and skin sensitivity at the foot had not been studied longitudinally, there was no data on which to base an a-priori power calculation prior to recruitment. However, improvements in plantar pressure variables with FO use for 3 months was demonstrated with a sample of 30 (Zhai et al., 2016) which is less than the sample size of 41 who completed this study. Although on visual inspection no differences in responses were observed between foot types, recruitment did not target specific foot types and so the majority of participants had a neutral foot posture. Given the importance of sub-groups in certain conditions (Selfe et al., 2016), it could be argued that the mixed sample of foot postures masked a possible effect of the FO in a specific foot type sub group.

7.5 Conclusion

This study found no detectable change in soft tissue morphology or skin sensitivity after wearing FO for three months. Future study on the long term effects of FO on the neuromuscular system could consider using alternative measurement techniques and stratifying the sample based on sub groups. Although this study does not explain the mechanism of FO benefits, it challenges the belief held by some that FO make muscles smaller (and weaker).

Chapter 8 DISCUSSION

8.1 Summary

This chapter reiterates the aims of this thesis before highlighting the key findings. Each key finding is then discussed. Points of novelty and limitations are addressed, followed by future directions for research, implications for clinical practice and a final conclusion.

8.2 Review of thesis aims

Foot orthoses (FO) with a medial wedge or medial arch support are commonly used to treat musculoskeletal pathologies by altering external forces applied to the foot, which could consequently alter internal forces generated by muscles. However, little is known about whether systematic changes in FO geometry result in systematic changes in activation of lower limb muscles. The aim of this PhD was therefore to investigate the effects of FO on selected lower limb muscles.

Four studies were performed to answer the following research questions:

1) what is the level of evidence that footwear, FO and taping alter lower limb muscle activity during walking and running? The outcomes of a literature reviewed highlighted poor reliability of measuring peroneus longus (PL) and abductor hallucis (ABH) EMG signals with surface sensors was identified. Also, the reliability of the dynamic medial longitudinal arch (MLA) angle had not been established. Hence a second research question was posed:

2) is PL and ABH EMG data and MLA angle measurement reliable in walking?

The remaining research questions were specific to the effect of FO:

3) what are the immediate effects of variations in FO geometry on EMG of select muscles of the lower limb and foot and ankle biomechanics and

4) what are the long-term effects of FO on soft tissue morphology, plantar pressure and skin sensitivity.

Each of these will be discussed in turn highlighting the important advances in knowledge and also how these have shaped the thesis overall.

8.3 Key findings

- The literature suggests 1) Rocker shoes decrease tibialis anterior activity and increase triceps surae activity; 2) foot orthoses may decrease activity of tibialis posterior and increase activity of peroneus longus; 3) other footwear and taping effects are unclear.

- PL EMG profile and discrete data variables are reliable when using ultrasound guidance for sensor location and small sensors. The general pattern of ABH activation and the range of motion (ROM) of the MLA are reliable.
- Foot orthoses with medial heel wedging can immediately and simultaneously, reduce external eversion moment and peak tibialis posterior (TP) EMG amplitude, with no effect on the EMG of other muscles of the lower limb including PL.
- Three months of FO use had no effect on selected foot and ankle soft tissue morphology, plantar pressure nor foot skin sensitivity.

8.4 Discussion of key findings

8.4.1 *The effect of footwear, foot orthoses and taping on muscle activity of the lower limb*

Specialised footwear, FO and taping are external devices that can all be used to treat some of the most common conditions of the lower limb and so evidence for their effect on lower limb muscle activity was reviewed together.

The review found that during walking rocker shoes decrease tibialis anterior (TA) activity and increase triceps surae (TS) activity in early stance. Rocker footwear like Masai Barefoot Technology (MBT), which is curved under the heel in the sagittal plane, unsurprisingly appear to exert their effects on muscles that act primarily in the sagittal plane. Tibialis anterior EMG amplitude has been found to be reduced in early stance in walking by ~30-40% ($p< 0.05$) with MBT versus flat heeled conventional footwear (Forghany et al., 2014, Price et al., 2013, Sacco et al., 2012), by contrast, the activity of medial gastrocnemius and soleus were shown to increase with some rocker shoes by 5.5-23% (Forghany et al., 2014, Koyama et al., 2012, Price et al., 2013). Rocker shoes shift the centre of pressure anteriorly at heel contact, increasing the external dorsiflexion moment, reducing the work required from TA and increasing the work from TS, accounting for the decrease and increase of EMG activity respectively (Forghany et al., 2014, Koyama et al., 2012, Price et al., 2013). Rocker shoes are thus proof that altering the external load applied to the foot can affect EMG activity of specific muscles of the lower limb.

There is some limited evidence that FO decrease activity of TP in early stance and increase activity of PL in mid-late stance, but there is otherwise little evidence in the literature of an effect of FO on lower limb muscle activity during walking. The evidence for significantly reduced TP activity with a FO, compared to a shoe alone, came from only one study (Murley et al (2010a)) that found peak TP amplitude reduced by 12-19% with prefabricated and custom FO respectively. The lack of research measuring the effect of FO on TP activity most likely stems from the need for invasive indwelling EMG to record from this muscle. The evidence for the effect of FO on PL EMG in walking is mixed, with some reports of increased activity, albeit not a linear response to heel wedging magnitude (Murley and Bird, 2006, Murley et al., 2010a). Other studies have found little effect of FO on PL amplitude (Akuzawa et al., 2016, Telfer et al., 2013a). The contrasting results may reflect differences in the FO design or material and whether or not that caused change in external joint moments. Although FO may increase amplitude and duration of PL activity during running (Baur et al., 2011, Kelly et al., 2011), the literature is limited by low between-session reliability of PL EMG (Barn et al., 2012, Moisan and Cantin, 2016, Murley et al., 2010b).

The review identified a number of other limitations to previous studies: no study investigated the effects on intrinsic foot muscle activity nor morphology over time; the effects of varying different aspects of FO geometry is unknown; few studies simultaneously recorded kinematics and kinetics with EMG to relate the effect on external forces to internal forces; and many studies had low statistical power. These limitations informed the rationale for the subsequent studies in this thesis. In Chapter 6 the immediate effect of FO on lower limb muscle activity was investigated, including activity of TP, and the effect of varying medial heel wedging and arch height, and kinematics and kinetics were recorded simultaneously with EMG data. As a precursor this, in Chapter 5, whether the EMG of PL and abductor hallucis (an intrinsic foot muscle) can be measured reliably with surface EMG was investigated. Chapter 7 investigated the long-term effects of FO on foot soft tissue morphology, including intrinsic foot muscle thickness and cross-sectional area and plantar fascia thickness.

8.4.2 *The reliability of peroneus longus surface EMG, abductor hallucis EMG and medial longitudinal arch angle*

Work in Chapter 5 found that PL EMG profile, peak amplitude and timing of the peak were repeatable between sessions during barefoot and shod walking. The protocol involved use of ultrasound imaging to identify the borders of PL and facilitate placement of the ENG electrode over the muscle belly, and the use of small EMG sensors (Delsys Trigno™ Mini). Both these approaches were intended to minimise cross talk from other muscles. The between-day reliability was better than in Barn et al. (2012), perhaps due to differences in power, study populations, and/or normalisation. The protocol used in Chapter 5 could be confidently used to study PL activation in repeated measures study designs, like interventions, but also increases the validity of findings in cross-sectional observations with one measurement session (such as in Chapter 6) as reliability underpins validity.

The overall pattern of ABH activation, which was assessed visually and with the coefficient of multiple correlation, was more consistent between days than peak amplitude and the timing of the peak. Therefore when using the Delsys Trigno™ Mini it may be more appropriate to study EMG of ABH qualitatively by visual inspection of the RMS signal over the gait cycle than with discrete variables. Reliability of discrete variables may have been improved by cutting holes in the shoe around the mini sensor head, which could have reduced possible artefact. A hole around the sensor head was not done in this thesis because several holes were already made in the shoe for kinematic markers to calculate MLA angle and it was thought that the shoe might lose its integrity if further holes were made in it and be less supportive. Study of ABH activation over time requires more consideration than that of PL, especially as there was large variability in the profile of ABH EMG between participants, which was similar to previous work in straight line walking (Reeser et al., 1983). In studying ABH EMG, the variable ability to perform an isolated contraction of this muscle should also be considered, especially if normalisation would be performed with respect to an MVC.

The pattern of MLA angle and the ROM was consistent between sessions. The drop in MLA angle recorded in late stance in barefoot is similar to the pattern in an earlier study of multi-segment foot kinematics barefoot (Leardini et al., 2007). The mean MLA angle in this thesis was around 140-150° which is less than the 170° in the study by Leardini et al., but more than the 131.5° ±11.8° in a fluoroscopy study (Balsdon et al., 2016). Given the difference in methodology between using x-rays and motion capture and possible variation in marker placement within motion capture, it seems appropriate to compare ROM between conditions rather than focusing on absolute values especially in relation to the literature.

As the PL EMG profile and discrete variables were repeatable between sessions, the protocol was used to study the immediate effect of FO in Chapter 6. For ABH the general activation pattern was reliable, but not discrete variables, so only the EMG profile was studied in relation to the immediate effect of FO. The ROM of the MLA was also reliable, so included in the kinematic variables assessed in Chapter 6.

8.4.3 *Foot orthoses with medial heel wedging can immediately reduce external eversion moment and peak tibialis posterior (TP) EMG amplitude, with no effect on the EMG of other muscles of the lower limb including PL*

Tibialis posterior activity decreased in early stance by 17% (p=0.001) with a FO with an additional 8 mm increase in medial heel wedging and by 14% (p= 0.047) with a FO with both a 6 mm increase in arch height and an 8 mm increase in medial heel wedging. The reduction in TP activity with FO with medial heel wedging was of a similar magnitude to the 12-19% reported with custom and pre-fabricated FO relative to shoes by Murley et al. (2010a). However, in a study published since conducting the literature review, there was no effect of a polypropylene customised FO with a 4-mm medial skive at 15° and a 5° extrinsic medial wedge on TP EMG (Maharaj et al., 2018). Unlike much of the previous literature, kinematics and kinetics were also recorded in this recent work, and OpenSim software used to scale a multi-segment foot model for each participant. There was no effect of the FO on the STJ displacement or STJ moment relative to the shoe condition. Therefore, it would appear the FO in question was not effective in altering external forces, which would account for a lack of change in the internal work required of TP and a lack of change in TP EMG.

The FO tested in this thesis had relatively extreme modifications that are unlikely to be routinely used in clinical practice. However, they serve as a proof of concept that it is possible to reduce TP activity with medial heel wedging. Future work could establish the minimal degree of medial heel wedging required to significantly reduce TP EMG activity, which could be informative for clinical practice. If the intention of FO was to treat TPTD then reducing TP activity would theoretically be beneficial because it would mean less force going through the TP muscle tendon unit, which could facilitate tendon healing. A reduction in TP activity in early stance in particular is clinically meaningful because the TP acts eccentrically in this phase. During eccentric contractions tendons act as mechanical buffers, absorbing energy and length changes which could damage the muscle (Konow et al., 2011), however this means the tendon itself is susceptible to strain-induced tendinopathy (Maharaj et al., 2018).

The reduced TP activity with medial heel wedging was accompanied by a reduced external ankle eversion moment, which provides a possible link between FO design, external forces and muscular demand. The reduction in TP EMG amplitude with the FO with increases in medial heel wedging and arch height was smaller than the reduction in peak eversion moment. However a difference in the magnitude of change between EMG and joint moment is not surprising given that there is not a linear relationship between muscle force and EMG data, likely differences in the moment arm of the TP and possible effects of the FO on energy storage in the TP tendon that could not be accounted for (which could influence eversion moment, but not be seen in EMG).

There was no effect of FO on PL, MG, TA nor ABH EMG. This is generally consistent with the literature review in Chapter 3. However, contrasting with this thesis, some studies have found an increase in PL activity with FO (Murley and Bird, 2006, Murley et al., 2010a) . The difference could be due to whether the FO increased the excursion of the centre of pressure in the mediolateral direction, making the participants more unstable. Alternatively, it could be due to differences in the method of recording PL EMG, as, unlike the method used in this thesis, other protocols have been shown to be unreliable (Barn et al., 2012, Murley et al., 2010b). Additionally, the presence of a fine-wire electrode in the TP appeared to have some influence on the relative effect of the FO on PL EMG in this thesis. The presence of a fine-wire electrode in both TP and PL in the study by Murley et al. (2010a) could potentially confound the effect of the FO on PL EMG, perhaps through restricting ROM or altering proprioceptive and/or nociceptive feedback from the muscle. Future work could consider comparing trials with and without fine-wire electrodes.

Although no effect of FO on ABH EMG was found, the result was somewhat inclusive. While the reliability study in this thesis indicated good reliability of the profile of ABH EMG using the Trigno™ Mini sensor, despite the potential for some artefact from shoe contact in late stance, the FO in Chapter 6 took up more volume in the shoe than the flat inlay in Chapter 5. The FO used in Chapter 6 with medial heel wedging and both medial heel wedging and increased arch support in particular took up substantial volume in the shoe, and interfered with the Trigno™ Mini sensor. It was necessary to have a small number of trials of each condition with the fine-wire electrode in TP, with the view to minimise collection time from TP due to the issue of a reduction in amplitude recorded by the fine-wire electrode overtime. Conditions were then repeated to obtain more trials with moment data. This effectively meant testing 10 conditions and repeatedly changing the shoes and FO, which disrupted the adhesion of the Trigno™ Mini sensor over ABH for many participants and limited the amount of quality data. Consequently, although there appeared to be an increase in ABH amplitude with all FO conditions in late stance, this was based on only nine participants, so we are unable to draw any substantial conclusions from this. If future work was specifically concerned with ABH using surface EMG rather than indwelling EMG, then it would be advisable to reduce the number of total trials (and not simultaneously recording from TP), or modify the testing footwear to have holes around the sensor.

Given the challenges of recording EMG of intrinsic foot muscles, future work might adopt alternative approaches to studying the response of muscles to FO. Indeed the design of the final study in this thesis, using ultrasound to investigate morphology changes in intrinsic foot muscles after 3months of FO use was a novel attempt to achieve this.

8.4.4 *Nil effect of three months of foot orthosis use on soft tissue morphology, plantar pressure and skin sensitivity*

The following US variables were measured prior to and after three months of FO use (or control): The thickness and cross-sectional area (CSA) of ABH; the Achilles tendon at the insertion on the calcaneus (ATINS) and the mid-portion (ATMID) and flexor digitorum brevis (FDB) and the thickness of flexor hallucis brevis (FHB); the proximal plantar fascia (PFINS) and the plantar fascia in the mid foot (PFMID). The US variables did not change over time in either the FO or control group. A lack of change in muscle morphology in the FO group could be a reflection of a true lack of effect of the FO on soft tissue morphology or the measurement approach was insensitive, especially given the high MDD for CSA measurements. Alternatively, effects could have occurred in structures or musculoskeletal properties not measured. It might also be that the duration of FO use was insufficient to drive changes in morphology. However participants typically wore their FO throughout the working day, arguably equivalent to patients in clinical practice. The changing demand on the neuromuscular and sensory systems imposed by altered loading with the FO may be within the normal scope of muscle and mechanoreceptor behaviour and so not sufficient to change soft tissue morphology or sensory function. No change in the FO group conflicts with a previous conference publication of an intervention with customized FO, whereby CSA decreased in the FO group for FDB and ABH (Protopapas and Perry, 2018), however MDDs were not reported, which were critical to interpreting the statistically significant differences in this thesis and concluding there was no evidence of a change due to the FO.

The strength of intrinsic foot muscles was not assessed directly in this study, however muscle size measured with ultrasound is a common indirect means of assessing intrinsic foot muscle strength (Soysa et al., 2012). It is reasonable to conclude, therefore, that as intrinsic foot muscle size did not change after three months of FO use, there were likely no changes in intrinsic foot muscle strength. There is no universally accepted gold standard of assessing foot muscle strength and it is difficult to isolate the intrinsic and extrinsic foot muscle contributions to measurements (Soysa et al., 2012). Using plantar pressure to assess strength, significant correlations were found between ultrasound measures of muscle size and strength (Mickle et al., 2016). Hallux flexor strength was significantly correlated with FHL CSA and thickness ($r = 0.515$-0.606, $p<0.003$) and lesser toe flexor strength was significantly correlated with FDB cross-sectional area and thickness ($r = 0.369$-0.501, $p<0.03$) and FHB thickness ($r = 0.552$, $p<0.001$). The lack of change in intrinsic foot muscle morphology with three months FO use thus provides indirect evidence to challenge the belief held by some that FO make foot muscles weaker.

Three months of FO use also resulted in no change in skin touch sensitivity (monofilament or neurothesiometer scores), which could mean the plantar pressure change was insufficient to alter skin sensitivity, any change in sensitivity was not detectable using the approaches used, or the change in load was simply within the natural variability in sensory function.

8.5 Novelty

The studies presented in this thesis have a number of novel elements. Firstly the eligibility criteria of the literature review was more specific to external devices intended to alter loading of the foot compared to the previous review by Murley et al. (2009b). The review in this thesis included only FO with a medial arch profile and or a medial heel/foot wedge, and footwear with modifications in the shape or material of the sole. In this way we have gained a better understanding of how modifications in shoe or FO design in the sagittal or frontal plane can alter activation of muscles acting primarily in those planes. Unlike the review in this thesis, Murley et al. also allowed comparisons to a barefoot control, which is less externally valid than a shoe only control. Another review included only MBT footwear and not other rocker shoes (Tan et al., 2016).

Secondly, novel protocols were developed in this thesis including use of ultrasound guidance and small sensors to reliably record EMG of the PL. This work has been accepted for publication (Reeves et al., 2019). The reliability supported its use in the subsequent study of the immediate effect of FO on lower limb EMG. In contrast, discrete PL EMG variables have been measured in the study of the effects of FO (Barn et al., 2014, Murley et al., 2010a) despite previously demonstrating poor reliability with the same protocols (Barn et al., 2012, Murley et al., 2010b). The overall data collection protocol in the immediate effects of FO study in Chapter 6 was also novel. To account for the drop in signal amplitude with time that was identified, only three trials of each condition were collected with the fine-wire electrode present in TP. Further trials with the electrode removed were then used to obtain more joint moment data. The issue of reduced recorded EMG amplitude over time was further explored in a spin off project in late 2018. With the fine-wire electrode there was a progressive drop in tibialis anterior amplitude by 30 minutes and a continued to drop until 50 minutes, while the amplitude of the surface electrode remained consistent over the 50 minutes. The methodological rigor of the approaches taken gives confidence in the validity of the findings that there was a reduction in TP EMG amplitude in early stance with medial heel wedging FO and a lack of effect of FO on PL EMG. Future research should be mindful of the drop in signal amplitude over time demonstrated with the prefabricated fine-wire electrodes used in this study.

Thirdly, few studies in the literature have simultaneously collected kinematics and kinetics with EMG when investigating the immediate effect of FO, except in research by Telfer and colleagues (2013a, 2013b). However, no TP EMG was collected in their work and only Maharaj et al. (2018) has reported this data. In this thesis a possible link between reduced external force (reduced ankle eversion moment) was found in conjunction with reduced peak TP amplitude using FO with medial heel wedging. Additionally, the effect of specific aspects of FO geometry on lower limb EMG had not previously been investigated. The most similar approaches to those in this thesis would be a comparison between pre-fabricated FO and custom FO by Murley et al. (2010a) and systematically increasing the degree of heel wedging from 6° lateral to 10° medial in 2° increments by Telfer et al. (2013a, 2013b). Tibialis posterior peak amplitude decreased by 19% with the pre-fabricated FO (with a 6 mm medial heel wedge) and 12% with customised FO (20° inverted wedge) (Murley et al., 2010a), which was a similar to the 14-17% decrease in peak TP amplitude with 8 mm medial heel wedging in this thesis. A lack of effect of FO on surface EMG of lower limb muscles in this thesis is in agreement with the findings of Telfer et al. (2013a).

Lastly, studies of the long-term effects of FO on muscles using ultrasound are rare and the effect of FO with medial arch support on skin sensitivity had not previously been investigated. The final study in this thesis took a mechanistic approach to determine the effects of wearing a FO that increased peak plantar pressure in the medial arch and decreased peak plantar pressure in the medial heel by at least 8%. Prescribing FO according to a minimum change in plantar pressure ensured that a change in external load relative to participants' conventional footwear was achieved, and is a novel approach in itself. An alternative approach would have been to ensure a dose response based on change in eversion moment. However, unlike the wedged FO, the standard FO design did not significantly reduce peak ankle eversion moment in the immediate effects study in this thesis. The standard FO used in the long-term effects study had increased arch support, but not medial wedging. The mechanism of therapeutic effect of the FO may relate to the redistribution of plantar pressure on the foot sole and be different to medial wedged FO. Therefore a threshold for effect of the FO based on plantar pressure was considered appropriate. Furthermore, assessing plantar pressure changes in response to a FO using a portable system was more practical than recording joint moments in a gait laboratory for the long-term study with a large sample size. A lack of FO studies delivering a "kinetic dose" was identified as a limitation to current FO research (Griffiths and Spooner, 2018). The kinetic (plantar pressure) immediate responses to FO was measured in both the FO and control groups in this thesis and were similar, which demonstrates that the control group can be considered true controls.

8.6 Limitations

None of the studies in this thesis included a clinical sample, which limits the generalisability of the findings to patient populations. It is possible that there could be a detectable effect of FO on muscle activity and soft tissue morphology in cases of foot pathology and/or deformity, which could be accompanied by weaker muscles than healthy individuals. The reason healthy participants were recruited was to investigate the mechanistic action of different FO geometries and to establish the potential of a change in mechanical load to alter neuromuscular function without the confounding effects of pathology.

Variation between and within clinical conditions could wash out any mean effects of FO in either the short or long term. For instance inflammatory cytokines have the ability to impede muscle function in patients with rheumatoid arthritis which is often associated with TPTD (Barn et al., 2012). The findings of immediate effects of FO study in this thesis are based on relatively young participants (mean age: 31 ± 7 years) and an older participant of 54 years of age was excluded as an outlier due to a relatively inactive TP during walking. If functional decline of TP with age is common it may limit the transferability of the results of this study to patients with rheumatoid arthritis, tibialis posterior tenosynovitis and acquired flat foot as these individuals are often middle-aged or older (Barn et al., 2013, Jordan et al., 2014).

The long-term effects of FO on muscle morphology and skin sensitivity may be different in patients experiencing pain compared to the effects on healthy, active individuals. As FO can be effective in treating common clinical conditions, it is possible that pain could be reduced over a three-month period of FO use, and with a reduction in pain could come an increase in physical activity levels and subsequent changes in muscle morphology. As such FO could indirectly affect muscle morphology in patients by changing internal rather than external load. Furthermore, in clinical practice FO are often prescribed in combination with exercises. The FO can facilitate a desirable position from which strength can be improved with exercises, after which FO may be removed. Future work with a focus on external validity could consider the response of muscle morphology and skin sensitivity to FO in patients when the FO are used in combination with typical exercises.

The methods used within this thesis are not without their limitations. It has been suggested that the presence of a fine-wire electrode in a muscle could affect gait patterns (Semple et al., 2009) and as such the findings of an effect of FO on muscle activity may not be externally valid. Although the presence of the fine-wire electrode in the TP muscle appeared to have an effect on the relative difference between conditions in peak PL EMG, the kinematics were comparable in the trials with the fine-wire electrode present and the trials after the electrode had been removed. There also appeared to be no difference in general TP EMG profiles or kinematics between participants who found the electrode particularly uncomfortable and those who were unaware of it. One might imagine if the electrode presence in the muscle affected gait, it would be most apparent in those who found it most uncomfortable, which was not the case. Additionally, the activation patterns of other shank muscles recorded using indwelling EMG have been shown to be similar to that recorded with surface EMG, indicating the activation pattern is not notably affected by the presence of an electrode in the muscle (Bogey et al., 2003, Bogey et al., 2000, Chimera et al., 2009, Onmanee, 2016). Tracking individual fascicle and MTU length change in the TP over stance in additional to indwelling EMG (Maharaj et al., 2016, Maharaj et al., 2017b, Maharaj et al., 2018) would give a more in depth picture of the effect of FO on TP function than EMG alone. However this is a highly specialist technique, requiring significant expertise not least because obtaining a clear image of individual fascicles to track using dynamic ultrasound with such a deep muscle as TP is very challenging and was not feasible in this thesis.

It is well established that motion capture with skin mounted markers involves skin movement/soft tissue artefact as the skin moves relative to the underlying bones (Milner, 2007). However it has been shown that the difference between bone pin data and skin mounted marker data is smaller than individual bone motion and the motion of a rigid segment, implying modelling the foot as a single segment is an oversimplification (Nester et al., 2007a, Nester, 2009). Multi-segment kinematics, beyond calculation of MLA, was beyond the aim of this thesis. Instead it was demonstrated that concurrent changes in kinematics, kinetics and TP EMG is possible with medial wedged FO, whilst acknowledging the limitations of non-invasive motion capture.

In the long term effects study the MDD for some CSA ultrasound measures was relatively high, particularly FDB (40.36 mm^2), which may have masked small changes in FDB CSA with FO if they were not detectable. Perhaps MDD of FDB CSA would have been smaller with MRI, however using repeat MRI measurements with over 50 participants would not have been feasible, due to MRI being far more expensive and less accessible than ultrasound. Skin touch sensitivity monofilament testing has demonstrated low reliability (ICC 0.46-0.61) on the back (Ellaway and Catley, 2013) and natural variability between testing sessions could have contributed to a lack of change in skin sensitivity with FO use. The method was designed to give a practical measure of global skin sensitivity of the regions of the foot, which would be similar to clinical practice and currently no better measure of skin touch sensitivity exists. Finally the Pedar in shoe plantar pressure measurements are only able to record pressure in the vertical direction and thus neglect shear forces in the medial/lateral direction. It is impossible to state whether there could have been differences in shear forces between the FO and flat inlay that would vary over time with FO use, however the plantar pressures in the long term effects study were not the major variables as muscle morphology and skin sensitivity were of primary interest.

8.7 Future research directions

Clear future directions would be to investigate the effect of FO on muscle activity and morphology in a pathological context and to explore the influence of sub groups. Reduced activity in the TP in early stance with FO with medial heel wedging could be beneficial for patients with TPTD, however it is possible that altered muscle function in these patients could affect how they respond to FO. Thus, studying the immediate effects of FO on TP EMG in TPTD is a worthwhile future research plan. Medial heel wedging significantly reduced TP activity while increasing arch support alone did not. A future randomised control trial could compare the clinical efficacy of medial heel wedging versus increased arch support in TPTD patients. Participants in the immediate effects of FO study in this thesis were not sub grouped based on foot posture and had mostly a neutral foot posture. Yet the reduction in peak TP EMG amplitude in early stance with medial wedging was similar to the reduction in TP activity with FO seen in a group of pes planus participants (Murley et al., 2010a). The long term effects study in this thesis also involved mostly those with a neutral foot so it was not possible to create large enough sub groups based on foot posture. Future work could investigate whether the response to FO over time differs with foot posture.

As a proof of concept, the immediate effects of FO in this thesis used extreme modifications in FO geometry, which were beyond what would likely be prescribed in clinical practice. A logical next step would be to establish the minimal magnitude of medial heel wedge to reduce peak TP EMG amplitude and to determine whether there is a linear dose response of medial wedge height on TP EMG. The modified prefabricated FO in the study by Murley et al. (2010a) which decreased peak TP amplitude by 19% had a 6 mm medial heel wedge, compared to the 8 mm wedge in this thesis. Whether reduced TP activity is possible with an even smaller amount of medial heel wedging has not been established. To investigate the possibility of a dose response a study design like that of Telfer et al. (2013a, 2013b) would be prudent, in which the magnitude of wedging was systematically increased from 6° lateral to 10° medial in 2° increments.

Studying the long-term effect of FO on skin sensitivity may in part be currently restricted by the current limitations of available measurement techniques. Participants reported “getting used to” the FO after around a week of use and that they then noticed the difference when wearing footwear without the FO. Anecdotally it would seem that some neurological adaptation to the redistribution of plantar pressure with FO occurs. For instance increased contact of the medial arch with the FO versus a shoe alone may increase stimulation of mechanoreceptors in that area and perhaps feedback from these receptors is down-weighted as the CNS adapts to the increased information. Participant perception of the FO was not captured in skin touch sensitivity thresholds. Future advances in neuromechanics may allow us to quantify how the sensory system adapts to mechanical changes in loading.

A universally accepted theory of the therapeutic effect of FO has yet to be established. None of the current clinical and biomechanical theories of the mechanisms behind the beneficial effect of FO appear to be well supported by empirical evidence and explain the effect of FO for the variety of conditions that FO can be used to treat. The reduction in external ankle eversion moment with medial wedges in this thesis supports the concept of reduced internal moment required from internal structures, reducing the force level through soft tissue. However the “tissue stress theory” relates specifically to the moment about the subtalar joint axis and given the complex motion of the foot at multiple joints (Lundgren et al., 2008), considering a reduction in torque about the STJ alone may be too simplistic. The external moment in this thesis was recorded about the ankle joint, while recording the moment about the STJ itself is possible (Maharaj et al., 2016, Maharaj et al., 2017b, Maharaj et al., 2017a, Maharaj et al., 2018), it involves advanced modelling procedures. The reduced peak TP EMG amplitude with medial heel wedging supports the idea of a neuromuscular component to the mechanisms of FO together with reduced loading, however reduced amplitude alone does not fit into the proposed defined paradigms like the preferred movement pathway or muscle tuning, which relate to maintaining an optimum kinematic pattern and vibration in soft tissue respectively. Reduced TP EMG in early stance has a logical benefit to TPTD, but does not explain how FO can be effective for other conditions. In the case of plantar fasciitis for example, one mechanism of FO effect could be simply the redistribution of plantar pressure and the reduction in plantar pressure at the heel which is often the site of pain. It is possible that different mechanisms of FO are responsible for their clinical efficacy in different conditions and could vary with FO geometry and even between individuals, given the large variability in response to FO (Donoghue et al., 2008, Mills et al., 2009). Future FO should continue to see the effect of FO as multifactorial and could work towards more unified theories than exist at present.

8.8 Implications for practice

The finding of reduced peak TP EMG amplitude in early stance with FO with medial heel wedging could influence the choice of FO design in treating TPTD. If the goal of the treatment is to reduce the force through the TP muscle tendon unit in order to give the tendon a chance to heal, then clinicians could consider prescribing a medial heel wedge as this may be more effective than increased arch support. The reduction in TP EMG amplitude with medial heel wedging of 14-17% was accompanied by reductions in peak maximum eversion moment of 30-38%. Although such changes are substantial, a threshold for a clinically meaningful effect has not been established (Telfer et al., 2013b). Nonetheless FO medial heel wedging are readily available for the clinician. Pre-fabricated FO made of EVA or thermoplastic, as used in this thesis, are common in clinical practice (Nester et al., 2017). The medial heel wedging used in this study were incorporated into the FO construction using a CAD/CAM system. However, wedge additions can be easily stuck onto a pre-fabricated standard FO, thus are accessible to clinicians. Although a lack of change in the variables measured in the long term effect of FO study in this thesis does not provide any explanation of the mechanism of FO effect over time, it may help in the clinical management of patients as the lack of change in muscle morphology provides novel indirect evidence to counter the belief held by some that FO make feet weaker, a belief that could be a barrier for the willingness of some patients to wear FO.

8.9 Conclusion

This PhD has demonstrated a plausible link between foot orthosis design, external force and activity of the tibialis posterior, which could direct treatment practice. A reliable protocol for recording EMG from the peroneus longus was developed along with a thorough method for the challenging technique of recording from the tibialis posterior. Reduction in tibialis posterior EMG in early stance was found with medial wedging, which is valuable information in prescribing foot orthoses for the treatment of tibialis posterior tendon dysfunction. Additionally a crucial limitation in indwelling EMG was identified, in that the amplitude recorded from fine-wire electrodes can reduce over time irrespective of any intervention being tested, which is a vital consideration for future protocols. Longer term effects of foot orthosis use on soft tissues and skin sensitivity were not found. However there was likewise no evidence that foot orthoses make intrinsic foot muscles smaller and as muscle size is related to muscle strength, no reason to believe foot orthoses weaken muscles and subsequently no reason to discourage their use in the clinic.

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Appendix

Ethics approval

Reliability study: HSCR 16-92

Immediate effects study: HSR1617-36

Long-term effects study: HSR1718-009 (University of Salford)

REB NUMBER: 17-08-019 (University of Guelph)

Within session reliability was assessed by visual inspection of plots of individual gait cycles of shoe and barefoot data in the reliability study in Chapter 5

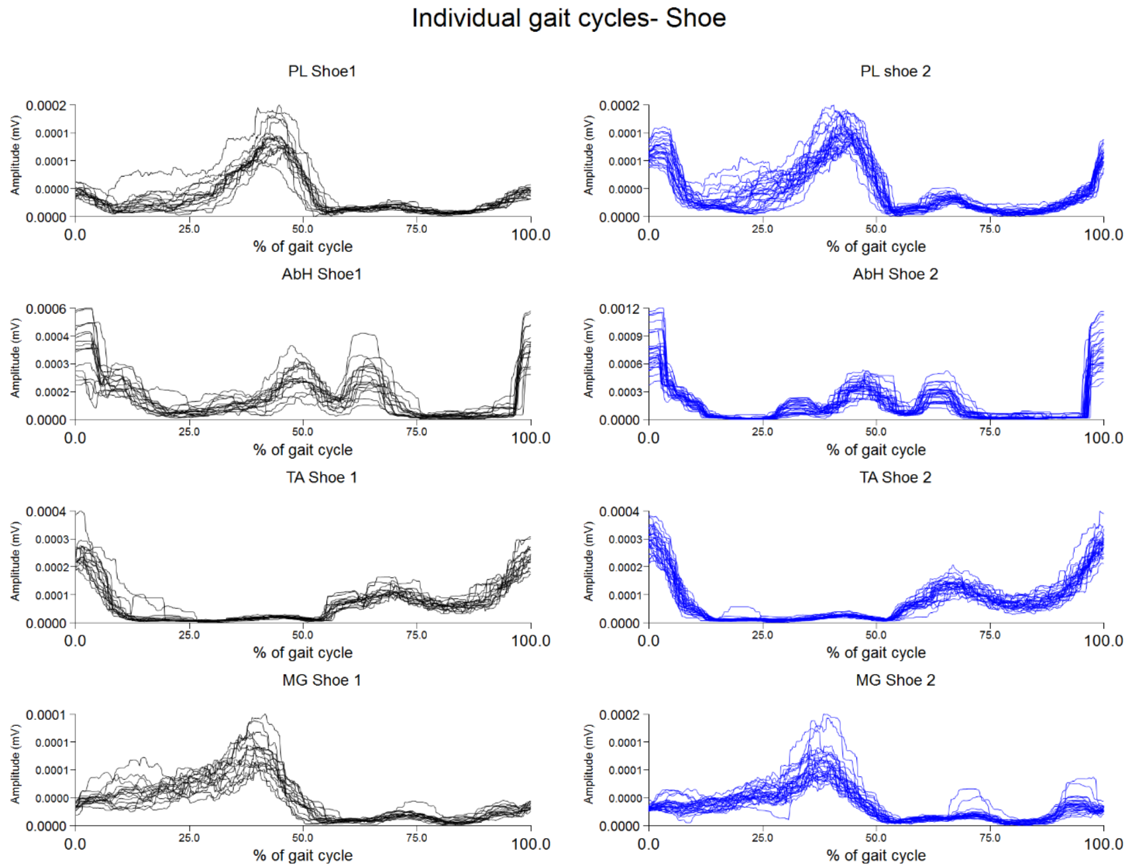

Figure 0-1. Example plots of individual gait cycles in the shoe condition from one participant in session one (black lines) and session two (blue lines) for peroneus longus (PL), abductor hallucis (AbH), tibialis anterior (TA) and medial gastrocnemius (MG)

Problem EMG data and TP over an hour

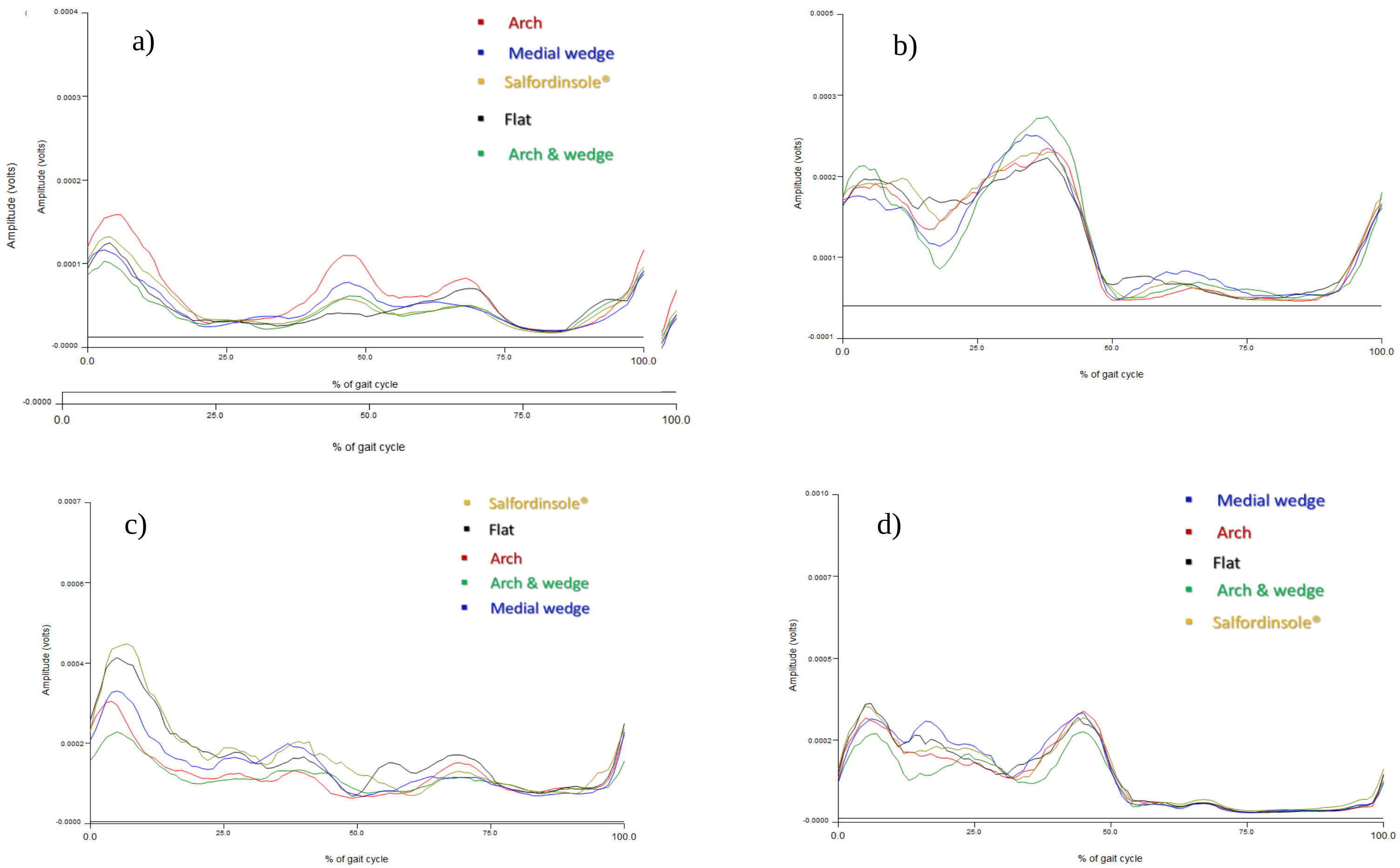

Figure 0-2 Tibialis posterior EMG for participants a) AL131017; b) CF150217; c) CP121017; d) CS061017. Black lines: flat inlay; yellow lines: Salfordinsole®; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging; condition order in legend

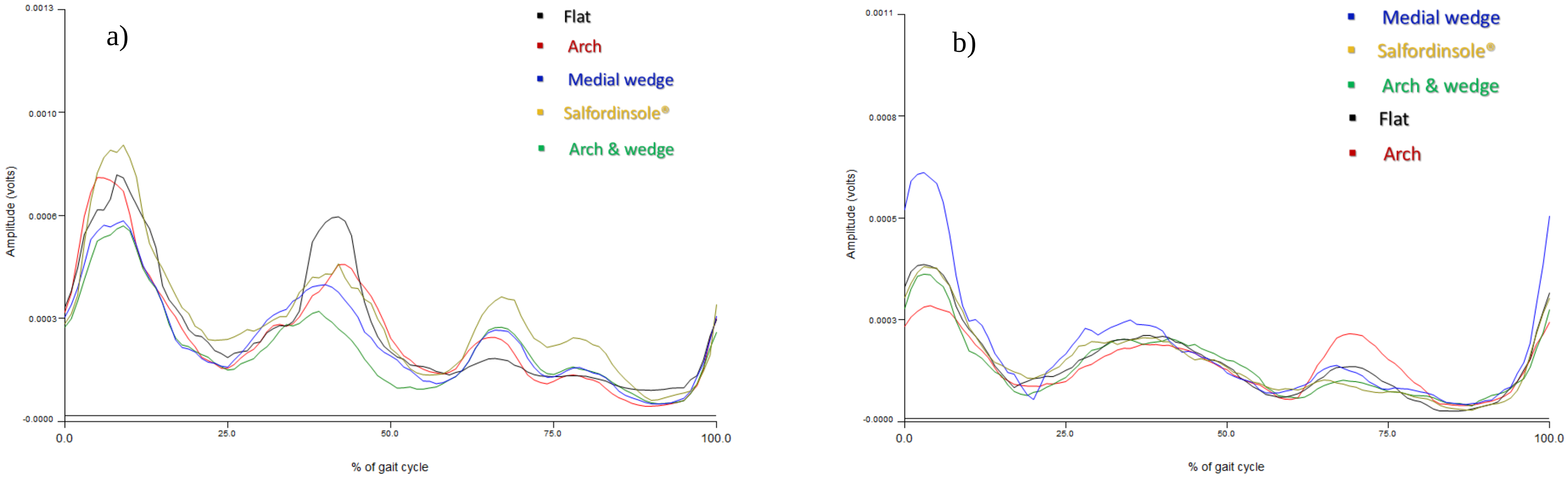

Figure 0-3 Tibialis posterior EMG for participants a) JA160217; b) SV020217. Black lines: flat inlay; yellow lines: Salfordinsole®; red lines: 6 mm increase in arch height; blue lines: 8 mm increase in medial wedging; green lines: 6 mm increase in arch height combined with 8 mm increase in medial wedging; condition order in legend

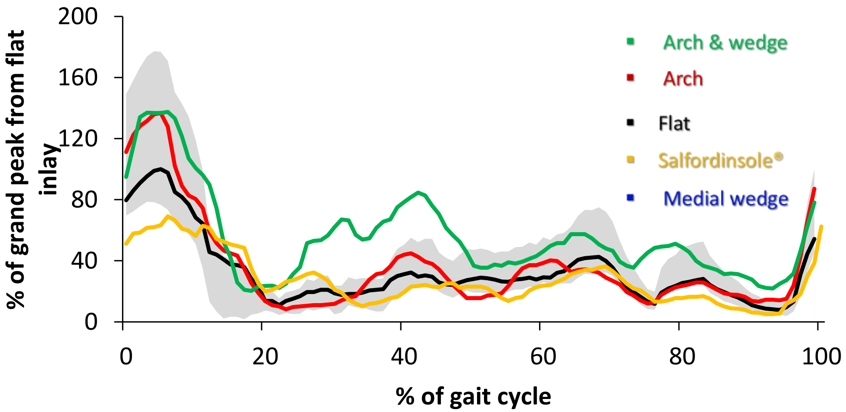

Figure 0-4. Tibialis posterior EMG from excluded participant HS181017 after first protocol modification. Medial wedge data discarded

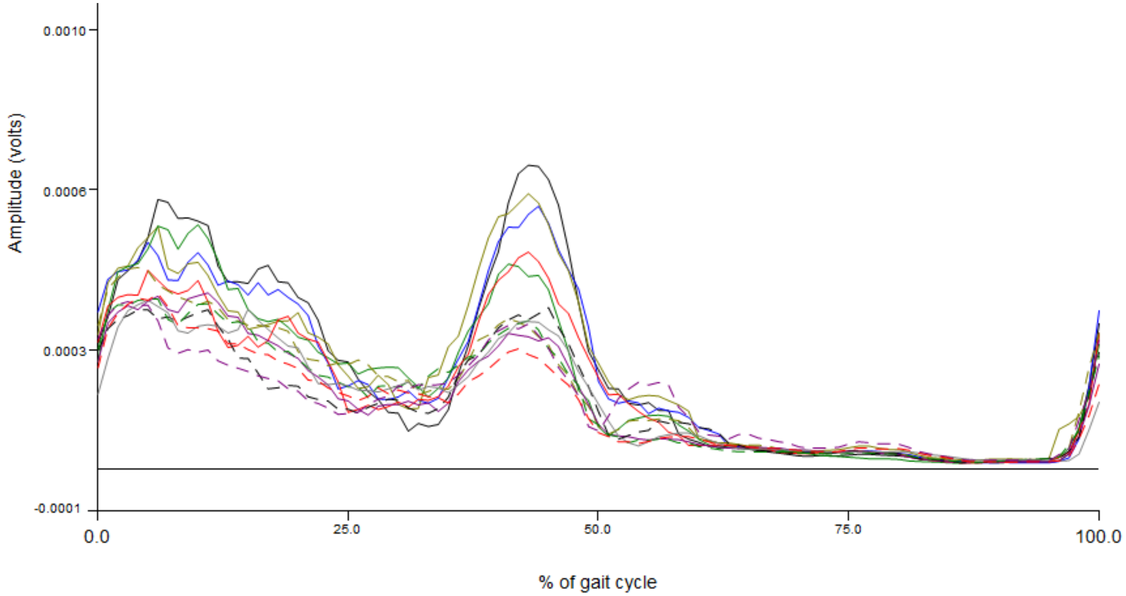

Figure 0-5. Tibialis posterior EMG over an hour in a single shod condition. Legend in Table 1. Each recording consists of six trials with multiple gait cycles (around 15 per trial).

Table 0-1. Trial legend for tibialis posterior EMG over an hour

Recording no.	Time	Colour/type
1	15:11	Black solid
2	15:16	Blue solid
3	15:20	Yellow solid
4	15:25	Green solid
5	15:30	Blue dash-dot
6	15:35	Red solid
7	15:40	Yellow dash
8	15:45	Purple solid
9	15:50	Black dash
10	15:55	Green dash
11	16:00	Purple dash
12	16:05	Grey solid
13	16:10	Red dash

Tibialis anterior EMG over time with fine-wire and surface electrodes

Below is the initial results of an additional project comparing the performance of surface and fine-wire electrodes over time submitted as a conference publication to the International Society of Posture and Gait Research Congress 2019 and the Ontario Biomechanics Conference 2019 (below).

Reeves J, Starbuck C, Rafiq W, Bent L, Nester C. Surface and fine-wire electrode performance over time when recording from the tibialis anterior in walking. (2019). 16th Annual Ontario Biomechanics Conference; Nottawasaga Inn, Alliston Ontario March 8-10.

SURFACE AND FINE-WIRE ELECTRODE PERFORMANCE OVER TIME WHEN RECORDING FROM THE TIBIALIS ANTERIOR IN WALKING

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Introduction: The consistency of indwelling electromyography (EMG) over time using fine-wire electrodes is not documented. Activation patterns of shank muscles recorded using indwelling EMG is like that recorded with surface EMG [1]. Yet, indwelling EMG may be less consistent over time vs. surface EMG, perhaps due to changes in electrode capacity with exposure to the internal environment. Our experience using indwelling EMG suggests that there is a reduction in indwelling EMG amplitude after 20-40 mins of walking using certain fine-wire electrodes. The aim of this study was to characterise the EMG signal over time when recording from the tibialis anterior (TA) during walking using fine-wire and surface electrodes.

Methods: Five healthy males performed ten walking trials of 5 mins with 1 min breaks. Fine-wire electrodes (Chalgren Enterprises Inc., CA, USA, 50 mm long, 25 gauge) were inserted into the right TA and connected to a spring contact sensor (Delsys, Inc., MA, USA). A Delsys Trigno™ Mini sensor was attached near the insertion site and an IMU sensor was fixed to the distal shin to determine foot contact from acceleration. EMG was collected at 2000 Hz and bandpass filtered at 10-500 Hz. Windows of 75 ms (surface) and 250 ms (fine-wire) were used to calculate the route mean squared envelope per gait cycle and averaged per trial. Amplitude was normalized to the mean of the peak per gait cycle in trial 1 and time normalized to the gait cycle.

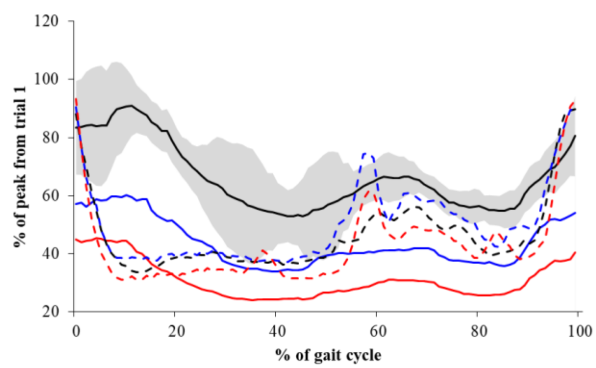

Figure 1. Mean tibialis anterior EMG over the gait cycle recorded with fine-wire electrodes (solid lines) and surface electrodes (dashed lines). Black line= trial 1; blue line= trial 5; red line= trial 10. Shaded grey area= standard deviation from trial 1

Results: Figure 1. shows how the EMG profile of the TA across the gait cycle decreased in amplitude with progressive walking trials when recorded with indwelling EMG. By trial 5 (up to 30 mins) amplitude reduced by ~40% and by trial 10 (up to 50 mins), amplitude reduced by ~50%. In contrast, the EMG of TA recorded with surface EMG remained relatively consistent.

Discussion and Conclusions: Reduced signal amplitude over time with indwelling EMG and not surface EMG suggests that the recording capacity of these fine-wire electrodes diminishes with continuous use, rather than altered activation of the muscle itself. Any change in signal properties over time is a vital consideration for any repeated measures study design using indwelling EMG, as a change in the recording capacity of the electrode would confound the effect of the independent variable being tested.

Reference: [1] Chimera, N. et al. (2009). J ELECTROMYOGR KINES 19(6), e494-e499.

Conference paper on the relationship between toe grip strength and muscle size

Below is a conference abstract to be presented at the Footwear Biomechanics Symposium 2019

THE RELATIONSHIP BETWEEN TOE GRIP STRENGTH AND INTRINSIC MUSCLE MORPHOLOGY

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Introduction

Intrinsic foot muscle strength is thought to be important in foot function and may contribute to foot posture (Angin et al., 2018). Muscle morphology (cross sectional area (CSA) and thickness) is often used as a surrogate measure of muscle strength, as measuring toe strength directly is difficult (Mickle et al., 2013). Uritani et al. (2014) developed a toe grip dynamometer as a quick and portable means of assessing toe grip strength directly. It is unknown whether toe grip strength measured with this device relates to muscle morphology.

Purpose

The aim of this study was to explore the relationship between intrinsic foot muscle size and toe grip strength measured using a toe grip dynamometer.

Methods

Twenty-nine healthy participants (Males= 9, mean ±SD age: 30 ±10 years; height: 1.65 ±0.07 m; mass: 67.3 ±13.2 kg; Foot Posture Index: 3 ±4; shoe size 7 ±2) were included in this analysis.

Toe grip strength of the right foot was tested using a T.K.K.3362 toe-grip dynamometer (**Figure 0-6**, Takei Scientific Instruments, Niigata, Japan) as used previously (Uritani et al., 2014). Participants placed as many toes as comfortable on the bar and practiced pulling back on the bar maximally. Participants then did 3 maximum efforts and the highest value (kg) was recorded.

Figure 0-6. Toe grip dynamometer from Uritani et al., 2014

Ultrasound images of the right foot were recorded to obtain CSA and thickness of the abductor hallucis (AbH), flexor hallucis brevis (FHB) and the flexor digitorum brevis (FDB) at their thickest points using a MyLab 70 Xvision (13 MHz linear array transducer, Type, LA523, Esoate Europe, UK). Two images were taken per structure. For thickness, a mean of three measurements were taken per image and for CSA a mean of two measurements per image. Images were analysed in ImageJ (NIH, USA). Correlations were calculated to explore variables related to toe grip strength and each other. A backward stepwise regression analysis was then performed using input variables that were significantly correlated to toe grip strength ($p<0.05$).

Results
Mean ±SD toe grip strength was: 12.5 ±4.4 kg and 15.5 ±5.4 kg for females and males respectively. Height and shoe size were significantly correlated with toe grip strength ($p<0.001$, $R=0.57$ and $p=0.003$, $R=0.48$ respectively), however shoe size and height were significantly correlated ($p<0.001$, $R=0.74$), as such shoe size did not contribute to further variance. Of the ultrasound variables FHB thickness (**Figure 0-7**) was significantly correlated with toe grip strength ($p=0.02$, $R= 0.54$). The regression analysis found height ($p= 0.02$) and FHB thickness ($p= 0.04$) were predictors of toe grip strength ($R^2= 0.52$).

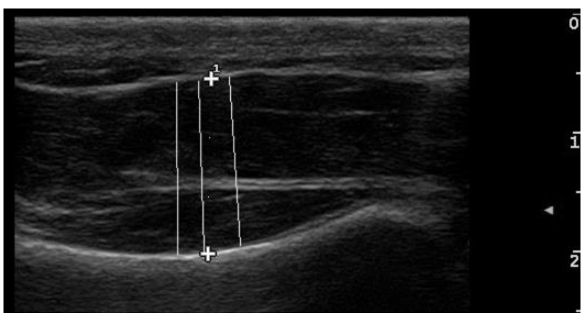

Figure 0-7. Ultrasound image of the flexor hallucis brevis

Discussion
Height and FHB thickness were predictors of toe grip strength, explaining 52% of the variance. The design of the dynamometer likely explains why FHB was related to toe grip strength, while FDB variables were not. The grip bar of the dynamometer is perpendicular to the long axis of the foot and it is easier to grip the bar with the hallux than the lesser toes, thus FHB likely contributes more to the output of the dynamometer than FDB. Extrinsic foot muscles, like flexor hallucis longus, probably also contribute to toe grip strength output.

Unlike the previous study sex, age and weight were not predictors of toe grip strength in this study, despite comparable mean values (Uritani et al., 2014). However, as a secondary analysis of a larger study, our study had a smaller sample size and age range than the previous work, which may explain a lack of a relationship for these other variables with toe grip strength. Similarly our sample had a mostly normal foot posture, which may explain why there was no relationship between FPI and toe grip strength, despite a proposed relationship between soft tissue morphology, as a surrogate of foot/toe function, and foot posture (Angin et al., 2018).

This study has demonstrated a relationship between FHB thickness and toe grip strength using a toe grip dynamometer. This could inform the decision process of measures to include in future research on toe/foot function, e.g. in using FHB thickness as a surrogate measure of strength in studying toe deformities.

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