Effect of foot posture and foot orthoses on the electromyographic activity of selected lower limb muscles during walking

George Samuel Paul Murley

Bachelor of Podiatry (Hons)

A thesis submitted in total fulfilment of the requirements for the degree of Doctor of Philosophy

Department of Podiatry

Faculty of Health Sciences

La Trobe University

Bundoora, Victoria 3086

Australia

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Summary

This thesis presents a series of five related research projects investigating the electromyographic (EMG) activity of selected lower limb muscles during walking. Initially, a systematic literature review identified key deficiencies with current evidence for the relationship between foot posture, foot orthoses and footwear on lower limb muscle activity during walking. The findings from this review indicated that a rigorous approach of classifying foot posture was required to investigate the effect of foot posture and foot orthoses on lower limb muscle function during walking.

Subsequently, a foot posture screening protocol was designed to recruit participants with normal- and flat-arched feet. The foot posture screening protocol was then used to recruit participants to a series of three laboratory-based EMG studies. The first EMG study explored the between-session reliability of tibialis posterior, peroneus longus, tibialis anterior and medial gastrocnemius EMG during walking. The findings indicated that although EMG from these muscles is relatively stable within a testing session, the between-session reliability is generally poor. Consequently, the last two EMG studies were conducted with a single-session design.

Of these two studies, the first compared muscle activity from participants with normal-arched feet to participants with flat-arched feet, and identified significant differences in tibialis posterior, peroneus longus and tibialis anterior EMG amplitude during walking. The second EMG study investigated whether two types

of foot orthosis, a prefabricated and a customised device, changed muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet. Both foot orthoses significantly altered tibialis posterior and peroneus longus EMG amplitude, however only the prefabricated orthosis changed peroneus longus EMG amplitude towards a pattern observed with normal-arched feet. Further research is needed to determine if these changes are beneficial to patients with foot pain and disability.

Statement of authorship

I, George Murley, declare that except where reference is made in the text of the thesis, this thesis contains no material published elsewhere or extracted in whole or in part from a thesis submitted for the award of any other degree or diploma.

I, George Murley, declare that no other person's work has been used without due acknowledgment in the main text of the thesis. This thesis has not been submitted for the award of any degree or diploma in any other tertiary institution. With regard to the extent of collaboration with another person or persons, although the publications involve joint authorship, I have made a significant and leading contribution to the work, equivalent to that expected for a traditional thesis – I am first author on all five publications presented in the thesis.

All research procedures reported in the thesis were approved by the relevant Ethics Committee or Safety Committee or authorised officer (Ethics ID: FHEC06/205)

Signed…………………………………………………………Date………………..

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Aim of the thesis

The primary aim of this thesis was to determine the relationship between foot posture, foot orthoses and lower limb electromyographic activity during walking.

Objectives of the thesis

The following objectives were adopted to systematically address the primary aim of the thesis:

(i) Conduct a systematic literature review and identify key deficiencies with current evidence related to the relationship between foot posture, foot orthoses and lower limb electromyographic activity during walking

(ii) Design a protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements

(iii) Determine the relative and absolute reliability of investigating electromyography from selected leg muscles during walking

(iv) Identify the most stable electromyographic parameters including normalisation techniques that reduce variability

(v) Explore differences in lower limb electromyographic activity between normal and flat-arched foot posture during walking

(vi) Establish the short-term effects of customised and prefabricated foot orthoses on lower limb electromyographic activity during walking

(vii) Determine whether foot orthoses change muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet

Outline of thesis

To achieve the objectives of the thesis, a series of five related studies were undertaken. These studies plus complementary information are presented in eight chapters.

Chapter 1 outlines the background to the thesis, provides an overview of the literature and states the aims and objectives of the thesis.

Chapter 2 (Study 1) presents a systematic review of literature related to the relationship between foot posture, foot orthoses and lower limb electromyographic activity during walking. The purpose of this review was to inform researchers planning future studies of relating to foot posture and interventions (e.g. foot orthoses). Chapters 3–7 were designed to advance deficiencies identified in the systematic literature review.

Chapter 3 (Study 2) presents a protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements. This protocol was used to screen potential participants and to identify those with normal-and flat-arched foot posture for inclusion in studies 3–5 (presented in Chapters 5, 6 and 7 respectively).

Chapter 4 provides an overview of the electromyographic methodology employed for studies 3–5 (presented in Chapters 5, 6 and 7 respectively).

Chapter 5 (Study 3) presents a reliability study for the assessment of lower limb electromyography during walking. This study explores variability issues including within- and between-session error and the effect of different normalisation techniques on between-participant variability. Findings from this study influenced the design of studies 4 and 5 (presented in Chapters 6 and 7 respectively).

Chapter 6 (Study 4) presents a study that compared lower limb electromyography from young adults with normal- and flat-arched foot posture during walking.

Chapter 7 (Study 5) presents the final study that compared the effect of customised and prefabricated foot orthoses during walking in young adults with flat-arched foot posture.

Chapter 8 provides a discussion of the main findings including limitations and concludes with recommendations for a future research agenda.

The following table presents an overview of the research undertaken in this thesis to show how the project evolved from the systematic literature review and how participants were involved in specific studies. In addition, a brief summary of the main findings from each study is presented.

Structure of the research undertaken and summary of findings

Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review **(Chapter 2)**

Key findings: Some evidence exists that: (i) pronated feet demonstrate greater activation of invertor musculature and decreased activation of evertor musculature; (ii) foot orthoses increase activation of tibialis anterior and peroneus longus, and may alter lower back muscle activation. Studies were of only moderate methodological quality with significant deficiencies in basic reporting of effect size and error.

To address some of these issues, the sequence of studies below were undertaken

Participant recruitment

The foot posture of ninety-one asymptomatic young adults was assessed using two clinical measurements (normalised navicular height and arch index) and four radiological measurements taken from anterior-posterior and lateral x-rays (talus-second metatarsal angle, talo-navicular coverage angle, calcaneal inclination angle and calcaneal-first metatarsal angle). Normative foot posture values were taken from the literature and used to recruit participants with normal-arched feet. Data from these participants were subsequently used to define the boundary between normal- and flat-arched feet. This information was then used to recruit participants with flat-arched feet.

A protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements **(Chapter 3)**

Key findings: The values obtained from the two clinical and four radiological measurements established two clearly defined foot posture groups. Correlations among clinical and radiological measures were significant ($p<0.05$) and ranged from $r=0.24$ to 0.70. Interestingly, the clinical measures were more strongly associated with the radiographic angles obtained from the lateral view. This foot screening protocol provides a coherent strategy for researchers planning to recruit participants with normal- and flat-arched feet.

From the 91 participants screened, 30 qualified for the normal-arched foot posture study and attended two EMG sessions approximately two weeks apart.

Reliability of lower limb electromyography during overground walking: a comparison of maximal- and sub-maximal normalisation techniques **(Chapter 5)**

Key findings: Time of peak EMG amplitude for all muscles displayed relatively narrow limits of random error. However, reliability of peak and root mean square amplitude parameters for tibialis posterior and peroneus longus displayed unacceptably wide limits of random error, regardless of the normalisation reference technique. Moderate limits of random error for tibialis anterior and medial gastrocnemius amplitude were obtained with no normalisation and with sub-maximal reference contractions, compared to traditional maximum isometric contractions. Timing and amplitude EMG parameters for all muscles displayed low to moderate coefficient of variation within each test session (range: 7-25%). Overall, between-participant variability was minimised with sub-maximal normalisation values. These results demonstrate that re-application of electrodes results in large random error between sessions, particularly with tibialis posterior and peroneus longus. Researchers planning studies of these muscles with a repeated-test design (e.g. to evaluate the effect of an intervention) must consider whether this level of error is acceptable.

Foot posture influences the electromyographic activity of selected lower limb muscles during gait **(Chapter 6)**

Key findings: Statistically significant differences in EMG amplitude were detected for tibialis posterior, peroneus longus and tibialis anterior. Differences in muscle activity in people with flat-arched feet may reflect neuromuscular compensation to reduce overload of the medial longitudinal arch.

From the 91 participants screened, 30 qualified for the flat-arched foot posture study and were issued with prefabricated and customised foot orthoses.

Effect of prefabricated and customised foot orthoses on the electromyographic activity of selected lower limb muscles during gait **(Chapter 7)**

Key findings: The foot orthoses significantly altered tibialis posterior and peroneus longus EMG amplitude. However, only the prefabricated orthosis changed peroneus longus EMG amplitude towards a pattern observed with normal-arched feet. Otherwise, few differences were found between the prefabricated and customised orthoses.

Publications by the candidate relevant to the thesis

The eight chapters presented in this thesis include five publications.

Chapter 2 has been published as:

Murley GS, Landorf KB, Menz HB, Bird AR. Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review. *Gait and Posture* 2009, 29(2):172-87

Chapter 3 has been published as:

Murley GS, Menz HB, Landorf KB. A protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements. *Journal of Foot and Ankle Research* 2009, 2(22):1-13.

Chapter 5 has been published as:

Murley GS, Menz HB, Landorf KB, Bird AR. Reliability of lower limb electromyography during overground walking: a comparison of maximal- and sub-maximal normalisation techniques. *Journal of Biomechanics* 2010, 43:749-56.

Chapter 6 has been published as:

Murley GS, Menz HB, Landorf KB. Foot posture influences the electromyographic activity of selected lower limb muscles during gait. *Journal of Foot and Ankle Research* 2009, 2(35):1-9.

Chapter 7 has been published as:

Murley GS, Landorf KB, Menz HB. Do foot orthoses change lower limb muscle activity in flat-arched feet towards a pattern observed in normal-arched feet? *Clinical Biomechanics* 2010; 25(7):728-36.

Additional publications by the candidate during candidature

Murley GS, Bird AR. The effect of three levels of foot orthotic wedging on the surface electromyographic activity of selected lower limb muscles during gait. *Clinical Biomechanics* 2006, 21(10):1074–1080.

Murley GS. Letter to Editor. Re: Anomalous tibialis posterior muscle, functional or functionless? *The Foot* 2007, 18(2):119-120.

Munteanu SE, Strawhorn AB, Landorf KB, Bird AR, Murley GS. A weightbearing technique for the measurement of ankle joint dorsiflexion with the knee extended is reliable. *Journal of Science and Medicine in Sport* 2009, 12(1):54-59.

Murley GS, Buldt AK, Trump PJ, Wickham JB. Tibialis posterior EMG activity during barefoot walking in people with neutral foot posture. *Journal of Electromyography and Kinesiology* 2009, 19(2):e69-e77.

Semple R, Murley GS, Woodburn J, Turner DE. Tibialis posterior in health and disease: a review of structure and function with specific reference to electromyographic studies. *Journal of Foot and Ankle Research* 2009*,* 2(24):1-8.

Levinger, P, Murley GS, Barton CJ, Cotchett M, McSweeney SR, Hylton HB. A comparison of foot kinematics in people with normal- and flat-arched feet using the Oxford Foot Model. *Gait and Posture* 2010, 32(4):519-23.

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Ethical approval

Ethical approval for the research undertaken in the thesis was obtained from a single application to the La Trobe University Human Ethics Committee (Ethics ID: FHEC06/205) (Appendix 1) and it was registered with the Radiation Safety Committee of the Victorian Department of Human Services (Appendix 2).

Participants involved provided informed written consent (Appendix 3) after reviewing the participant information form prior to their involvement (Appendix 4).

The x-rays performed as part of the investigations presented in Chapters 3, 5, 6 and 7 were conducted in accordance with the Australian Radiation Protection and Nuclear Safety Agency Code of Practice for the Exposure of Humans to Ionizing Radiation for Research Purposes (2005).[1]

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[1] Australian Radiation Protection and Nuclear Safety Agency - Exposure of Humans to Ionizing Radiation for Research Purposes. [cited 29 May 2009]; Radiation Protection Series No. 8:[Available from: www.arpansa.gov.au/pubs/rps/rps8.pdf].

1.0 Introduction and literature review

1.1 Foot posture

Human foot posture is characterised by the alignment of the bones of the foot relative to each other, as well as how they relate to the lower leg. Variations from normal foot posture are thought to influence foot and lower limb function during gait, predisposing individuals to injury [1]. However, for researchers and clinicians in the field there is no consensus for which parts of the foot skeleton define foot posture [1]. Several methods for assessing and categorising foot posture have been developed such as; visual observation, footprint parameters, arch height and radiographic measurements [1]. When considering this array of measurements for categorising foot posture, it is evident that no section of the foot seems more compelling in research and clinical practice than the structure of the medial longitudinal arch (MLA).[2]

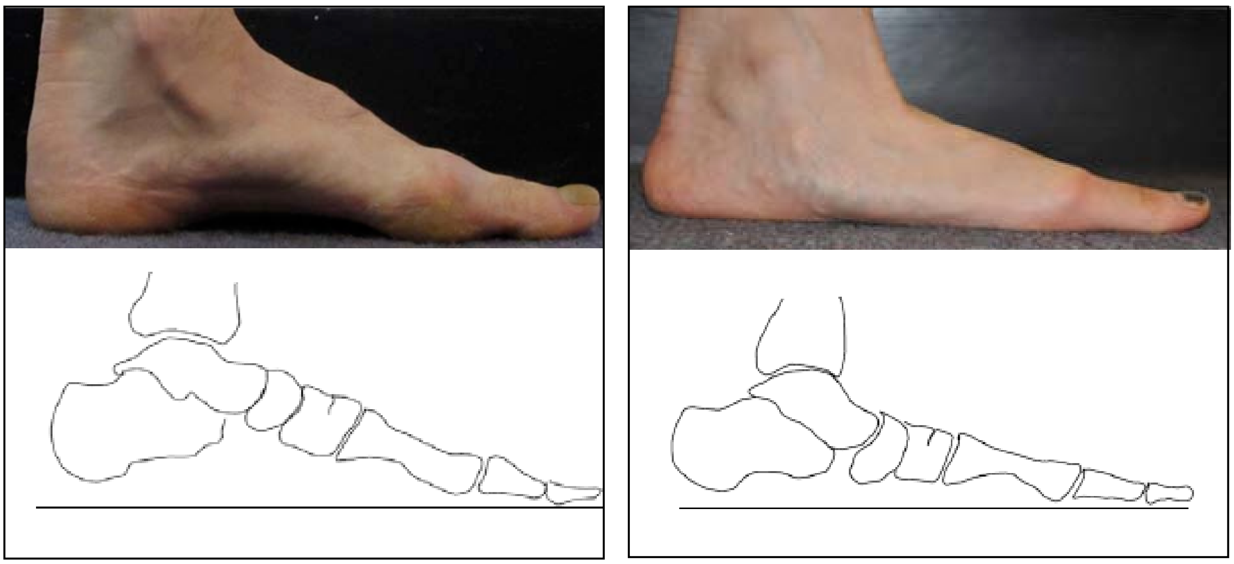
Irrespective of the techniques used to categorise foot posture, there is evidence that the MLA varies considerably between individuals [2] and is influenced by a range of factors including ethnicity [3,4], shoe-wearing habits [5,6] and systemic conditions, such as neurogenic [7] and rheumatological disease [8]. Undoubtedly, foot posture is also determined by complex interactions between the bones of the foot and the forces applied to them by passive and active tissues and external forces.

[2] The terms foot posture and MLA are used interchangeably in the literature review to describe the structure of the medial longitudinal arch.

In adults who have developed normally from a physical standpoint, foot posture is most commonly ‘slightly pronated’ [2]. However, foot posture changes over an individual’s life. For example, a study by Redmond and colleagues [2] reported that at either end of one’s life – during early childhood and old age – foot posture is on average pronated, compared to the general adult population. It is unclear, however, whether normal and pronated feet are distinct entities or simply two points along the same continuum of foot posture. Figure 1.1 shows examples of normal- and flat-arched feet illustrated from pictures and line drawings of x-rays.

Figure 1.1. Photographs and line drawings of x-rays (medial view) taken from two people: (a) normal-arched foot; and (b) flat-arched foot.

(a) normal-arched foot (b) flat-arched foot



Since there is a lack of consensus on how to best assess and classify foot posture, it is not surprising that the terminology used to describe common variants in foot posture is diverse. For example, the terms ‘pronated’, ‘excessively pronated’, ‘hyperpronated’, ‘valgus’, ‘everted’, ‘pes planus’, ‘pes plano-valgus’, ‘flat-arch’ and ‘flat-foot’ have all been used frequently to describe the postural appearance of the foot shown in Figure 1.1 (b). For this thesis, the terms ‘normal-arch’ and ‘flat-arch’ will be used – these terms reflect that the research is focused on the posture of the MLA.

As mentioned earlier, some variations in foot posture, such as flat- or high-arched foot type, are thought to cause tissue stress that results in injury [9]. It is also believed that musculoskeletal injuries related to altered foot posture can become debilitating and reduces the capacity of individuals to exercise and participate in activities of daily living. Several lower limb injuries associated with abnormal foot posture are commonly treated with foot orthoses. Therefore, to assist our understanding of both the causes of overuse conditions of the lower limb and the effects of various interventions to treat these conditions (i.e. foot orthoses), it is important to investigate whether foot posture and foot orthoses influence lower limb movement and injury.

Accordingly, the following two sections (1.2 and 1.3) present an overview of research that has investigated; (i) the effect of *foot posture* on the lower limb biomechanics and injury, and (ii) the effect of *foot orthoses* on lower limb biomechanics and injury. Where possible, systematic reviews of relevant literature are presented at the beginning of each section to reflect their role in providing the highest level of evidence [10,11].

Chapter 1 is followed by the first published study in this thesis titled: *Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: A systematic review* (Chapter 2).

1.2 Effect of foot posture on lower limb biomechanics and injury

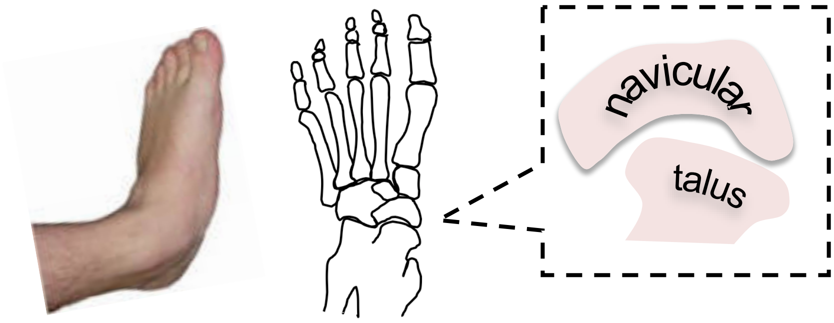
1.2.1 Foot posture and lower limb biomechanics

It has been suggested that pronated foot posture leads to excessive motion and reduced joint congruency [9]. This may place greater stress on soft tissue structures, such as ligaments, tendons and muscles, to maintain foot stability during gait [9]. For example, forefoot varus, characterised by inversion of the forefoot relative to the hindfoot, is thought to induce pronation of the subtalar and midtarsal joints to allow the first metatarsal head to become plantigrade during stance [12]. These events are expected to increase internal rotation of the tibia and femur and place greater stress on tissues that oppose these movements, such as tibialis posterior (Figure 1.2).

With this in mind, this section summarises the literature that has investigated the effect of foot posture on lower limb biomechanics, specifically related to kinematics and plantar pressure measurements during walking. For clarity, only key studies involving healthy participants during walking are included.

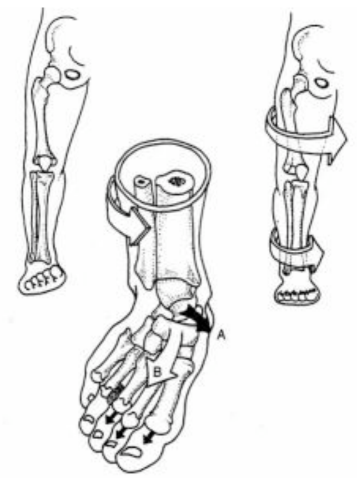
Figure 1.2. Potential relationship between flat-arched foot posture, lower limb motion, muscle activity and injury.

Flat-arched foot



1. Reduced joint congruency with flat-arched foot posture. For example, the talus can be adducted relative to the articulation with the navicular.

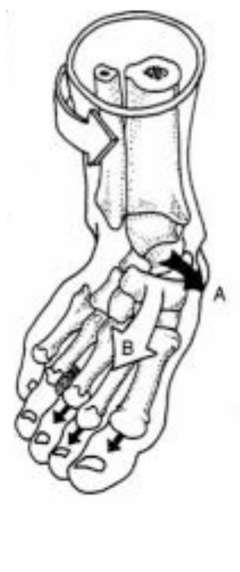
Compensatory motion in lower extremity



Picture taken from Michaud (1998) Foot orthoses and other forms of conservative foot care, Massachusetts.

2. Reduced joint congruency may lead to hyper mobility or excessive motion of the foot and lower limb. For example, flat-arched foot posture influences movement and position of different lower limb segments during walking.

Additional load is placed on related tissues



Picture taken from Michaud (1998) Foot orthoses and other forms of conservative foot care, Massachusetts. Michaud proposed that as the subtalar joint pronates the talus is forced to adduct and plantarflex (A). With these movements, the calcaneonavicular ligament and the plantar talonavicular joint capsule are placed under greater load. (B) Anterior displacement of the talus causes the navicular and first three rays to move forward and abduct relative to the fourth and fifth rays – this may increase tensile force through the plantar aponeurosis and cause injury.

3. Reduced joint congruency and excessive motion may place greater stress on soft tissue structures, such as ligaments, tendons and muscles, to maintain foot stability during gait.

4. Additional muscular support (i.e. from tibialis posterior) during gait is required to stabilise the joint and reduce excessive tissue stress.

5. Fatigue and damage of these controlling muscles may result in the development of various injuries.

Kinematic studies

The literature that investigated lower limb kinematics has predominantly studied the effect of foot posture on position and movement of the forefoot, the arch, the rearfoot and the tibia during walking. The following section presents an overview of four key studies [12-15] that investigated people with features of either normal- or flat-arched foot posture using multi-segment three-dimensional video motion analysis.

Two studies have investigated the effect of forefoot varus on lower limb kinematics during walking [12,14] (Table 1.1). One of these studies modelled the foot as three rigid segments and compared 10 children with forefoot varus to 11 children with normally-aligned feet [12]. This study reported findings from 56 kinematic variables and found that forefoot varus was associated with significantly less hip adduction and extension during the loading response and midstance, respectively. The second study [14] modelled the foot as two rigid segments and compared seven subjects with normal foot posture to 14 subjects with abnormally pronated foot posture. Participants' foot posture was classified using several goniometric measurements and the navicular drop difference comparing subtalar joint neutral with the relaxed position. The results indicated that subjects with pronated feet were significantly more inverted than the normal group at initial contact, however during midstance phase the rearfoot of the pronated group was significantly more everted.

Two further studies investigated relationships between foot posture and lower limb motion [13,15] (Table 1.1). The first and most recent of these studies modelled the foot as four rigid segments and compared 11 adults with 'typical' foot posture to 11 adults with ‘low-mobile’ (i.e. flat-arch and mobile) foot posture [13], categorised using arch height and arch mobility measurements. They reported that the low-mobile foot group exhibited significantly less abduction excursion of the ‘calcaneonavicular complex’ during midstance, and increased inversion excursion of the rearfoot during pre-swing, compared to those with normal foot posture.

The second study modelled the foot as two rigid segments and compared 15 males with a history of musculoskeletal symptoms linked to pes planus (i.e. flat-arched feet) with 18 healthy males. This study found that the participants with pes planus displayed significantly greater rearfoot plantarflexion during contact phase, less forefoot adduction at toe off, and less forefoot range of motion in the transverse plane over total stance, compared to the normal participant group [15].

Among these studies, several different methods of classifying foot posture were adopted, making it difficult to compare findings from otherwise similar studies. In addition, several of the classification strategies lack validity, such as the subjective clinical assessment and goniometric assessment of forefoot to rearfoot alignment [1]. While this indicates that consensus is lacking on a clear strategy for classifying foot posture, one consistent feature of these studies was the inclusion of an apparent flat-arch or pronated foot group. Other issues with this literature include: (i) given the large number of bones in the foot arch, it is unclear how many segments are required to accurately capture movement using a multi segment foot model and; (ii) several studies included a small sample size for the number of variables tested, potentially leading to Type II statistical error.

In summary, the literature indicates that flat-arched foot posture influences position and movement of different lower limb segments during walking, although the precise nature of this relationship is far from clear.

Table 1.1. A summary of key studies investigating the relationship between foot posture and lower limb kinematics during walking

Author/s	Participants	Foot posture measurements	Kinematic measurements	Key findings
Alonso-Vazquez et al [12]	10 children with forefoot varus compared to 11 'control' children without forefoot varus	Forefoot varus measured with goniometer (frontal plane alignment of forefoot relative the plantar aspect of the heel)	56 variables from 3-dimensional analysis of the lower limb	• Significantly less hip adduction and extension during loading response and midstance, respectively, with forefoot varus deformity
Cobb et al [13]	11 participants with 'typical' foot posture compared to 11 with 'low-mobile' foot posture	- Arch ratio - Relative arch deformity ratio	96 kinematic variables related to the rearfoot, calcaneonavicualar, medial forefoot and first metatarsophalangeal complexes	• Significantly decreased calcaneonavicular complex abduction excursion in participants with low-mobile feet compared to those with typical feet, during midstance • Significantly increased rearfoot complex inversion excursion in participants with low-mobile feet compared to those with typical feet, during pre-swing phase
Houck et al [14]	7 participants with normal-foot posture, 14 with abnormally pronated foot posture (18 female, 3 male)	- Goniometric measurement of forefoot to rearfoot angle - Rearfoot to ground angle - Navicular drop difference comparing subtalar joint neutral to relaxed	- Calcaneal eversion - First metatarsal dorsiflexion	• Significantly more inverted rearfoot in participants with pronated feet compared to those with normal feet, during initial contact • Significantly more everted rearfoot in participants with pronated feet compared to those with normal feet, during midstance
Hunt and Smith [15]	15 males with a history of musculoskeletal symptoms associated with 'pes planus' were compared to 18 normal males	- Subjective clinical assessment	Numerous foot and ankle parameters (not clearly defined)	• Pes planus group displayed significantly greater rearfoot plantarflexion at 21% of stance phase compared to the normal group • Pes planus group displayed significantly less forefoot adduction at toe off compared to the normal group • Pes planus group displayed significantly less forefoot range of motion in the transverse plane over total stance compared to the normal group

Plantar pressure studies

The second notable area of research pertaining to the effect of foot posture on lower limb biomechanics is the investigation of plantar pressure measurements during gait. Four of studies have utilised plantar pressure measurements to investigate whether foot posture influences dynamic function [16-19] (Table 1.2). As with the kinematic studies discussed in the previous section, these studies were similarly heterogeneous in their method of classifying foot posture.

Classification of foot postures were based on; visual observation [16], radiographic measurements [17], foot print measurements [18], and arch height measurements [19]. Despite the different methods of categorising foot posture, there was some consensus among these studies that high-arched or cavus foot postures display significantly lower pressure parameters in the medial arch region and increased pressure parameters under the heel and forefoot region during walking, compared to normal- and flat-arched feet.

Table 1.2. A summary of key studies investigating the effect of foot posture on plantar foot pressure during walking

Author/s	Participants	Foot posture measurements	Dynamic plantar pressure measurements	Key findings
Burns et al [16]	70 participants (41 female, 29 male). 30 had normal-arched foot type, 30 had idiopathic pes cavus and 10 had neurogenic pes cavus foot posture	Foot posture index	- Contact time - Contact area - Peak pressure - Pressure-time integral	• Compared to the group with normal-arched feet, the group with idiopathic cavus feet exhibited a smaller contact area beneath the midfoot, greater rearfoot peak pressure, greater pressure-time integral for the whole foot, the forefoot, midfoot and rearfoot
Cavanagh et al [17]	48 symptom-free participants	27 radiographic measures of foot posture	- Peak plantar foot pressure	• A higher-arched foot is associated with increased peak plantar pressure under the rearfoot and forefoot
Rosenbaum et al [18]	30 participants free of injuries. 10 with a normal-arched, 10 with flat-arched and 10 with high-arched feet.	Foot form index (foot print)	- Peak plantar pressure under eight regions of the foot	• Compared to the groups with normal-arched and flat-arched feet, the high-arched feet demonstrated the least pressure in the midfoot region
Teyhen et al [19]	1000 participants (566 males, 434 females). 693 had normal-arched feet, 142 had high or very high-arched feet and 165 had flat or very flat-arched feet.	Arch height index	- 200 plantar pressure measurements	• A higher-arched foot is associated with increased pressure in the lateral forefoot, increased lateral excursion of gait line, increased force-time integral of the lateral hind foot and first metatarsal region

Summary

To summarise the information presented in this section, there is limited evidence that flat-arched and high-arched feet exhibit functional differences during gait, compared to either normal-arched feet and to each other. The major issue when reviewing this literature relates to the array of methods employed to classify foot posture. In addition, some studies may have failed to identify ‘normal’ or true ‘extremes’ of arch height due to issues relating to the validity of the tests. Only in the last decade has normative foot posture data for various validated clinical and radiolographic measurements been published [2,20-23]. The availability of such data means that it is now easier for researchers to quantify and categorise normal and extremes of foot posture based on distribution of foot posture within the population. Whilst there still remains uncertainty regarding the effect of foot posture on lower limb biomechanics during walking, foot posture has long been considered to play a role in predisposition to injury.

1.2.2 Foot posture and lower limb injury

In section 1.2.1 the relationship between foot posture and lower limb biomechanics was discussed. A framework was provided detailing how flat-arched foot posture may influence joint congruency and subsequent changes in motion and stresses on soft tissue structures. It is widely thought that these events may contribute to musculoskeletal injury. Therefore, to further investigate the significance of variations in foot posture, the following section presents an overview of the literature relating to the effect of foot posture on lower limb injury.

One systematic review [24] and two narrative literature reviews [25,26] have investigated the relationship between foot posture and injury (Table 1.3). Each review presented a descriptive account of the research findings from the included studies and none pooled data to conduct meta-analyses.

Only the systematic review [24] conducted a methodological assessment of the included studies. This review assessed the association between foot type and tibial stress injuries among three prospective [27-29] and six retrospective studies [30-35] of military and sporting populations. The authors concluded that although no definitive link between foot posture and increased risk of tibial stress injuries was evident, very low- and very high-arched feet are likely to increase the risk of tibial stress injuries compared to normal-arched feet.

Another review investigated foot characteristics associated with lateral ankle injury, such as lateral ankle sprain [25]. The study populations included military recruits, infantry recruits, college athletes and physical education students. Four studies used a prospective design [36-39] and five used a retrospective design [40-44]. Only cavovarus foot posture (i.e. high-arched foot) had a significant association with lateral ankle injuries. Other anthropometric measures associated with lateral ankle injuries included increased foot width and increased calcaneal eversion range of motion.

Table 1.3. Overview of the key systematic reviews and literature reviews investigating the relationship between foot posture and lower limb injury

Author/s	Details of studies included in review	Study populations	Types of injury	Meta-analysis or descriptive?	Key findings
Barnes et al [24]	9 studies with sample sizes ranging from 40 to 505. - 3 studies were prospective - 6 studies were retrospective	6 sporting 3 military	Tibial stress fractures	Descriptive	• No definitive link between foot posture and increased risk of tibial stress injuries • Very low- and very high-arched feet may increase the risk of tibial stress injuries compared to normal arched feet
Morrison et al [25]	9 studies with sample sizes ranging from 13 to 390 - 4 studies were prospective - 5 studies were retrospective	1 military recruits 1 infantry recruits 2 college athletes 1 physical education students 4 mixed or unable to determine	Inversion ankle injury	Descriptive	• Cavovarus foot deformity, increased foot width and increased calcaneal eversion range of motion may be associated with inversion ankle injury
Murphy et al [26]	9 studies with sample sizes ranging from 40 to 423 - 7 studies were retrospective - 1 study was a non-randomised trial - 1 study was a randomised controlled trial	3 military 2 runners 1 football 1 cross country 1 physical education students 1 soccer 1 field hockey 1 lacrosse 1 basketball	All injuries	Descriptive	• Five studies reported an association between foot morphology and injury; four showed no association • Risks for lower extremity stress fracture includes having high arches or a supinated foot type

One further review [26] investigated various intrinsic and extrinsic risk factors for lower extremity injury, with 'foot morphology' included as a sub-category of intrinsic factors. The review included eight prospective studies [27,35,36,45-49] and one clinical trial [50]. The study populations comprised military recruits, runners, physical education students and participants involved in various individual sports. The review concluded that five of the studies reported an association between foot morphology and injury [27,35,45,46,51] and four studies showed no association [36,47-49]. One risk factor identified for lower extremity stress fracture was having a high-arched or a supinated foot type [27,46].

In summary, these three reviews indicate that both low- and high-arched foot posture are associated with exercise-related lower extremity injury. However, all three reviews reported limitations with the included studies such as small sample size, over-representation of one sex, and the infrequent use of established methods for classifying injury severity. Moreover, a fundamental concern with the studies reviewed is when they were conducted there were no rigorously validated methods of classifying foot posture. Only recently have such techniques for rating foot posture become available. For example, one such technique that has been shown to be valid [52] and reliable [2,53-55], the Foot Posture Index (FPI), was first described by Redmond and colleagues [54] in 2006. Following this, several prospective studies were published which included validated tests for assessing foot posture – these were not included in the aforementioned reviews. The more recent studies can be divided into those that classified foot posture based on *static* foot posture measurements (Table 1.4) and those that classified foot posture based on *dynamic* measurements (Table 1.5). Studies using *static* foot posture measurements included clinical tests such as arch height and footprint measurements [56-61], and *dynamic* foot posture measurements included pressure mats to evaluate plantar pressures and forces [62,63].

To illustrate the relationship between foot posture and injury, patellofemoral pain can be used as an example, which has a high incidence compared to other musculoskeletal injuries affecting the lower limb. Patellofemoral pain is estimated to represent 9-19% of exercise related injuries [34,64]. Several studies have investigated the relationship between foot posture and this condition [58,60,62,64,65]. Lun and colleagues [60] conducted a prospective study over 6 months comprising 87 runners. Compared to runners without injury, the six participants who developed patellofemoral pain displayed significantly less static forefoot varus of the left foot. Although, the finding of an association between less left-sided forefoot varus and patellofemoral pain should be interpreted with some caution, as this result may just reflect a 'chance finding', even though the study was probably under-powered with only six participants.

Other prospective studies utilising both static and dynamic measurements reported that only dynamic measurements, for instance plantar pressures recorded during running, were associated with patellofemoral pain [62,65]. Unfortunately, these dynamic measurements are not necessarily associated with features of either a pronated or supinated foot. For example, Thijs and co-workers [62,65] conducted two prospective studies; one comprising 143 'novice recreational runners' and the other involving 84 'officer cadets'.

Table 1.4. Additional key studies investigating the relationship between static foot posture measurements on lower limb injury

Author/s	Participants	Static foot posture measurements	Types of injury	Study design/ follow-up	Key findings
Burns et al [57]	131 triathletes - 12 supinated foot - 111 normal foot - 8 pronated foot	• Foot Posture Index • Valgus index	All overuse and traumatic injuries	Two studies: (i) 6 month retrospective/ no follow-up (ii) Prospective/ 10 week follow-up	• Supinated foot type significantly greater likelihood of sustaining overuse injury during competition
Levy et al [59]	512 newly entered cadets - 33 pes planus	• Arch height based on midfoot ratio	All lower limb injuries	Retrospective/ 46 months follow-up	• Left-sided pes planus significantly related to left-sided midfoot injuries, right-sided midfoot injuries and left knee injuries • Right-sided pes planus significantly related to right knee injuries
Lun et al [60]	87 runners - 4 pes cavus - 58 neutral - 53 pes planus	Subjective clinical categorisation • *Longitudinal arch:* - pes planus - pes cavus - neutral • *Standing ankle pronation:* - neutral - mild - moderate - severe	All lower limb injuries	Prospective/ 6 months follow-up	• Significantly lesser left forefoot varus in runners with patellofemoral pain syndrome compared with non-injured runners

Table 1.4. Continued – additional key studies investigating the relationship between *static* foot posture measurements on lower limb injury

Author/s	Participants	Static foot posture measurements	Types of injury	Study design/ follow-up	Key findings
Michelson et al [61]	196 college athletes from 10 different sports - 56 pes planus	• Harris mat footprint with six regions demarcated	All lower limb injuries	Prospective/ unable to determine follow-up	• Pes planus not a risk factor for any injury of the lower extremity
Nakhaee et al [56]	47 professional male runners - 28 normal - 5 low-arched - 14 high-arched	• Clinical navicular drop test	All lower limb injuries	Cross-sectional/ retrospective	• Higher or lower medial longitudinal arch are not definite risk factors for sports related injuries
Pierrynowski et al [66]	53 healthy university students	• Subtalar joint inclination angle estimated with infrared markers	Foot and knee	Cross-sectional/ no follow-up	• Significantly greater subtalar joint inclination angle for the knee injury group
Tomaro et al [67]	5 males and 15 females	• Subtalar joint motion measured as a ratio of transverse plane motion to frontal plane motion	Foot, leg and knee	Cross-sectional/ no follow-up	• Significantly lower subtalar joint ratio in participants with foot symptoms compared to leg and knee symptoms
Witvrouw et al [64]	282 physical education students (151 males, 131 female)	• Footprint measured with pedograph; forefoot to rearfoot alignment	Patellofemoral pain syndrome (PFPS)	Prospective/ 24 month follow-up	• Foot posture not significantly different between participants with PFPS compared to participants without PFPS

Table 1.5. Additional key studies investigating the relationship between *dynamic* (i.e. gait related) foot posture measurements and lower limb injury

Author/s	Participants	Dynamic foot posture measurements	Types of injury	Study design/ follow-up	Key findings
Levinger and Gilleard [58]	27 female participants (13 with patellofemoral pain syndrome, 14 asymptomatic controls)	• Triplanar motion of the rearfoot relative to the tibia	Patellofemoral pain syndrome (PFPS)	Cross-sectional comparative/ no follow-up	• Significantly delayed peak rearfoot eversion, earlier peak dorsiflexion, lower peak ground reaction force, prolonged rearfoot eversion during stance phase with PFPS compared to controls
Thijs et al [65]	143 novice recreational runners (54 men, 89 women)	• Rollover pattern using pressure plate and foot posture index	Patellofemoral pain syndrome (PFPS)	Prospective/ 10 week follow-up	• Significantly higher peak vertical force underneath lateral heel and 2nd and 3rd metatarsals in runners who developed PFPS compared to uninjured runners
Thijs et al [62]	84 officer cadets (65 men, 19 women)	• Plantar pressure from pressure plate	Patellofemoral pain syndrome (PFPS)	Prospective/ 6 week follow-up	• Significantly more laterally directed pressure, shorter time to maximum pressure on fourth metatarsal, slower maximum velocity of change in latero-medial direction of centre of pressure with PFPS compared to uninjured cadets
Van Ginckel [63]	129 able-bodied novice runners (19 men and 110 women)	• Force distribution from pressure plate	Achilles tendinopathy	Prospective/ 10 week follow-up	• Significant decrease in total posterior-anterior displacement of centre of force, laterally directed centre of force during foot flat with Achilles tendinopathy compared to uninjured runners

The key findings from these studies were that, compared to uninjured participants, those who developed patellofemoral pain displayed significantly higher peak vertical force underneath the lateral heel and the 2nd and 3rd metatarsal; more laterally directed pressure distribution; shorter time to maximum pressure on the 4th metatarsal; and slower maximum velocity of change in latero-medial direction of centre of pressure.

While these findings suggest that a relationship between foot function and injury may exist, such measures of foot function are difficult to assess and cannot readily be applied to clinical practice. There are numerous other prospective studies that report specific aspects of foot posture or foot function to be associated with the development of lower limb injuries [57-59,66,67] (Table 1.4 and 1.5), however, these studies are difficult to compare because they are not homogenous in the type of injuries studied and the methods of classifying foot posture.

In summary, although it is widely accepted that high-arched and flat-arched foot posture increases the likelihood of developing specific lower limb injuries, there is scant evidence to support this notion at this stage. Further prospective research is required, using valid and reliable methods of classifying foot posture. In this case, the classification of abnormal foot posture should be based on population-based data rather than hypothetical deformities.

Despite the uncertainty surrounding the relationship between foot posture and lower limb injury, foot orthoses are widely used for the management of musculoskeletal injuries associated with foot posture anomalies. The basis for using foot orthoses is to normalise lower limb motion, forces and muscle activity [68], thereby reducing symptoms. If foot orthoses are shown to be effective at reducing symptoms and modifying biomechanical parameters, then this may add support to the theory that foot posture and lower limb injury are related.

With this in mind, the following section presents an overview of the literature that has investigated; (i) the biomechanical effects, and (ii) the clinical effectiveness of foot orthoses for treating lower limb injury.

1.3 Effect of foot orthoses on lower limb biomechanics and injury

1.3.1 Foot orthoses and lower limb biomechanics

Foot orthoses are medical devices that are widely used to treat conditions affecting the foot and lower limb. The terminology used to describe different types of foot orthoses is complex because a wide range of materials and manufacturing processes are available. These devices include, but are not exclusive to; accommodative, functional and pre-fabricated foot orthoses [69]. Due to the diverse range of devices there is no consensus for using a particular device for treating specific conditions [68-71].

Despite the lack of consensus, it is generally believed that foot orthoses support and align the foot, thus improving function [72]. Over the last three decades, a substantial amount of research has been undertaken in gait laboratories to determine whether foot orthoses do indeed support, align and improve function of the foot.

Laboratory-based studies have principally used three techniques for investigating the effect of foot orthoses on lower limb function. These techniques are: (i) kinematics; (ii) kinetics or plantar pressures; and (iii) electromyography [73].

The following section presents an overview of the kinematic and kinetic effects of foot orthoses during gait. For clarity, only key studies involving healthy participants during walking are included in this section of the literature review (section 1.3.1), as non-healthy participants may walk differently due to pain, which may confound or mask any direct effects of the foot orthoses. The electromyographic effects of foot orthoses are presented separately in a systematic review in Chapter 2 as the first published study in this thesis.

(i) Kinematic studies

Several studies have investigated the effect of foot orthoses on lower limb motion during walking [74-78]. Various styles of foot orthoses have been studied, these include; ‘simple insoles’ [74], semi-customised and customised foot orthoses [75,78], moulded foot orthoses with different wedging configurations [76] and participants’ own existing semi-rigid and rigid foot orthoses [77] (Table 1.6). The majority of studies focused on the capacity of foot orthoses to control rearfoot motion during stance phase.

Table 1.6. A summary of studies investigating the effect of foot orthoses and lower limb kinematics during walking

Author/s	**Participants**	**Type of foot orthosis** (Control condition)	**Kinematic measurements**	**Key findings**
Branthwaite et al [74]	9 active males without history of orthotic therapy	- Simple insoles' - Biplanar insole - Cobra insole (Trekking sandals)	4 variables related to foot dorsiflexion, eversion, adduction angles and eversion velocity	• Biplanar insoles significantly reduced maximum eversion compared to the trekking sandals alone
Davis et al [75]	19 healthy recreational runners	- Semi-custom - Custom (Neutral running shoes)	- Peak eversion angle - Peak eversion velocity - Eversion excursion - Eversion duration	• Eversion excursion decreased significantly with semi-custom compared to the custom and no orthotic (shoe only) conditions
Johanson et al [76]	22 participants (9 males, 13 females) with forefoot varus deformities	- Moulded orthosis with rearfoot and forefoot posting - Moulded orthosis with rearfoot posting - Moulded orthosis with forefoot posting - Moulded orthosis with no posting (running shoe)	- Maximum calf-to-calcaneus angle in pronation (CCAmax) - Degrees of pronation between heel strike and CCAmax - Maximum calcaneus-to-vertical angle in eversion (CVAmax) - Degrees of eversion between heel-strike and CVAmax	• CCAmax and CVAmax decreased significantly with both forefoot and rearfoot posting compared to forefoot posting alone • CCAmax decreased significantly with all orthotic conditions compared to the shoe only condition • CCAmax decreased significantly with all orthotic conditions (except unposted shell) compared to the shoe only condition
McCulloch [77]	10 participants (5 males, 5 females) with existing functional foot orthoses	- 7 subjects wore rigid orthotic inserts - 3 wore semi-rigid orthotic inserts (personal athletic shoes)	- Time to maximum pronation - Time to heel rise - Rate of pronation during first 10% of stance - Rate of pronation during second 10% of stance - Maximum pronation, ankle dorsiflexion and knee flexion	• Pronation reduced significantly throughout stance with foot orthoses • Duration of stance time significantly increases with foot orthoses

Table 1.6. – Continued. A summary of studies investigating the effect of foot orthoses and lower limb kinematics during walking

Author/s	**Participants**	**Type of foot orthosis** (Control condition)	**Kinematic measurements**	**Key findings**
Zifchock and Davis [78]	37 participants (17 males, 20 females) recreational runners without foot orthoses	- Semi-custom - Custom (neutral running shoes)	- Peak eversion angle - Peak eversion velocity - Eversion excursion - Eversion duration	• Eversion excursion decreased significantly with semi-custom and custom compared to no orthotic (shoe only)

There is some evidence that indicates foot orthoses decrease aspects of rearfoot eversion including; peak eversion [74], eversion excursion [75,78], and eversion duration [77]. However, there is insufficient evidence that different styles of foot orthoses, such as prefabricated and customised orthoses, have differerent affects on rearfoot motion during walking.

One common limiting factor with studies investigating the effect of foot orthoses on lower limb kinematics is that most had small sample sizes. Two studies investigated ten or less participants, while the remaining three studies included between 19 and 37 participants. Such small sample sizes may have led to low statistical power, potentially leading to type II statistical error. Another concern with these studies relates to measurement error. That is, measuring foot motion using 2- and 3-dimensional analysis techniques may be inaccurate due to the relative skin movement over the small bones of the foot.

In addition to these issues, there has been further criticism of previous studies that investigated the kinematic effects of foot orthoses. For example Nigg [79] has stated that:

> *"Evidence suggests that the concept of aligning the skeleton with shoes, inserts, and orthotics should be reconsidered. They produce only small, not systematic, and subject-specific changes of foot and leg movement"* (*page 2).*

Nigg [79] went on to theorise that impact forces during the stance phase could provide an input signal that produces ‘muscle tuning’ shortly before the next contact with the ground. The associated muscle activity may minimise soft tissue vibration and/or reduce joint and tendon loading. Nigg [79] also proposed that kinematic studies fail to detect significant changes in motion because compensatory ‘muscle tuning’ maintains the kinematic and kinetic situations for a given task. Taking this issue into account, Nigg [79] proposed a new paradigm in 2001 suggesting that the relationship between foot pronation, injury, and the success of foot orthoses is more likely to be related to alterations in muscle activity than joint kinematics.

While recognising the contribution of Nigg to current thinking, there are some issues with his views on the kinematic and kinetic effects of foot orthoses. For example, since Nigg’s paradigm was published in 2001, there is now evidence that significant motion occurs in the midfoot and forefoot [80], and therefore the kinematic effects of foot orthoses may be of considerable magnitude when considered across all the joints of the foot.

(ii) Plantar pressure studies

Four key studies have investigated the effect of foot orthoses on plantar pressure measurements [81-84]. Two studies included asymptomatic participants [81,82] while the other two studies included participants issued with foot orthoses for musculoskeletal complaints [83,84] (Table 1.7). These studies investigated a range of orthoses, including semi-rigid customised foot orthoses [81-84], prefabricated foot orthoses [82] and flat non-cast insoles with medial heel wedging - these were compared to barefoot [83] or shoe-only ‘control’ conditions [81,82,84]. Two studies involved participants who exhibited ‘excessive pronation’ or flat-arched foot posture [81,82]. One study assessed participants that had musculoskeletal problems but ‘normal’ foot posture [83], while the other study did not indicate what type of foot posture participants displayed [84].

The findings of these studies indicate that contoured foot orthoses have systematic effects on plantar pressures. More specifically, contoured foot orthoses *increase* plantar pressures in the medial arch and *decrease* plantar pressures under the heel and forefoot region [81,82]. While some of the weaknesses evident in these studies, for example small sample size, limit the conclusions that can be drawn from this research, foot orthoses appear to have a relatively predictable effect on plantar pressure during gait.

Plantar pressure measurements may have an important role in assisting our understanding of both the causes of overuse conditions of the lower limb and the effects of various interventions to treat these conditions. For example, it is known that elevated plantar pressure is a causative factor in the development of certain types of plantar ulcers in people with diabetes [85]; and related to this, some offloading interventions prevent and heal such foot ulcers and reduce plantar pressure [86]. However, currently it is unknown whether musculoskeletal injuries are related to alterations in the magnitude or distribution of plantar pressures.

Table 1.7. A summary of key studies investigating the effect of foot orthoses on plantar foot pressure during walking

Author/s	**Participants**	**Type of foot orthosis** (Control condition)	**Plantar pressure measurements** (Measurement apparatus)	**Key findings**
Bennett et al [83]	22 participants with history of foot and leg problems	- Running shoes - Prescription orthoses with 4° medial wedge - (Barefoot)	(Electrodynogram) - Peak plantar pressure - Time to peak pressure	• Maximum pressure was loaded earlier (5%-7%) on the lateral border of the foot during the gait cycle
Reed and Bennett [84]	27 participants issued with foot orthoses for musculoskeletal complaints	- 'Root' style device [orthosis with midfoot support] - 'Blake' style device [orthosis with rearfoot support] - (Shoe only)	(Electrodynogram) - Peak pressure - Time to peak pressure - Pressure duration - Duration of load - Total pressure per second	• Duration of some of the components of stance phase was altered • Initiation of loading beneath the medial forefoot was delayed • Reduction in the total duration of loading at discrete sites beneath the heel and forefoot • Effects of the two orthoses were similar
Redmond et al [81]	22 healthy individuals with excessive pronation	- Modified Root device/customised orthosis - Non-cast insole with 6 degree varus rearfoot wedge - (Thin sole athletic shoe)	(Novel Pedar ®) - Peak pressure - Maximum mean pressure - Pressure time integral - Maximum force - Force time integral - Area - Time	• Customised foot orthoses increase pressure area in the midfoot compared to the shoe only condition • Customised foot orthoses decrease medial and lateral forefoot and heel pressures, compared to the shoe only condition
Redmond et al [82]	15 participants with flat-arched feet	- Semi-rigid customised - Semi-rigid prefabricated - (Shoe only)	(Novel Pedar ®) - Peak pressure - Maximum mean pressure - Pressure time integral - Maximum force - Force time integral - Area - Time	• Customised and prefabricated foot orthoses increase force and force time integral in the midfoot region, compared to the shoe only condition • Customised and prefabricated foot orthoses decrease peak, maximum mean pressure, pressure time and force time integrals in the medial and lateral forefoot region, compared to the shoe only condition

(ii) Electromyographic studies

The third technique commonly used to investigate the effect of foot orthoses on lower limb function is electromyography, which is the main focus of this thesis. For this reason, Chapter 2 provides a comprehensive review of literature in the area utilising systematic review methodology.

1.3.2 Foot orthoses and lower limb injury

As mentioned in Section 1.3.1, foot orthoses are a medical device used widely to treat conditions affecting the lower limb and foot. Foot orthoses have been investigated for both their effect on *preventing* lower limb injuries and their effect on *treating* lower limb injury. The use of foot orthoses to *prevent* lower limb injuries has been investigated by three recent systematic reviews [68-70]. Further, the use of foot orthoses to *treat* lower limb injuries was investigated by two of these reviews [69,70] and one additional review [71] (Table 1.8). Six other Cochrane systematic reviews [87-93] related to foot orthoses could be included in this discussion, however the data related to foot orthoses in these reviews was extracted by a single systematic review by Hume and colleagues [69] and has therefore been superseded by this review.

The following section presents a summary of the four key systematic reviews that investigated the effect foot orthoses on lower limb injury [68-71]. They predominantly included studies that investigated military populations and community-dwelling participants from non-sporting populations. The participants in these trials received foot orthoses for either the treatment of a lower limb injury or prior to an exercise period.

Table 1.8. Overview of systematic reviews and literature reviews investigating the effect of foot orthoses on lower limb injury

Author/s	Details of studies included in review	Study populations	Classification of foot orthoses	Types of injury	Meta-analysis or descriptive?	Key findings**
Collins et al [70]	23 randomised trials, including 8 that evaluated prevention of lower limb overuse injury and 15 that investigated the treatment of injury	Prevention studies - Military personnel Treatment studies - Mixed	- Control - Prefabricated customised - Prefabricated semi-rigid - Prefabricated silicon - Custom (casted)	*Prevention studies* Lower limb: - pain or injury - stress fractures - back pain/injury - 'problems' (stress fractures, ankle sprains and foot problems *Treatment studies* - Plantar fasciitis - Antero-medial knee pain - Anterior knee pain - Lesser metatarsalgia - Morton's neuroma - Heel pain - Primary metatarsalgia - Myofascial pain syndrome of peroneus longus - Heel spur syndrome	Meta-analysis	• Pooling of data supported the use of foot orthoses for preventing lower limb overuse conditions in military populations • Insufficient evidence to support or refute the use of foot orthoses for treating overuse injuries
Landorf and Keenan [68]	12 randomised or quasi randomised trials	11 Military personnel 1 Soccer referees	*Shock-absorbing:* - Heel pad - Prefabricated *Motion-controlling:* - Semi-rigid prefabricated - Semi-rigid - Customised soft	Varied – some evaluated injury broadly, others focused on specific injuries such as medial tibial stress syndrome or stress fracture	Descriptive	• Motion-controlling foot orthoses decrease the incidence of stress fractures (particularly femoral) and shin splints • Sock-absorbing insoles or heel pads do not prevent injury

** Key findings were deemed by the reviewer to be of either statistical or clinical significance

Table 1.8. Continued - overview of systematic reviews and literature reviews investigating the effect of foot orthoses on lower limb injury

Author/s	Details of studies included in review	Study populations	Classification of foot orthoses	Types of injury	Meta-analysis or descriptive?	Key findings**
Hume et al [69]	15 randomised controlled trials, controlled clinical studies, uncontrolled clinical studies, Cochrane and systematic reviews	Military and mixed populations	- Customised semi-rigid - Customised soft - Prefabricated semi-rigid - Prefabricated soft	- Plantar fasciitis - Tibial stress fractures - Patellofemoral pain syndrome	Descriptive	• Plantar fasciitis – rigid and semi-rigid foot orthoses have a moderate beneficial effect; soft foot orthoses have a small beneficial effect • Patellofemoral pain – soft foot orthoses have a small beneficial effect • Posterior tibial stress fractures – semi-rigid and soft foot orthoses have a small beneficial effect
Hawke and Burns [71]	11 randomised controlled trials and non-randomised controlled trials	Mixed – mean age in treatment groups ranged from 13 to 63 years	- Sham - Custom-made - Prefabricated	- 5 plantar fasciitis - 3 rheumatoid arthritis - 1 pes cavus - 1 hallux valgus - 1 juvenile idiopathic arthritis (JIA)	Descriptive	• Custom-made foot orthoses were effective for painful pes cavus, rearfoot pain in rheumatoid arthritis, foot pain in JIA and painful hallux valgus • Prefabricated orthoses just as effective as custom-made orthoses for JIA

** Key findings were deemed by the reviewer to be of either statistical or clinical significance

With respect to the statistical approach used to summarise the data, all the reviews presented rigorous statistical analyses to compare interventions, such as confidence intervals and or risk/odds ratios. However, only Collins and colleagues [70] conducted meta-analyses from pooled data.

The first review by Collins and colleagues, which included meta-analyses, reviewed 23 randomised controlled trials (RCTs) [70]. Studies were separated according to those that investigated prevention [29,94-99] and those that investigated treatment [92,100-113] of lower limb injuries. Various styles of foot orthoses were reported, ranging from prefabricated silicon inserts to customised (cast) foot orthoses. Pooling of data in this review supported the use of foot orthoses to prevent lower limb overuse conditions in military populations. However, insufficient evidence was available to support or refute the use of foot orthoses for the treatment of overuse injuries.

The second review by Landorf and Keenan [68] supported the conclusions of Collins and colleagues [70] that foot orthoses have a role in injury prevention [68]. This study reported evidence from 12 randomised and quasi-randomised trials [29,94-98,114-119], including some additional studies compared to the work of Collins and colleagues [70]. The review suggested that motion-controlling foot orthoses decrease the incidence of stress fractures (particularly femoral) and shin splints. In contrast to the motion-controlling foot orthoses, they concluded that shock-absorbing insoles or heel pads do not prevent injury.

The third and more recent review by Hume and co-workers [69] differs to the conclusion by Collins et al [70] that ‘insufficient evidence supports or refutes the use of foot orthoses in the treatment of overuse injuries’. In this investigation, 15 clinical trials (both controlled and uncontrolled) were reviewed. Foot orthoses were categorised as customised or prefabricated, with the additional classification of soft or semi-rigid. Specific overuse injuries were investigated, which included; plantar fasciitis, patellofemoral pain and tibial stress fractures. The authors concluded that for some conditions, rigid and semi-rigid foot orthoses have a moderate beneficial effect, and soft foot orthoses have a small beneficial effect.

Finally, the fourth study was a Cochrane systematic review [71] that investigated 11 randomised and non-randomised controlled trials [100,105,107,108,120-126] related to the use of custom-made foot orthoses for the treatment of foot pain. Unlike the other systematic reviews discussed above, this review also included studies that evaluated systemic conditions (e.g. juvenile idiopathic and rheumatoid arthritis) and foot pain related to foot posture or deformity (e.g. pes cavus and hallux valgus), rather than predominantly exercise-related injuries. Customised foot orthoses were found to be effective for painful pes cavus, rearfoot pain in rheumatoid arthritis, foot pain in juvenile idiopathic arthritis and painful hallux valgus. Prefabricated foot orthoses were reported to be just as effective as custom-made orthoses for juvenile idiopathic arthritis.

The trials included in these four reviews need to be interpreted in light of several limitations. The most common issues included; the lack of consensus for the terminology used to define different types of foot orthoses, the relative short intervention periods, and a lack of consistent outcome measures. Despite these shortcomings, the reviews provide high-level evidence that foot orthoses are effective for preventing and treating some lower limb injuries.

1.4 Summary of the research problem

With the previous literature reviewed in mind, there are two key rationales for undertaking this thesis. Firstly, variations in foot posture are associated with lower limb injury, and secondly, foot orthoses aimed at altering foot posture are effective for preventing and treating some lower limb injuries. Therefore, there is a need to explore the physiological response of the body to variations in foot posture and interventions that might affect foot posture, such as foot orthoses (Figure 1.3). Such research may assist in our understanding of both the mechanism of overuse conditions of the lower limb and the effects of various treatment approaches. This will ultimately lead to more effective interventions for the prevention and treatment of lower limb injury. Accordingly, this thesis investigates the effect of foot posture and foot orthoses on lower limb muscle activity during walking.

Figure 1.3. Flow chart presenting the relationship between key concepts presented in Sections 1.2 and 1.3 of the literature review.

Section 1.2 Foot posture influences lower limb biomechanics and increases the risk of lower limb injury

Section 1.3 Foot orthoses alter lower limb biomechanics and are effective for preventing and treating lower limb injuries

Key research question: What is the muscular response of the body to variations in foot posture and foot orthoses?

Chapter 2 Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review

CHAPTER 2

Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review

Preface

Sections 1.2 and 1.3 of the literature review provide an overview of the effect of foot posture and foot orthoses on lower limb biomechanics and injury. The literature presented in section 1.2 indicates there is limited evidence that variations from ‘normal’ foot posture influence lower limb motion and plantar pressures, and increase the risk of injury. The literature presented in section 1.3 indicates that foot orthoses influence lower limb motion and plantar pressures, and are effective for preventing and treating some lower limb injuries. Although there were methodological limitations in several of the studies reviewed, evidence from this body of literature provides a case for investigating the relationship of foot posture and foot orthoses on muscle activity during gait.

Accordingly, the primary aim of the systematic review presented in this chapter was to determine whether there is evidence that foot posture, foot orthoses and footwear affect lower limb muscle activity during walking or running. The review was undertaken to inform the design of subsequent studies in this thesis, identify methodological issues within the literature, and to identify gaps in current evidence.

The study in this chapter has been published:

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Review

Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: A systematic review

George S. Murley [a,b,*], Karl B. Landorf [a,b], Hylton B. Menz [b], Adam R. Bird [a,b]

[a] Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, VIC 3086, Australia
[b] Musculoskeletal Research Centre, Faculty of Health Sciences, La Trobe University, Bundoora, VIC 3086, Australia

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Electromyography

ABSTRACT

The aim of this systematic review was to evaluate the literature pertaining to the effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running. A database search of Medline, CINAHL, Embase and SPORTDiscus without language restrictions revealed 504 citations for title and abstract review. Three articles were translated to English and a final 46 articles underwent a two-tiered quality assessment. First, all articles were scored for their reporting of electromyographic methodology using a set of standards adopted by the International Society of Electrophysiology and Kinesiology. Thirty-eight articles displayed adequate reporting of electromyographic methodology and qualified for detailed review including a second quality assessment using a modified version of the Quality Index. These included six studies investigating the effect of foot posture, 12 the effect of foot orthoses and 20 the effect of footwear on lower limb muscle activity during walking or running. Meta-analysis was not conducted due to heterogeneity between studies. Some evidence exists that: (i) pronated feet demonstrate greater electromyographic activation of invertor musculature and decreased activation of evertor musculature; (ii) foot orthoses increase activation of tibialis anterior and peroneus longus, and may alter low back muscle activity; and (iii) shoes with elevated heels alter lower limb and back muscle activation. Most studies reported statistically significant changes in electromyographic activation, although these findings were often not well supported when confidence intervals were calculated. Most important, however, is that there is a need for further research of more rigorous methodological quality, including greater consensus regarding standards for reporting of electromyographic parameters.

Contents

* Corresponding author at: Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, VIC 3086, Australia. Tel.: +61 3 9479 5776; fax: +61 3 479 5768.
E-mail address: g.murley@latrobe.edu.au (G.S. Murley).

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1. Introduction

Some variations in foot morphology, such as flat- or high-arched foot type, have long been recognised to cause tissue stress that results in injury [1]. Although there are many variations of flat- or high-arched feet that may or may not be functionally abnormal, some prospective studies provide evidence that flat- or high-arched feet increase the risk of lower limb injury [2–7]. However, previous systematic reviews have found a lack of agreement between studies that have evaluated the association between foot posture and injury, with almost as many studies supporting a link as there were studies not supporting a link [8,9]. Despite this uncertainty, it is widely accepted that foot posture, as well as other extrinsic factors such as age and skill level, combine to influence the risk of injuries in sport [8,9].

The mechanism linking variations in foot structure and musculoskeletal injury remains unclear. Nevertheless, several lower limb injuries associated with abnormal foot posture are widely treated with foot orthoses and footwear modification. Systematic reviews have found evidence that foot orthoses can prevent some lower limb overuse injuries, particularly femoral stress fractures and shin splints [10,11]. However, these reviews highlight that further research in this area is still required, particularly in the form of high quality randomised controlled trials (RCTs).

Whilst the use of RCT methodology is fundamental for determining the efficacy of interventions [12,13], laboratory-based biomechanical studies are required to explore the physiological response of the body to variations in foot posture and interventions such as foot orthoses. Laboratory-based studies may lead to further insights regarding the underlying mechanism that causes injury (e.g. altered plantar pressures and motion). This information can subsequently be used to develop effective interventions.

The biomechanical literature has principally focused on three techniques for evaluating the effect of foot posture, foot orthoses and footwear on lower limb function. These techniques include: (i) kinematics; (ii) kinetics or plantar pressures; and (iii) electromyography (EMG) [14]. Skeletal muscle function has obvious interactions with bone, joint, tendon, energy consumption and fatigue. Muscle activation may have a more complex relationship with overuse injury. Therefore, the aim of this systematic review was to determine whether there is evidence that foot posture, foot orthoses and footwear affect lower limb muscle activity during walking or running. Accordingly, the purpose of this review was to inform researchers planning future study of related conditions (i.e. foot posture) and interventions (i.e. foot orthoses and footwear).

2. Methods

2.1. Search strategy

To identify studies relating to the effect of foot posture, foot orthoses and footwear on dynamic lower limb muscle activity, an electronic database search was performed using OVID including Medline (1982–2007), Medline® In Process and Other Non-Indexed Citations (December 2007), CINAHL (1982–2007), Embase (1988–2007) and SPORTDiscus (1830–2007). A set of search terms were explored and derived from Medical Subject Headings (MeSH). To broaden the search strategy, some search terms were truncated and wildcard symbols were applied (Table 1). A random search of online biomechanically-related journals was conducted to ensure the database search was sensitive to relevant articles. The final database search was completed without language restrictions.

2.2. Inclusion criteria

To identify relevant studies, all titles and abstracts yielded from the search strategy were assessed by a single reviewer (GSM). Studies were included for the subsequent quality assessment if the following criteria were all satisfied:

i. Main outcome measure for muscle activity was either EMG or muscle function MRI during walking or running;
ii. Independent variables included either variation in foot posture, foot orthoses or footwear;
iii. Hypothesis testing with statistical analysis was carried out;
iv. Human participants without neurological disease were tested;
v. Participant sample size was greater than $N = 1$.

Studies assessing the effect of postural perturbations on muscle activity were not included in this review.

Table 1
Search strategy

Ovid interface (504 citations – all titles and abstract reviewed) – Updated 13/12/2007
OVID – CINAHL (<1982 to December week 1 2007>), *EMBASE* (<1988–2007 week 49>), *Ovid MEDLINE® In-Process & Other Non-Indexed Citations* (<December 12, 2007>), *Ovid MEDLINE®* (<1966 to November Week 2 2007>), *SPORTDiscus* (<1830 to November 2007>)
1. (Foot or pes) and (dysfunction or type or posture or flat or pronat$ or supinat$ or arch$ or cavus or planus or planovalgus or evert$ or invert$ or motion or structure)
2. (Foot or shoe or ankle) and (orthot$ or insert or wedg$ or orthos$ or insole or brace)
3. Shoe$
4. Electromyograph$ or EMG or IEMG or muscle function or mfMRI or (muscle and function MRI)
5. Walk$ or run$ or gait or locomotion or jog$
6. (1 or 2 or 3) and (4 and 5)
7. Remove duplicates from 6

Table 2
Phase 1: Reporting of EMG methodology

Criteria	Variables assessed
1. Surface EMG sensors [16,17]	Shape, size, material, construction
2. Sensor placement [16,17]	Skin preparation, patient position, placement and fixation, testing connection, reference electrode
3. Sensor location [16,17]	Orientation on muscles
	Consideration of cross talk—provided reference article or stated
4. Detection equipment [16,17]	Filters (type, kind, bandwidth, order)
Rectification method	Full wave, half wave
Sampling	Manufacturer/type of analogue-to digital (A/D) conversion board, sampling frequency, number of bits, input amplitude range
Amplitude processing	Smoothing, average rectified value, root mean square, integrated EMG
5. Gait velocity [18]	Velocity controlled (either self-selected or fixed velocity)

2.3. Quality assessment

As there were no validated assessment checklists available for this type of review (laboratory-based biomechanical studies), the quality assessment procedure was adapted from other sources [15–17]. A two-phase quality assessment was conducted on relevant articles. The first phase included criteria specific to reporting of EMG methodology using five criteria adapted from the recommendations of SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) [16] and the International Society of Electrophysiology and Kinesiology [17].

Two reviewers (GSM and ARB) independently scored the reporting of EMG-related methodological variables (Table 2). The first three criteria related to: (i) surface EMG sensors; (ii) sensor placement; and (iii) sensor location. Criterion (iv) evaluated signal processing and criterion (v) assessed whether the participants' walking and running velocity were controlled during the experiment, as walking velocity is known to influence the amplitude characteristics of the EMG signal [18]. These five criteria were considered important in the review process because each is known to affect the quality of the recorded EMG signal [16,18]. The reviewers (GSM and ARB) met when all studies had been scored, discussed any discrepancies in scoring and a final score was obtained. Articles that scored at least 3/5 qualified for the second part of the methodological quality assessment. This was done using a modified version of the Quality Index [15] – a 27-item checklist for assessing the methodological quality of both randomised and non-randomised studies of health care interventions. The index has demonstrated high internal consistency for non-randomised studies (Kuder Richardson—20 reliability coefficient = 0.88) and good test-retest (r = 0 .88) and inter-rater (r = 0.75) reliability.

Using the Quality Index, only a reduced subset of the original 27 items were determined to be relevant across the following subscales: *Reporting* (Items 1, 2, 3, 4, 5, 6, 7, 10); *External validity* (Items 11, 12); *Internal validity (Bias)* (items: 15, 16, 18, 20); and *Internal validity (confounding)* (Items 21, 22, 23, 24). Items 4, 14, 23 and 24 were only relevant for studies with interventions (i.e. footwear or foot orthoses) and were not applied to studies comparing participant baseline characteristics (i.e. foot posture). Intervention studies were not assessed by questions relating to principal confounders (Item 5) or selection bias (Items 21, 22), as participants in these studies were trialled in all the interventions without a control group. Subsequently, the total maximum score attainable for each of the three categories was: 15 for foot posture; 16 for footwear; and 16 for foot orthoses. To allow comparison of study scores across categories, the summated score for each study was expressed as a percentage. Appendix A includes the scores obtained by each study.

2.4. Data synthesis

The included studies lacked homogeneity in relation to the techniques for classifying foot posture, the types of foot orthoses, footwear and the EMG parameters included in analyses. Accordingly, pooling the data and meta-analysis were not performed. Another issue that restricted quantitative summary of the findings was under-reporting of the mean effect size and error. Where possible, mean differences with 95% confidence intervals were calculated for studies that reported statistically significant findings with the raw mean scores and error, or, mean effect with error for the intervention or group comparison.

3. Results

3.1. Search results

The electronic database search yielded 504 citations included for title and abstract review. A full text review was completed for 75 articles including four publications translated to English from three different languages (German, Dutch and French). This was reduced to 46 once the inclusion/exclusion criteria were taken into account and then 38 once the EMG methodological reporting was evaluated. These 38 articles qualified for detailed review, which included the second quality assessment using the modified version of the Quality Index. A summary of the search results is presented in Fig. 1.

As evident from Fig. 1, the final 38 studies included six that investigated the effect of *foot posture*, 12 that investigated the effect of *foot orthoses* (or a component of an orthosis), and 20 that investigated the effect of *footwear* on lower limb muscle activity during walking or running. Of the 38 studies, 37 used EMG to record muscle activity and one study also utilised muscle function MRI [19]. The Appendix A includes a tabulated summary of the studies included in the final review.

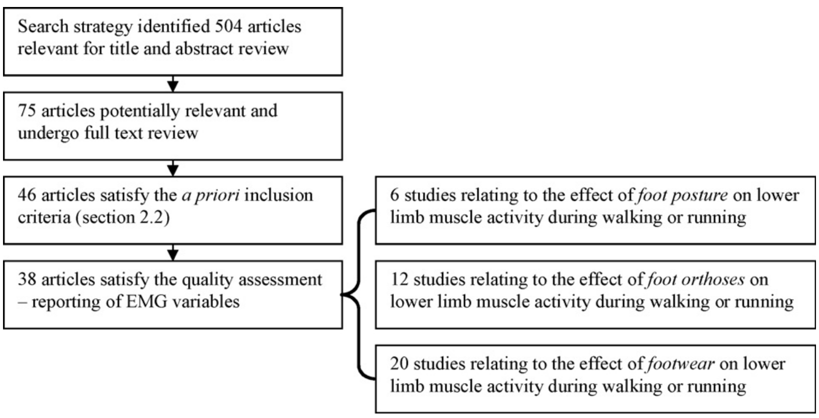

Fig. 1. Search results through the review process.

Table 3
Reporting of inter-rater reliability for EMG methodological criteria

Criteria item with Cohen's kappa and *p*-value				
(1)	(2)	(3)	(4)	(5)
0.351 $p < 0.02$	0.628 $p < 0.001$	0.607 $p < 0.001$	0.612 $p < 0.001$	0.482 $p < 0.01$

Note: Inter-rater agreement for all items = 83%.

3.2. Quality assessment

3.2.1. Phase 1 – Reporting of EMG methodology

The median score for reporting of EMG methodology was 4/5 with 30 articles scoring at least 4/5 and six articles scoring 2/5 or less. Criterion 1, relating to sensor construction, and criterion 3, relating to sensor location, were the most and least frequently reported criteria, respectively. The two reviewers that assessed the reporting of EMG variables demonstrated 83% overall agreement and moderate to good inter-rater reliability (Cohen's kappas ranging from 0.351 to 0.628, $p < 0.02$) (Table 3).

3.2.2. Phase 2 – Methodological quality (modified Quality Index)

The mean score obtained from all studies using the modified Quality Index was 56%. Most studies (25/38) rated 65% or less (50% or less $n = 14$, 51–65% $n = 11$, 66–75% $n = 10$, greater than 75% $n = 2$). The two methodological limitations found across all three categories were the samples were not generalizable (Item 12) and assessors were not blinded (Item 15).

In the *foot posture* category, the main outcome measure was clearly described in the introduction or methods section (Item 2) of only three studies [20–22]. Similarly, only three studies [21,23,24] reported actual probability values (*p*-values) for the main outcomes (Item 10). None of the six studies identified the source of the participants, including whether they were a random sample of a specific population (Item 11) or the time period during which the participants were recruited (Item 22). The internal validity (bias) of the foot posture studies was problematic as only one study [22] indicated the accuracy (validity and reliability – Item 20) of the main EMG outcome measure.

In the *foot orthoses* category, none of the studies blinded participants to the type of foot orthoses (Item 14) or concealed the intervention, when randomly allocated, from both the participants and research staff during data collection (Item 24).

Table 4
Difference in means with 95% confidence intervals for comparisons of conditions

Category	Author/s	Conditions	Muscle (EMG parameter)	%Difference in means (direction of change)	95% CI
Foot posture	Cornwall and McPoil [20]	Late pronators vs early pronators	Tib. anterior (timing)^W	−16.9 (earlier minimum with early pronators)	−35.4 to 1.59
	Williams et al. [21]	High-arch vs low-arch	Vast. lateralis (timing)^R	−7.75 (earlier onset with high-arched feet)	−12.6 to −2.9
Foot orthoses	Nawoczenski and Ludewig [29]	Custom foot orthoses vs control	Tib. anterior (amplitude)^R	37.5 (increased amplitude with custom foot orthoses)	28.3 to 46.7
			Bic. femoris (amplitude)^R	−11.1 (decrease amplitude with custom foot orthoses)	−13.7 to −8.3
	Tomaro and Burdett [30]	Custom foot orthoses vs control	Tib. anterior (duration)^W	2.6 (longer duration with custom foot orthoses)	−3.3 to 8.5
Footwear	Chiu and Wang [26]	Nursing shoe A vs nursing shoe C	Med. gastroc. (amplitude)^W	−13.0 (lower amplitude with shoe A)	−16.1 to −9.0
		Shoe B vs shoe C	Med. gastroc. (amplitude)^W	−16.0 (lower amplitude with shoe B)	−19.2 to −12.8
	Gefen et al. [31]	High-heel shoe vs low-heel shoe wearers	Per. longus (median frequency)^W	−20.0 (faster decrease in high-heel wearers)	−33.9 to −6.1
			Lat. gastroc. (median frequency)^W	29.0 (greater difference in median frequency between med. and lat. gastroc. for high-heel wearers)	18.5 to 39.5
	Li and Hong [32]	Negative-heel vs normal heel	Bic. femoris (amplitude)^W	3.3 (greater amplitude with negative-heeled shoes)	2.5 to 4.1
			Tib. anterior (amplitude)^W	5.1 (greater amplitude with negative-heeled shoes)	−1.0 to 11.3
			Lat. gastroc. (amplitude)^W	7.0 (greater amplitude with negative-heeled shoes)	−0.2 to 14.1
			Lat. gastroc. (duration)^W	8.9 (longer duration with negative-heeled shoes)	1.3 to 16.6
			Tib. anterior (duration)^W	17.5 (longer duration with negative-heeled shoes)	10.0 to 24.9
	O'Connor et al. [19]	Varus midsole vs neutral midsole	Tib. anterior (amplitude)^R	16 (greater amplitude with varus midsole)	−2.2 to 3.4 (Millivolts $\times 10^{-2}$)
		Varus midsole vs neutral midsole	Soleus (amplitude)^R	18 (greater amplitude with varus midsole)	−0.3 to 1.3 (Millivolts $\times 10^{-2}$)
		Valgus midsole vs neutral midsole	Soleus (amplitude)^R	4 (greater amplitude with valgus midsole)	−0.6 to 0.8 (Millivolts $\times 10^{-2}$)
	Serrao and Amadio [33]	Athletic shoe vs barefoot	Vast. lateralis (timing)^W	4.2 (greater time to reach maximum with athletic shoes)	−8.1 to 16.4
			Vast. lateralis (timing)^R	10.2 (greater time to reach maximum with athletic shoes)	−11.4 to 31.8
			Med. gastroc. (timing)^W	18.8 (greater time to reach maximum with athletic shoes)	−25.8 to 63.4

Note: Only studies that published sufficient data are included. *R* = Running; *W* = walking.

In the *footwear* category, less than half of the studies reported actual *p*-values for the main outcomes (Item 10) and only one study indirectly blinded participants to the type of footwear [25]. Four studies [19,26–28] indicated the validity or reliability (Item 20) of the main EMG outcome measure. Again, none of the studies concealed the intervention from both the participants and research staff (Item 24).

3.2.3. Reporting of effect size and error

The mean effect and error were not reported in any format for 7/38 studies. Six studies reported the mean effect without error in either table or graph format, 12 reported the mean and error in table format, 10 reported mean and error in a graph, and only two reported the mean effect with 95% confidence interval in a graph or ensemble EMG trace. The 95% confidence interval was calculated for nine studies (Table 4).

3.3. Overview of included studies

The muscles investigated commonly included lower back (erector spinae), gluteal region (gluteus maximus and minimus), thigh region (biceps femoris, rectus femoris and vastus lateralis) and lower leg (tibialis anterior, gastrocnemii and peronei). A wide range of EMG parameters were analysed across the studies including temporal (onset, duration and time to minimum) and intensity (wavelet analysis, integrated and normalised peak amplitude) related characteristics. Furthermore, the EMG signals were evaluated at a range of different stages of the gait cycle (pre-heel-strike phase, propulsion phase, etc.).

3.3.1. Foot posture

In the studies evaluating foot posture, participant sample sizes varied from 18 to 43. The age range of participants was reported in 5/6 studies, most of which were young adults (i.e. 25–35 years), although in one study participants were aged 40–71 years with moderate to long standing rheumatoid arthritis [24]. The different methods of classifying foot posture included the arch index [23], the arch ratio [21], radiographic alignment [24], two-dimensional video analysis [20] and subjective clinical observation [22,34]. Two studies assessed participants running [21,23] and four studies assessed participants during walking [20,22,24,34].

Williams et al. [21] conducted one of the two running studies and compared 20 flat-arched to 20 high-arched runners. High-arched runners displayed significantly earlier EMG onset for vastus lateralis compared to low-arched runners. The other running study [23] included 15 normal-arched, 12 flat-arched and 16 high-arched participants using a treadmill and found no significant differences in EMG amplitude for vastus lateralis or other lower limb muscles.

All studies that evaluated participants during walking reported significant findings for flat-arched foot posture. These comprised the following comparisons: males with symptomatic pronated foot posture compared to males without symptoms [22]; early pronators compared to late pronators [20]; flat-arched compared to normal-arched [34]; and valgus deformity compared to normal alignment in rheumatoid arthritis [24]. Early pronators displayed a significantly shorter 'time to minimum' EMG amplitude for tibialis anterior [20]. One study [34] reported a "greater level of activity for most lower limb muscles" with no numerical or quantitative data such as group mean or error. Hunt and Smith [22] reported EMG amplitude across five data points with ensemble EMG curves displaying 95% confidence intervals. They found participants with pronated foot posture had *greater* EMG amplitude for tibialis anterior, extensor digitorum brevis, lateral gastrocnemius and soleus in some phases of the gait cycle. In other phases of the gait cycle, the pronated foot also displayed *lower* EMG amplitude for extensor digitorum longus, soleus, medial and lateral gastrocnemius, peroneus longus and brevis. In the study evaluating valgus foot deformity in people with rheumatoid arthritis [24] the participants displayed greater EMG amplitude for tibialis posterior, flexor hallucis longus, flexor digitorum brevis and lower EMG amplitude for peroneus brevis during stance phase compared to a normally aligned group.

3.3.2. Foot orthoses

In the studies evaluating foot orthoses, participant sample sizes ranged from 9 to 40 with most studies reporting an even distribution of males and females and all involved a young adult sample (i.e. 20–35 years). The experimental foot orthoses varied considerably and included the following: customised and prefabricated foot orthoses [29,30,35,36]; external ankle support [37–39]; heel cups [40,41]; textured inserts [42]; medial/lateral wedging; and heel lifts [43,44]. Most studies assessed participants during walking [30,36,37,39–42,44], two studies evaluated participants while running [29,43], and one study included both walking and running conditions [38].

Customised foot orthoses were included in four studies, two with participants walking [30,36] and two during running [29,35]. These studies incorporated orthoses with semi-rigid polypropylene shells [29,30] or with varying levels of medial wedging [35,36]. One study [35] also included foot orthoses manufactured from ethylene vinyl acetate (dense foam) with and without wedging. Across these studies, a multitude of *post hoc* findings were reported, including significant increases in peroneus longus [35,36] and tibialis anterior EMG amplitude [29,35,36], and tibialis anterior duration [30]. Additionally, Mundermann et al. [35] reported a plethora of significant changes across the following variables: global high and low frequency EMG amplitude $\times 7$ muscles $\times 3$ stages of the gait cycle. One of these findings, relating to biceps femoris activity, conflicts with earlier research. Nawoczenski and Ludewig [29] reported biceps femoris EMG amplitude significantly *decreased* by 11.1% with customised foot orthoses during running, whereas Mundermann et al. [35], utilising wavelet analyses, demonstrated biceps femoris global EMG amplitude *increased* significantly with several types of customised foot orthoses.

With respect to external ankle supports, two studies reported a significant decrease in EMG amplitude for medial gastrocnemius, soleus and the peronei with these devices [37,38]. The first study evaluated Aircast® boots compared with barefoot during walking [37]. The second study assessed custom-made external ankle supports compared to no ankle support during both walking and running [38]. Kadel et al. [37] also reported significantly greater soleus and peroneus longus EMG amplitude with the fibreglass casts compared to barefoot walking.

Textured insoles [42] and soft heel cups [40,41] have been studied using EMG wavelet analyses during walking, predominantly to explore the relationship between altered plantar sensory feedback ('input signals') and muscle EMG. One study reported that textured insoles significantly reduced soleus and tibialis anterior amplitude at different periods of the stance phase [42]. Wakeling et al. [40,41] published two studies using data collected from the same 40 participants walking with either soft heel cups or hard sole shoes. The first study [41] revealed a significant increase in global EMG amplitude for biceps femoris, tibialis anterior and medial gastrocnemius with the soft heel cup. The second study [40] reported significantly greater high frequency EMG for tibialis anterior in the period just after heel-strike.

Finally, various configurations of foot wedging have been investigated during walking [44] and running [43] for their effect on lower back [44] and lower leg [43] muscle activity. Only significant changes were reported for erector spinae and gluteus medius EMG onset with different arrangements of heel lifts and lateral wedging under the forefoot [44], although these changes in EMG onset were only small (erector spinae: 4%; gluteus medius: 2%) relative to the length of the gait cycle.

3.3.3. Footwear

In the studies evaluating footwear, participant sample sizes ranged from 3 to 40. Most studies included only young adults with only three studies including participants with a mean age greater than 35 years. The style and features of the experimental footwear varied significantly and included: standard occupational type shoes or boots [26,45]; athletic footwear [19,25,27,28,33,46–48]; variable heel-height [31,32,49–52]; and unstable footwear design [53–55]. Twelve studies involved participants walking [25,26,31,32,49–56], seven studies involved running [19,25,27,28,46–48] and one study [33] included both walking and running conditions.

Athletic footwear with subtle design variations such as alterations in heel counter stiffness [46], midsole density or stiffness [25,27,28,47], midsole wedging [19,48] and participants' own shoes compared to barefoot [33] were investigated almost exclusively during running. One study [33] included a walking and running condition with only three participants assessed in their own running shoes compared to barefoot. The peak EMG amplitude for vastus lateralis and medial gastrocnemius occurred significantly earlier with the participants' own athletic shoes during walking. Another study [46] included 11 asymptomatic heel-strike runners tested in athletic shoes with and without a heel counter. EMG amplitude for triceps surae and quadriceps occurred significantly earlier with the heel counter removed. Two other studies investigated the effect of athletic shoes with variable levels of midsole stiffness on similar muscle groups (i.e. thigh and lower leg), however neither study reported any significant findings [47,52].

Recent advances in EMG wavelet analyses have also been used to quantify total EMG amplitude [27] and high/low frequency bands [28] while comparing various running shoe densities. von Tscharner et al. [27] found significantly greater EMG intensity for tibialis anterior pre-heel-strike and lower intensity post-heel-strike with running shoes compared to barefoot. The mean difference between these conditions was plotted on an ensemble EMG trace with a moving 95% confidence interval, although it was not clear which of the two shoe designs were compared to barefoot. Wakeling et al. [28] indicated that significant changes occurred in the intensity ratio between high and low frequency bands with different shoe materials, although they did not report the *post hoc* findings for any muscle or shoe effects. O'Connor et al. [19,48] also investigated the effect of running shoes, however the shoes incorporated a custom-made midsole aimed at inducing foot pronation and supination during different stages of the gait cycle while running. They utilised EMG to record muscle amplitude and temporal parameters [19,48], and muscle function MRI [19] to assess transverse relaxation time (i.e. a measure of metabolic activity and workload). The shoe with a medial-wedged midsole significantly increased tibialis anterior EMG amplitude compared to the neutral midsole. The neutral midsole significantly decreased the EMG amplitude for soleus compared to the medial and lateral-wedged midsoles.

Specific occupational footwear was evaluated by two studies in the form of nursing shoes [26] and clean room boots (rubber boot with polyurethane or PVC sole) [56]. In the first of these studies [26], 12 nursing staff utilised three nursing shoe styles while at work. The study found that shoes with 'arch support' produced a significant decrease in medial gastrocnemius EMG amplitude. In the second study, clean room boots with variable shock-absorbing and elastic properties were investigated under different walking conditions (e.g. carrying a load) [56]. Gastrocnemius EMG amplitude was significantly lower with heavier, more elastic and shock-absorbing boots when analysed as a function of time (i.e. after 60 min of walking). One other study recruited seven healthy participants and evaluated air-sole running shoes, air-cushioned street shoes and leather-sole street shoes [45]. This study found no significant changes in EMG amplitude or duration between the shoes.

The effect of variations in heel height on muscle activity in female participants was investigated by four studies, with heel heights ranging from 0 to 8 cm [49–52] and one additional study evaluated negative-heeled shoes [32]. With increasing heel height, the following changes were noted: greater peak EMG amplitude for erector spinae [50], decreased medial gastrocnemius and tibialis anterior peak EMG amplitude [51] and increased rectus femoris, soleus and peroneus longus root mean square (RMS) EMG amplitude [52]. One study reported no significant changes at all [49]. Lee et al. [50] included a 95% confidence interval (in a bar graph) for erector spinae, which illustrated consistent increases in peak EMG with increasing heel height (i.e. a systematic effect). Gefen et al. [31] included four habitual wearers of high-heeled and four habitual wearers of flat-heeled shoes and compared the medium EMG frequency of lower limb muscles during barefoot walking. Habitual high-heel wearers displayed a significantly faster decrease in median frequency for peroneus longus and lateral gastrocnemius after completing a fatiguing exercise compared to habitual low-heel wearers. In contrast to high-heeled shoes, Li and Hong [32] compared negative-heeled shoes to normal heeled shoes. They found the negative-heeled shoes caused significantly greater EMG amplitude for biceps femoris, tibialis anterior and lateral gastrocnemius, and longer EMG duration for lateral gastrocnemius and tibialis anterior.

Two other types of footwear that have received attention include unstable footwear and ankle destabilisation shoes. One study [53] utilised a mechanical destabilisation device under the heel of a shoe while walking to induce destabilisation of the rearfoot in nine healthy participants. A significant increase in tibialis anterior (7.8%), peroneus longus (6.0%) and peroneus brevis (2.1%) EMG amplitude was reported with the destabilisation shoe compared to barefoot. Two other studies compared the effect of unstable shoe designs (that incorporate a rounded sole in the anterior–posterior direction) to either participants' own shoes [54] or running shoes [55]. Romkes et al. [55] reported that the unstable shoe significantly altered rectus femoris, vastus medialis, vastus lateralis, tibialis anterior, medial and lateral gastrocnemius root mean square EMG amplitude within defined intervals of stance and swing phase, although they did not report any form of error measurement for the EMG variables. Nigg et al. [54] also recorded EMG from gluteus medius, biceps femoris, vastus medialis, medial gastrocnemius and tibialis anterior and found no significant changes in total EMG amplitude using wavelet analysis.

4. Discussion

Our review identified 38 articles, which evaluated either the effect of foot posture, foot orthoses, or footwear on lower limb muscle EMG or MRI during walking or running. In addition to the methodological issues (identified in the quality assessment) amongst the studies, there were also deficiencies in reporting effect size and clinical and statistical heterogeneity. This affected our ability to pool data and draw definitive conclusions from studies within each category.

4.1. Quality assessment and effect size

As 25/38 studies rated 65% or less, the majority of articles were of low to moderate methodological quality, especially in the category of external validity. Most studies included small sample sizes (i.e. less than 15) with inadequate reporting of mean effect size and confidence intervals. Despite the small sample sizes and under-reporting of measurement error or variance, many studies, perhaps incorrectly, proceeded to apply parametric statistical analyses to evaluate hypotheses. Such small sample sizes may have also led to statistical power issues, potentially leading to type II error. The issue surrounding statistical power in laboratory-based studies would be improved through *a priori* sample size estimation.

The lack of reporting of mean effect sizes with confidence intervals is a key deficit amongst the studies reviewed, although we acknowledge that such reporting has only recently become the accepted gold-standard. Most authors only presented probability values with mean effect and standard errors. The mean effect sizes and confidence intervals were calculated for nine of the included studies, all of which reported statistical significance for their findings (i.e. p-value less than 0.05). However, when considering the confidence intervals (Table 4), 10/20 *post hoc* comparisons in these studies had a confidence interval that included a zero value (i.e. the lower and upper confidence limits were less than and more than zero, respectively, indicating a non-significant finding). A further 4/20 comparisons had an effect size smaller than 5%, suggesting substantial uncertainty about whether these effects are clinically meaningful. Finally, due to the style of statistical reporting, it was difficult to identify results that reflected systematic and non-systematic effects across subjects in these studies.

4.2. Clinical and statistical heterogeneity

Our review found significant clinical heterogeneity existed between studies within each category. For example, when considering the category of foot orthoses, we directly compared the effect of customised foot orthoses manufactured from polypropylene to other studies that evaluated softer materials such as ethylene vinyl acetate (EVA). A further example is that of the foot posture category, where a wide range of techniques for classifying participants foot posture were used. One study [24] assessed foot posture with X-rays by measuring radiological alignment of the foot. In contrast, another study [22] based their inclusion on physiotherapists' clinical observations of foot posture.

In addition, due to significant differences in analyses such as the range of EMG parameters and data processing techniques, it was not feasible to compare quantitative aspects of the studies. For example, it is not possible to directly compare EMG data presented in the form of wavelet analysis to normalised EMG amplitude data. Accordingly, the data could not be pooled for meta-analysis, making it difficult to draw sound conclusions about the effect of foot posture, foot orthoses and footwear.

Only a limited number of studies clearly stated the rationale for including specific EMG variables, such as wavelet analysis [27,28,35,40–42,54] and median frequency [31]. When the range of EMG parameters from articles in our review are considered, it is clear that there is a need for a universal set of standards and recommendations outlining the most valid and reliable EMG parameters in gait research.

4.3. Foot posture studies

The relationship between foot posture and lower limb muscle activity is unclear. There was some evidence that pronated foot posture was associated with greater EMG amplitude for invertor muscles such as tibialis posterior [22,34], tibialis anterior [24] and flexor hallucis longus [24,34] when compared to normal or supinated foot posture. Conversely, pronated feet are associated with lower EMG amplitude for evertor muscles such as peroneus longus [22,24] compared to normal or supinated foot posture.

A major limitation of studies investigating the relationship between foot posture and lower limb muscle function is that at present, there is no universally accepted method for classifying foot posture that is both highly predictive of dynamic skeletal motion and associated with an increased risk of musculoskeletal injury [57]. It is therefore unclear whether the methods of classifying foot posture adopted by the reviewed studies are appropriate. It could be hypothesised that a dynamic method of classifying foot posture is required to evaluate whether abnormal foot posture is related to altered muscle activity.

4.4. Foot orthoses studies

The category of foot orthoses drew similar conclusions to the category of foot posture. Irrespective of the foot orthosis material, there is some evidence that peroneus longus and tibialis anterior EMG amplitude, and tibialis anterior duration is greater when wearing foot orthoses. These changes occurred in comparison to standard shoes alone during walking and/or sandals during running [29,30,35,36]. Other components of foot orthoses (i.e. those using hindfoot and forefoot wedging), textured insoles, heel cups and ankle bracing have also been reported to significantly affect lower limb or lower back EMG muscle function [37,38,40–42,44,58].

It is unclear, however, whether changes in muscle function using foot orthoses are consistent and predictable, even when the participants have similar foot posture [30,35,36]. Moreover, it is currently not known whether an increase or decrease in many of the measured EMG variables is beneficial or detrimental in relation to injury. While it makes intuitive sense that an intervention would be beneficial if it can bring muscle activity closer to that seen in a non-pathological population (measured via EMG), definitive evidence is still lacking. Accordingly, it is difficult to make conclusions about the effect of altered muscle function on clinically relevant conditions (e.g. tibialis posterior tendon dysfunction) [59].

4.5. Footwear studies

Numerous styles of footwear were included in the review, with the most commonly studied being shoes with varying heel height. Four of the five studies demonstrated significant changes in either lower back [50] or lower limb [32,51,52] EMG muscle activity with increasing heel height. Additionally, Gefen et al. [31] reported that peroneus longus and lateral gastrocnemius are more fatigable in habitual wearers of high-heeled shoes. Therefore, there is some evidence that extreme variations in heel height significantly affect the amplitude of lower back and fatigability of lower limb EMG muscle activity during walking.

Destabilisation [53] and unstable [54,55] footwear are designed to "enhance ankle stabilising musculature" [55] as part of injury treatment and prevention, however the actual effect of these shoes is far from clear. The two studies [54,55] that investigated the unstable shoe design provide conflicting findings regarding the effect of this footwear design on thigh and lower leg EMG activity during walking, although it should be noted that these studies utilised a different control condition. Therefore, we found no evidence to suggest this type of footwear has a systematic effect. Clearly, there is a need for further research on the efficacy of unstable shoe designs and destabilisation devices to determine whether they produce predictable and consistent changes in muscle activity.

A further eight studies investigated variation in athletic footwear design during running. The earliest and most recently published studies were from 1986 [49] and 2007 [32], respectively. Over this time, significant advances in muscle function analysis techniques such as wavelet analysis and muscle function MRI have occurred, which precludes the pooling of data extracted from earlier studies with similar methodology. Accordingly, no conclusions can be made with respect to the effect of athletic footwear on muscle function. As these newer techniques emerge and become more broadly accepted in the literature, there will be a need for greater consensus in reporting of important EMG parameters.

4.6. Limitations

The reporting of EMG variables – based on the recommendations of SENIAM [16] – were incorporated into this review to enable a basic quality assessment relating to EMG methodology. Due to the large number of studies, it was not feasible to contact the authors of articles who neglected to list in detail the EMG methodology. Accordingly, some of the excluded studies may have incorporated sound EMG methodological processes, but were excluded because they did not report these processes. Further, studies were not evaluated for the quality of each criteria (e.g. filtering process), thus reporting a criterion did not necessarily ensure that the methods used were of an acceptable standard. While only moderate inter-rater reliability was obtained between the two raters for scoring the reporting of EMG variables, any discrepancies in scoring were always followed up with a process of discussion, review of the relevant criteria, and finally, consensus re-rating. The Quality Index appeared to be the most relevant checklist available to assess the methodological quality of laboratory-based EMG studies. However, several items were identified as being irrelevant and were omitted from specific categories. This may have affected the overall validity of the checklist for conducting this type of review.

5. Conclusion

Lower limb muscle EMG is affected by some variations in foot posture, foot orthoses and footwear. Some evidence exists that: (i) pronated feet demonstrate greater activation of invertor musculature and decreased activation of evertor musculature; (ii) foot orthoses increase activation of tibialis anterior and peroneus longus, and may alter lower back muscle activation; and (iii) shoes with elevated heels alter lower limb and back muscle activation. However, there were substantial limitations in the data presented in the majority of studies reviewed. On the whole, the studies were of only moderate methodological quality with significant deficiencies in basic reporting of effect size and error. Additional issues that limit the conclusions that could be made from this review relate to clinical and statistical heterogeneity, which prevented pooling of data from similarly designed studies. There is, therefore, a need for greater consensus regarding standards for conducting and reporting of EMG studies.

Conflict of interest

The authors state that there are no conflicts of interest, which might have influenced the preparation of this manuscript.

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Appendix A

Posterior trunk: Er. spinae – erector spinae. *Anterior trunk*: Rec.abdom. – rectus abdominus. *Gluteal region*: Glut.maximus – gluteus maximus, Glut.medius – gluteus medius. *Posterior compartment thigh*: Sem.tend. – semitendinosis, Sem.memb. – semitendinosis, Bic.femoris – biceps femoris, Lat.hamst. – lateral hamstring. *Anterior compartment thigh*: Rec.femoris – rectus femoris, Vast.lateralis – vastus lateralis, Vast.medialis – vastus medialis. *Posterior compartment leg*: Gastroc. – gastrocnemius, Med.gastroc. – medial gastrocnemius, Lat.gastroc. – lateral gastrocnemius, Flx.d.longus – flexor digitorum longus, Tib.posterior – tibialis posterior, Flx.h.longus – flexor hallucis longus. *Anterior compartment leg*: Tib.anterior – tibialis anterior, Ext.h.longus – extensor hallucis longus, Ext.d.longus – extensor digitorum longus. *Lateral compartment leg*: Per.longus – peroneus longus, Per.brevis – peroneus brevis. *Foot*: Abd.hallucis – abductor hallucis, Flex.d.brevis – flexor digitorum brevis.

Summary of articles related to the effect of *FOOT POSTURE* on lower limb muscle activity during walking and running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Foot posture classification	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Backmann [23]	Whole sample data 16 female, 27 male 24.1 years (±4.11), 72.3 cm (±9.7), 72.2 kg (±0.1)	Static and dynamic Arch index –flat arch (n = 15) –normal arch (n = 12) –high arch (n = 16)	Semi.tend. Semi.memb. Vast.lateralis Gastroc.# Tib.anterior	Running	–Integrated EMG	53%	• No significant differences between groups
Cornwall and McPoil [20]	Individual sample data 1. *Early pronators* (6 females, 4 males) 25.7 years (±3.3), 174.3 cm (±10.1), 79.4 kg (±15.3) 2. *Late pronators* (6 females, 2 females) 29.3 years (±6.12), 169.8 cm (±10.4), 71.8 kg (±17.5)	2D video analysis 1. Early pronators reached maximum pronation within first 20% stance phase 2. Late pronators reached maximum pronation after 40% stance phase	Tib.anterior	Walking	–Time to minimum EMG	60%	*Early pronators* • Significantly 'shorter time to minimum' EMG for tib.anterior
Gray and Basmajian [34]	Whole sample data 1 female, 19 male[†]	Visual observation –normal arch (n = 10) –lat arch (n = 10)	Tib.posterior Flx.h.longus Tib.anterior Per.longus Abd.hallucis Flx.d.brevis	Walking	–Onset and offset –Level of activity (*nil, slight, moderate* and *marked*)	20%	*Flat-arched* subjects • Significantly greater EMG activity 'early instance for most muscles'[†]
Hunt and Smith [22]	Individual sample data 1. *Normal males* (n = 18) 25 years (±5), 1.78 cm (±.07), 78.3 kg (±10.8) 2. *Males with pronated static foot posture* (n = 15) 26 years (±7), 1.76 cm (±.07), 77.5 kg (±13.2)	Clinical assessment 1. Males without symptoms, foot orthoses or malalignment 2. Males with musculoskeletal symptoms attributed to their pronated static foot posture	Med.gastroc. Lat.gastroc. Soleus Tib.anterior Per.longus Per.brevis Ext.d.longus	Walking	–Normalised amplitude of five data points –Number of EMG peaks and troughs	60%	*Pronated foot posture* • Significantly greater EMG activity at heel contact for tib.anterior, at 40% stance for ext.d.longus, 80% for lat.gastoc., and soleus • Significantly lower EMG activity at 5% for per.longus, per.brevis, ext.d.longus at 10% for soleus and lat.gastroc., at foot-flat for med.gastroc.
Keenan et al. [24]	Individual sample data 1. *Normal hind-foot alignment* Age (range): 63 yers (40–71) (5 female, 2 male) 2. *Planovalgus foot alignment* Age (range): 60 years (44–76) (8 female, 2 male)	Plain film radiographs 1. Normal alignment –dorsoplantar view –lateral view 2. Valgus alignment –dorsoplantar view –lateral view	Gastroc.# Soleus Tib.posterior Flx.h.longus Tib.anterior Flx.d.brevis Per.brevis Per.longus	Walking	–Normalised amplitude calculated every 0.02 s –Duration	40%	*Valgus group* • Significantly greater EMG activity for tib.posterior, flx.h.longus and flx.d.longus • Significantly lower EMG activity for per.brevis

Appendix A (*Continued*)

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Foot posture classification	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Williams et al. [21]	Individual sample data 1. *High arch* (12 females 12 males) 28 years (±8.1), 66.5 kg (± 9.5), 1.72 (±0.1) 2. *Low arch* 27.7 years (±7.5), 72.7 kg (± 17.9), 1.74 (±0.1) (10 female, 8 male)	Arch ratio	Vast.lateralis Lat.hamst. Tib.anterior Lat.gastroc. Med.gastroc.	Running	–Onset –Amount of coactivation	73%	*High-arched* runners • Significantly earlier EMG onset for vast.lateralis

#Muscle unspecified.
†No further information available.
‡Several *post hoc* findings.
**Quality score derived from 15 items as no interventions were tested.

Summary of articles related to the effect of *FOOT ORTHOSES* on lower limb muscle activity during walking and running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Foot orthoses/test conditions	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Baur et al. [43]	17 males (running >50 km/week) 31 years (±8), 178 cm (±8), 73 kg (±17)	(1) Barefoot (2) Reference (running) shoe (3) Insoles without functional elements (EVA) (4), (5), (6), (7), (8) and (9) comprised components of EVA insoles with either/or part; cuboid notch, lateral forefoot wedge, 'shell form' or medial wedge	Med.gastroc. Lat.gastroc. Soleus Tib.anterior Per.longus	Running	–Onset, offset and duration –EMG amplitude	69%	• No significant differences between conditions
Bird et al. [44]	13 'right handed' participants (7 female, 6 male) 22.3 years (±3.4), 173.1 cm (±7.3), 71.6 kg (±7.8)	(1) 5° Lateral forefoot wedging (2) 5° Medial forefoot wedging (3) 2 cm Heel lift (4) Barefoot [1,2,3 were trialled as left foot, right foot and both feet separately]	Er.spinae Glut.medius	Walking	–EMG onset –Maximum normalised EMG	75%	• Er.spinae – significantly earlier EMG onset with *(bilateral) heel lifts and lateral forefoot wedging* compared to *barefoot* • *Glut.medius – significantly delayed EMG onset with unilateral and bilateral heel lifts compared to barefoot*
Kadel et al. [37]	12 adults without ankle pathology (4 female, 8 male)†	(1) Fibreglass 'cast' (2) Aircast walking 'boot' (3) Barefoot	Med.gastroc. Soleus Peroneals#	Walking	–Integrated and normalised EMG amplitude	44%	• Med.gastroc., soleus, per.longus – significantly lower EMG activity with *boot* compared to barefoot • Soleus and per.longus – significantly greater EMG activity with *cast* compared to barefoot • Med.gastroc. – significantly lower EMG activity with *boot* compared to *cast*
Kondradson and Højsgaard [38]	9 experienced runners (4 women, 5 men)	(1) Custom-made external ankle support	Per.longus	Walking and running	–EMG signal assessed at 5% intervals (up to 100%) for 'presence' of activity	38%	• Peroneals# – significantly lower pre-stance phase EMG activity with *external ankle support* compared to '*without ankle support*' during walking and medium-pace running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Foot orthoses/test conditions	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
	Age (range): 25 years (21–34)	(2) Without external ankle support†	Per.brevis				
Mundermann et al. [35]	21 volunteers – 'pronators' (12 female, 9 male) 25.4 years (±5.6), 170.2 cm (±6.7), 64.2 kg (±5.6)	(1) Control insert (2) Posting orthoses (3) Molding orthoses (4) Posting and molding orthoses (1, 2 and 3 were customised foot orthoses)	Bic.femoris Rec.femoris Vast.lateralis Vast.medialis Med.gastoc. Tib.anterior Per.longus	Running	–Wavelet analysis (high, low and global frequency bands during pre-heel-strike, post-heel-strike and propulsive phase)	75%	• Tib.anterior, vast.lateralis, vast.medialis and rec.femoris – significant changes‡ in global EMG intensity during pre-heel-strike with *orthoses*‡ compared to *control insert* • Tib.anterior, per.longus, med.gastroc., bic.femoris – significant changes‡ in global EMG intensity during post-heel-strike with *orthoses*‡ compared to *control Insert* • *Tib.anterior, per.longus – significant changes‡ in global EMG intensity during post-heel-strike with orthoses‡ compared to control insert*

#Muscle unspecified.
†No further information available.
‡Several *post hoc* findings.
**Quality score derived from 15 items as no interventions were tested.

Summary of articles related to the effect of *FOOT ORTHOSES* on lower limb muscle activity during walking and running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Foot orthoses/test conditions	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Murley and Bird [36]	17 Asymptomatic participants (10 females, 7 males) 23 years (±5), 170.2 cm (±9.65), 69.9 kg (±14.4)	(1) Barefoot (2) Shoe only (3) 0° inverted orthoses (4) 15° inverted orthoses (5) 30° inverted orthoses (3, 4 and 5 were customised foot orthoses)	Med.gastroc. Soleus Tib.anterior Per.longus	Walking	–Normalised amplitude –Onset	81%	• Tib.anterior – significant increase in EMG amplitude with *shoe only 0°, 15° and 30° inverted orthoses* compared to *barefoot* • Per.longus-significant increase in EMG amplitude with *15° inverted orthoses* compared to *shoe only*
Nawoczenski and Ludewig [29]	12 recreational runners (6 female, 6 male) 27.2 years (±9.9), 1172 cm (±0.1), 65.4 kg (±12.7)	(1) Customised foot orthoses (2) Running sandal	Bic.femoris Vast.lateralis Vast.medialis Med.gastroc. Tib.anterior	Running	–Mean 'root mean squared' (RMS) amplitude for first 50% of stance phase	69%	• Tib.anterior – significantly greater RMS EMG amplitude with *orthoses* compared to *sandal* • Bic.femoris – significantly lower RMS EMG amplitude with *orthoses* compared to *sandal*
Nurse et al. [42]	15 adults without ankle pathology (3 female, 12 male) 24.7 years (±2.9), 177 cm (±9), 74 kg (±12)	(1) Control insert (2) Textured insert	Bic.femoris Rect.femoris Vast.medialis Med.gastroc. Soleus Tib.anterior	Walking	–Wavelet analysis (total EMG intensity)	63%	• Tib.anterior and soleus – significantly lower global EMG intensity with *textured inserts* compared to *control insert*

Appendix A (*Continued*)

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Foot orthoses/test conditions	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Scheuffelen et al. [58]	12 healthy subjects with stable ankles†	1) 3× different stabilising shoes (2) 4× different ankle braces (3) Tape type 'fast-gips' (4) Jogging shoe	Med.gastroc. Tib.anterior Per.longus	Walking	–Integrated EMG and normalised amplitude	31%	• No significant differences between conditions
Tomaro and Burdett [30]	10 volunteers (7 female, 3 male) Age range: 25–30 years†	(1) Customised foot orthoses (2) Non-orthoses (athletic shoes)	Lat.gastroc. Tib.anterior Per.longus	Walking	–Duration of tib.anterior EMG activity –Mean RMS divided by duration of stance phase	56%	• Tib.anterior – significantly greater duration of EMG activity with foot orthoses compared to *non-orthoses*
Wakeling et al. [41]	40 subjects 20 female 25.8 years (±1.1), 67.6 kg (±1.6) 20 male 26.9 years (±1.0), 78.2 kg (±3.2)	(1) Control – hard sole shoe (2) Soft heel cup insert	Bic.femoris Rec.femoris Lat.gastroc. Tib.anterior	Walking	–Wavelet analysis (total EMG intensity)	63%	• Tib.anterior, bic.femoris, med.gastroc., – significantly greater global EMG intensity with *soft heel cup insert* compared to *control*
Wakeling and Liphardt [40]	40 subjects (20 female, 20 male) Female: 25.8 years (±1.1), 67.6 kg (±1.6) Male: 26.9 years (±1.0), 78.2 kg (±3.2)	(1) Control – hard sole shoe (2) Soft heel cup insert	Lat.gastroc. Tib.anterior	Walking	–Wavelet analysis (high frequency component)	69%	• Tib.anterior – significantly greater high-frequency EMG activity 0–60 ms after heel-strike with *soft heel cup insert* compared to control

#Muscle unspecified.
†No further information available.
‡Several *post hoc* findings.
**Quality score derived from 15 items as no interventions were tested.

Summary of articles related to the effect of *FOOTWEAR* on lower limb muscle activity during walking and running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Footwear/test conditions	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Chiu and Wang [26]	12 'healthy' females 23.3 years (±2.1), 158.4 cm (±4), 47.7 kg (±5)	(1) Nursing shoe A (2) Nursing shoe B (3) Nursing shoe C (A, B, and C had different sole, misole, upper and innersole characteristics)	Bic.femoris Rec.femoris Med.gastroc. Tib.anterior	Walking	–Mean normalised EMG amplitude	44%	• Med.gastroc. – significantly lower EMG amplitude with *shoe A and shoe B* compared to *shoe C (shoe A* and B included an arch support design†)
Forestier and Toschi [53]	9 healthy subjects 37 years (±12.0), 173 cm (±7), 68 kg (±17)	(1) Barefoot (2) Ankle destabilisation shoe	Med.gastroc. Lat.gastroc. Tib.anterior Per.longus Per.brevis	Walking	–Integrated and normalised EMG amplitude –Onset time (per. longus and brevis only)	50%	• Tib.anterior, per.brevis, per.longus-significantly greater EMG amplitude with *destabilisation shoe* compared to barefoot
Gefen et al. [31]	8 female subjects 26 years (±4), 55 kg (±5)† 4 habitual wearers of *high-heeled* shoes 4 habitual wearers of *flat-heeled* shoes	(1) Barefoot (2) 'Own' footwear† (3) Fatiguing exercise 1 (4) Fatiguing exercise 2	Med.gastroc. Lat.gastroc. Soleus Per.longus Tib.anterior Ext.h.longus	Walking	–Normalised EMG median frequency	69%**	• Med. and lat.gastroc. – significantly faster decrease in median EMG frequency for lat.gastroc., relative to med.gastroc., in habitual *high-heeled* wearers (after fatiguing exercises) compared to low-heeled wearers • Per.longus – significantly faster decrease in median EMG frequency for habitual *high-heeled* wearers

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Footwear/test conditions	Muscles (see key)	Walking or running	EMG variables	Quality score	Main findings
Jørgensen [46]	11 symptom-free heel-strike runners (5 female, 6 male) Age (range): 25.5 years (14–37)†	(1) Barefoot (2) Athletic shoe with rigid heel counter (3) Athletic shoe with heel counter removed	Hamstrings# Quadriceps# Triceps surae# Tib.anterior	Running	–Normalised EMG amplitude –Time to peak amplitude –No. of turns in EMG signal	63%	• Triceps surae# and quadriceps# – significantly earlier activity, greater amplitude and *no. of turns* with *heel counter removed* compared to shoes with *rigid heel counter*
Joseph [49]	6 subjects† Height range: 150–175 cm† Mass range: 50–67 kg†	(1) High-heeled shoes (heel height 1–2.5 cm) (2) Low-heeled shoes (heel height 5.5–8 cm)	Er.spinae Glut.maximus Glut.medius Bic.femoris Hip flexor* Soleus Tib.anterior	Walking	–Duration of EMG activity –Raw EMG amplitude	19%	• No significant differences between conditions
Komi et al. [47]	4 males with 'athletic background' 32 years (±9.4), 173.8 cm (±3.8), 72 kg (±5.9)	(1) Barefoot (2), (3) and (4) 'jogging shoes' (5) and (6) 'indoor shoes' (Indoor shoes comprised harder sole characteristics)	Rec.femoris Vast.medialis Lat.gastroc. Tib.anterior	Running	–Mean and integrated EMG	25%	• No significant differences between conditions
Li and Hong [32]	13 female subjects 23.1 years (±3.9), 50.2 kg (±5.3), 163 cm (±0.1)	(1) Normal shoes (2) Negative-heeled shoes	Er.spinae Rec.abdom. Bic.femoris Rec.femoris Lat.gastroc. Tib.anterior	Walking	–Mean and integrated EMG –Duration of EMG	56%	• Bic.femoris, lat.gastroc., tib.anterior – significantly greater EMG amplitude; lat.gastroc., tib.anterior – significantly longer duration with *negative-heeled* shoe compared to *normal* shoe

#Muscle unspecified.
†No further information available.
‡Several *post hoc* findings.
**Quality score derived from 15 items as no interventions were tested.

Summary of articles related to the effect *of FOOTWEAR* on lower limb muscle activity during walking and running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Footwear/test conditions	Muscles (see key)	Walking or running	EMG or MRI variables	Quality score	Main findings
Lin et al. [56]	12 healthy female students 24.2 years (±1.9), 52.0 kg (±5.8), 163 cm (±5.8)	(1) 'Clean room' *boot A* (2) 'Clean room' *boot B* (3) 'clean room' *boot C* Each with different shock-absorbing and elastic properties	Er.spinae Bic.femoris Rec.femoris Gastroc.# Tib.anterior	Walking	–Mean normalised EMG	50%	• Gastrocnemius# – significantly greater EMG amplitude with boot C and B compared to boot A as a function of time
Lee et al. [50]	5 healthy young women ('in their twenties')†	(1) *Low-heeled* shoes (0 cm) (2) *Medium-heeled* shoes (4.5 cm) (3) *High-heeled* shoes (8 cm)	Er.spinae (L_1/L_2) Er.spinae (L_4/L_5) Vast.lateralis Tib.anterior	Walking	–Peak and integrated EMG	50%	• Er.spinae (L_4/L_5) –significantly greater peak EMG as heel height† increased

Appendix A (*Continued*)

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Footwear/test conditions	Muscles (see key)	Walking or running	EMG or MRI variables	Quality score	Main findings
Lee et al. [51]	6 women ('regular wearers of high-heeled shoes') Age range: 20–31 years† Mean height (range): 160 cm (155–168)† Mean mass (range: 54.6 kg (48.1–63.5)†	(1) *Barefoot* (2) *2.5 cm* heeled shoes (3) *5.0 cm* heeled shoes (4) *7.5 cm* heeled shoes	Tib.anterior Med.gastroc.	Walking	–Normalised peak and mean peak EMG	50%	• Med.gastroc. tib.anterior – significantly lower mean peak EMG with 2.5 cm, 5.0 cm and 7.5 cm heeled shoes compared to barefoot • Med.gastroc., – significantly lower mean peak EMG with *2.5 cm and 5 cm* compared to *5.0 cm and 7.5 cm* heeled shoes, respectively • Tib.anterior –significantly greater mean peak EMG with *2.5 cm* compared to both *5.0 cm* and *7.5 cm*
Nigg et al. [54]	8 healthy subjects (3 female, 5 male) 28.0 years (±3.6), 169.5 cm (±6.4), 70.1 kg (±7.5)	(1) *Unstable shoe* (2) *Control shoe*	Glut.medius Bic.femoris Vast.medialis Med.gastroc. Tib.anterior	Walking	–Wavelet analysis (total EMG intensity)	69%	• No significant differences between conditions
O'Connor and Hamill [48]	10 males ('rearfoot strikers') 27 years (±5), 1.72 cm (±0.1), 72.6 kg (±5.3)	(1) Running shoe – *neutral* (2) Running shoe – *medial wedge* (3) Running shoe – *lateral wedge* (EVA rearfoot wedge tapered by 1 cm across heel of midsole + no heel counter on shoes)	Med.gastroc. Lat.gastroc. Soleus Tib.posterior Tib.anterior Per.longus	Running	–Integrated and mean EMG –EMG Onset and offset	56%	• No significant differences between conditions
O'Connor et al. [19]	10 males ('rearfoot strikers') 27 years (±5), 1.72 cm(±0.1), 72.6 kg (±5.3)	(1) Running shoe – *neutral* (2) running shoe – *medial wedge* (3) Running shoe – *lateral wedge* (EVA rearfoot wedge tapered by 1 cm across heel of midsole + no heel counter on shoes)	Med.gastroc. Lat.gastroc. Soleus Tib.posterior Tib.anterior Ext.d.longus Per.longus	Running	–MRI transverse relaxation times (T2). Average of 5 slices through muscle belly (scan occurred 3.6 ± 0.3 min after run completed) –Mean normalised EMG –EMG onset and offset	81%	• Tib.anterior – significantly greater mean amplitude with *medial wedge sole* compared to *neutral* sole • Soleus – significantly lower mean amplitude with *neutral wedged shoe* compared to *lateral* and *medial-wedged sole*

#Muscle unspecified.
†No further information available.
‡Several *post hoc* findings.
**Quality score derived from 15 items as no interventions were tested.

Summary of articles related to the effect of *FOOTWEAR* on lower limb muscle activity during walking and running

Author/s (date)	Participant characteristics (age, height, mass (±standard deviation))	Footwear/test conditions	Muscles (see key)	Walking or running	EMG or MRI variables	Quality score	Main findings
Romkes et al. [55]	12 healthy subjects (6 female, 6 male) 38.6 years (±13.2), 173.3 cm (±6.3), 77.4 kg (±12.3)	(1) Individuals' regular shoes (2) Masai barefoot technologies® (MTB-shoes)	Sem.tend. Rec.femoris Vast.lateralis Vast.medialis Med.gastroc. Lat.gastroc. Tib.anterior	Walking	–RMS from 16 equal intervals over gait cycle (normalised from barefoot condition)	50%	• Tib.anterior, med.gastroc., lat.gastroc., vast.lateralis, vast.medialis, rec.femoris – significantly greater RMS EMG activity with *MTB-shoes* compared to *regular shoes* (during part of contact phase) • Tib.anterior, med.gastroc., lat.gastroc. – significantly greater and rec.femoris – significantly lower RMS EMG with *MTB-shoes* compared to *regular shoes* (in part of swing phase)
Rosenbaum and Hennig [45]	7 healthy subjects (1 female, 6 male) 30.1 years (±3.0), 180.1 cm (±4.3), 71.4 kg (±6.3)	(1) Air-sole running shoe (2) Air cushion street shoe (3) Leather-sole street shoe	Med.gastroc. Soleus Tib.anterior	Walking	–Integrated EMG –EMG duration	50%	• No significant differences between conditions
Roy and Stefanyshyn [25]	13 subjects with weekly mileage >25 km week^{-1} 27 years (±5.1), 177.1 cm (±4.4), 73.2 kg (±5.4)	(1) Unmodified control shoe (2) Modified stiff shoe (3) Modified stiffest shoe	Vast.lateralis Rec.femoris Bic.femoris Med.gastroc. Soleus	Running	–RMS EMG amplitude from four intervals of stance phase	69%	• No significant differences between conditions
Serrao and Amadio [33]	3 'runners' 24.7 years (±3.2), 172 cm (±10), 73.7 kg (±10.1)	(1) Barefoot (2) Individuals' running shoes	Vast.lateralis Med.gastroc.	Walking and Running	–Normalised mean EMG	38%	• Vast.lateralis – significantly delayed peak EMG with *running shoes* compared to *barefoot* (during walking and running) • Med.gastroc. – significantly delayed peak EMG with *running shoes* compared to *barefoot* (during walking only)
Stefanyshyn et al. [52]	13 female subjects 40.6 years (±8.3), 164.1 cm (±5.6), 67.7 kg (±12.3)	(1) Flat shoe (1.4 cm heel height) (2) Low-heeled shoe (3.7 cm) (3) Medium-heeled shoe (5.4 cm) (4) High-heeled shoe (8.5 cm	Sem.tend. Bic.femoris Rec.femoris Vast.medialis Gastroc# Soleus Tib.anterior Per.longus	Walking	–RMS EMG amplitude	56%	• Per.longus, rec.femoris, soleus – significantly greater RMS EMG with *higher-heeled shoes* compared to *lower-heeled shoes*‡
von Tscharner et al. [27]	40 male 'runners' Weekly mileage >25 km wk$^{-1\dagger}$	(1) Barefoot (2) Neutral running shoe (3) Pronation control running shoe	Tib.anterior	Running	–Wavelet analysis (total EMG intensity)	63%	• Tib.anterior – significantly higher and significantly lower EMG intensity with *running shoes*† compared to *barefoot* during pre and post-heel-strike periods of gait cycle
Wakeling et al. [28]	6 'runners' (3 female) 23.3 years (±4.1), 165.7 cm (±1.5), 55.1 kg (±1.8) (3 male) 26.0 years (±2.5), 174 cm (±2.6), 73.4 kg (±1.4)	(1) Hard midsole running shoe (2) Soft midsole running shoe	Bic.femoris Rec.femoris Med.gastroc. Tib.anterior	Running	–Wavelet analysis (low and high frequency bands, pre-heel-strike)	69%	• Muscles# – significantly altered† total EMG intensity with different midsole† hardness pre-heel-strike

#Muscle unspecified.
†No further information available.
‡Several *post hoc* findings.
**Quality score derived from 15 items as no interventions were tested.

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A protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements

Preface

One of the methodological issues with several studies evaluated in the systematic review (Chapter 2) was the lack of valid and reliable methods adopted for classifying participants' foot posture. The absence of rigorous foot screening protocols utilised by the studies presented in the review is problematic, particularly for the six studies that investigated the effect of foot posture (i.e. as an independent variable) on lower limb muscle activity during gait. In order for subsequent studies in this thesis to systematically investigate the effect of foot posture and foot orthoses on lower limb muscle activity during walking, it was evident that a clear protocol was required to distinguish 'normal' and potentially 'abnormal' foot types.

Historically, the assessment of foot posture has revolved around the 'Root paradigm' of foot biomechanics [127]. In part, this involved obtaining static measurements such as forefoot to rearfoot alignment and neutral and resting calcaneal stance position. In the ideal 'normal' foot, the forefoot is parallel to the rearfoot and the calcaneus is aligned perpendicular to the supporting surface during relaxed standing. Furthermore, this paradigm proposed that during gait the subtalar joint of the normal foot rotated about a neutral position [127]. Over time, however, several studies reported significant reliability issues with these static measurements [128-131]. In-vivo bone pin studies, although based on very small sample sizes, have also demonstrated that 'normal' foot kinematics of specific joints and combinations of joints are highly variable [80]. This presents a significant challenge for studies planning to categorise foot posture from dynamic measurements, especially since larger scale normative data is not currently available from such invasive research. In the absence of a suitable dynamic method (with normative data), a series of clinical and radiographic measurements, based on normative data, were adopted for the study in this chapter.

With regard to the different types of foot posture in the adult population, recently published data indicates that normal- and flat-arched feet are more common foot types among healthy and disease affected people [2]. While high-arched feet are also susceptable to injury and warrant greater research, this foot type is far less common. For example, one study has shown from pooled population-based Foot Posture Index data that the distribution of foot postures is skewed toward 'pronated' [2]. Therefore, in order for the findings obtained in this thesis to have greater generalisability to the wider population, the research was focused on people with normal- and flat-arched feet.

Accordingly, the aim of the study in this chapter was to develop a foot screening protocol to distinguish between participants with normal- and flat-arched feet who would then subsequently be recruited into a series of electromyographic gait studies presented in Chapters 5-7.

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Methodology article

A protocol for classifying normal- and flat-arched foot posture for research studies using clinical and radiographic measurements

George S Murley*[1,2], Hylton B Menz[2] and Karl B Landorf[1,2]

Address: [1]Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, Australia and [2]Musculoskeletal Research Centre, Faculty of Health Sciences, La Trobe University, Bundoora, Australia

Email: George S Murley* - g.murley@latrobe.edu.au; Hylton B Menz - h.menz@latrobe.edu.au; Karl B Landorf - k.landorf@latrobe.edu.au

* Corresponding author

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Abstract

Background: There are several clinical and radiological methods available to classify foot posture in research, however there is no clear strategy for selecting the most appropriate measurements. Therefore, the aim of this study was to develop a foot screening protocol to distinguish between participants with normal- and flat-arched feet who would then subsequently be recruited into a series of laboratory-based gait studies.

Methods: The foot posture of ninety-one asymptomatic young adults was assessed using two clinical measurements (normalised navicular height and arch index) and four radiological measurements taken from antero-posterior and lateral x-rays (talus-second metatarsal angle, talo-navicular coverage angle, calcaneal inclination angle and calcaneal-first metatarsal angle). Normative foot posture values were taken from the literature and used to recruit participants with normal-arched feet. Data from these participants were subsequently used to define the boundary between normal- and flat-arched feet. This information was then used to recruit participants with flat-arched feet. The relationship between the clinical and radiographic measures of foot posture was also explored.

Results: Thirty-two participants were recruited to the normal-arched study, 31 qualified for the flat-arched study and 28 participants were classified as having neither normal- or flat-arched feet and were not suitable for either study. The values obtained from the two clinical and four radiological measurements established two clearly defined foot posture groups. Correlations among clinical and radiological measures were significant ($p < 0.05$) and ranged from $r = 0.24$ to 0.70. Interestingly, the clinical measures were more strongly associated with the radiographic angles obtained from the lateral view.

Conclusion: This foot screening protocol provides a coherent strategy for researchers planning to recruit participants with normal- and flat-arched feet. However, further research is required to determine whether foot posture variations in the sagittal, transverse or both planes provide the best descriptor of the flat foot.

Background

Foot posture, like most human anthropometric characteristics, varies considerably among children, adults and the older population [1]. Some variations in foot posture are associated with changes in lower limb motion [2,3] and muscle activity [4], and are strongly influenced by some systemic conditions, such as neurological [5] and rheumatological diseases [6]. These factors add weight to the view that functional differences exist between different foot types. Therefore, there is a need for strategies to accurately classify foot posture and define normal and potentially 'abnormal' foot types.

To address this issue, normative data are now available that classify foot posture using the following techniques: visual observation [1]; measurement of navicular height [7] or midfoot height [8]; footprint measures [7,9]; and angular measures derived from radiographs [10]. As interpretation of the clinical techniques is confounded by soft tissue overlying the skeletal structure of the foot, radiographic techniques are regarded as the gold-standard for assessing skeletal alignment of the foot in a static weight-bearing position [11]. Therefore, angular foot measurements derived from x-rays are often used to validate clinical measures of foot posture [8,12,13]. As such, it would be useful to have clinical measurements that accurately predict angular measurements derived from radiographs, as this process would reduce: (i) the expense of obtaining x-rays for a study; and (ii) unnecessary referral of participants for x-ray examination.

There have already been some attempts to address this issue. Menz and Munteanu [12] evaluated the association between three clinical measurements (arch index [9], foot posture index [2], and navicular height [14]) with three lateral-view x-ray measurements (navicular height, calcaneal inclination angle, and the calcaneal-first metatarsal angle) in 95 older participants. All three clinical measures demonstrated significant correlations with the x-ray measures, with the navicular height and arch index clinical measurements having the strongest correlations. In addition, Saltzman et al. [14] investigated the association between various measures of arch height and radiological measures for 100 patients with orthopaedic conditions (mean age, 46 years). The arch height measures were all reported to have good to strong correlations with angles derived from lateral x-ray views. Other clinical measures, such as the arch ratio have also been validated using x-rays [8]. However, further research is still required to validate clinical measures with additional angles of the foot, particularly angles assessed from the anterior-posterior view, and to validate measurements specific to the young adult population.

The major drawback for researchers is that the available literature does not provide a pathway for choosing a series of clinical and radiological measurements to screen participants' foot posture. A combination of validated clinical measurements and normative data would allow researchers to have a clear protocol to follow when screening participants' foot posture, whether for laboratory-based research or epidemiological studies.

Accordingly, the primary aim of this study was to develop a foot screening protocol using clinical and radiographic measurements for the purpose of recruiting participants with normal- and flat-arched feet for a series of laboratory-based gait studies. The secondary aim was to explore relationships between the clinical and radiographic measures of foot posture.

Methods

Participants

Ninety-one asymptomatic young adults were recruited (45 male and 46 female) aged 18 to 47 years (mean ± SD, 23.2 ± 5.6 years) (Table 1). The participants were without symptoms of macrovascular (e.g. angina, stroke, peripheral vascular disease) and/or neuromuscular disease, or any biomechanical abnormalities which affected their ability to walk. Ethical approval was obtained for the study from the La Trobe University Human Ethics Committee (Ethics ID: FHEC06/205) and it was registered with the Radiation Safety Committee of the Victorian Department of Human Services. The x-rays were performed in accordance with the Australian Radiation Protection and Nuclear Safety Agency Code of Practice for the *Exposure of Humans to Ionizing Radiation for Research Purposes* (2005) [15].

Participants were primarily recruited from the student and staff community at La Trobe University. The foot screening protocol was developed to recruit participants with normal-arched feet, which provided normative reference values for two radiographic measures of foot posture (talo-navicular coverage angle and calcaneal-first metatarsal angle). Data from these participants were subsequently used to define the boundary between normal- and flat-arched feet. This information was then used to recruit participants with flat-arched feet. Therefore, the foot screening protocol was developed by utilising: (i) published normative data for clinical and radiological measurements; and (ii) radiological measurements obtained from the first study investigating normal-arched feet (Figure 1 and 2). Participants with high-arched feet were not required for this study. Although high-arched feet are susceptive to injury and warrant greater research [16,17], this foot type is far less common than normal- and flat-arched feet [1], thus we chose to focus on two participant groups that would have greater generalisability to the wider population.

Stage 1: Clinical measurements

The first stage of the screening protocol involved two clinical measures of foot posture; (i) the arch index [9], and

Table 1: Participant anthropometric and foot posture characteristics

	Foot posture groups		
	Flat-arch n = 31	Normal-arch n = 32	Others n = 28
General anthropometric			
Gender ratio (female/male)	16/15	16/12	17/15
Age mean ± SD (years)	22.0 ± 4.3	23.5 ± 5.7	24.2 ± 6.7
Height mean ± SD (cm)	171.0 ± 10.0	169.7 ± 9.7	n/a
Weight mean ± SD (Kg)	73.3 ± 15.50	69.9 ± 13.6	n/a
Left or right foot count	16 right 15 left	14 right 18 left	13 right 15 left
Clinical measurements			
AI mean ± SD	0.30 ± 0.07*	0.24 ± 0.04*	0.23 ± 0.02
NNHt mean ± SD	0.18 ± 0.04†	0.27 ± 0.03†	0.25 ± 0.06
Radiographic measurements			
CIA mean ± SD (degrees)	16.1 ± 5.0#	20.9 ± 3.4#	24.9 ± 4.9
C1MA mean ± SD (degrees)	141.7 ± 6.7‡	132.8 ± 4.0‡	129.0 ± 7.7
TNCA mean ± SD (degrees)	27.5 ± 8.9^	12.5 ± 8.6^	13.0 ± 6.5
T2MA mean ± SD (degrees)	27.5 ± 10.2¥	13.3 ± 6.3¥	13.8 ± 5.3

AI – arch index, NNHt – normalised navicular height truncated, CIA – calcaneal inclination angle, C1MA – calcaneal first metatarsal angle, TNCA – talo-navicular coverage angle, T2MA – talus-second metatarsal angle.
Mean differences and 95% confidence interval (CI) expressed relative to normal-arch.
Statistically significant findings for comparisons listed below ($p < 0.001$):
* AI: mean difference 0.05, 95% CI 0.03 to 0.08
† NNHt: mean difference -0.09, 95% CI -0.11 to -0.07
# CIA: mean difference -4.8°, 95% CI -6.9° to -2.6°
‡ C1MA: mean difference 9.0°, 95% CI 6.2° to 11.7°
^ TNCA: mean difference 15.0°, 95% CI 10.7° to 19.3°
¥ T2MA: mean difference 14.2°, 95% CI 9.9° to 18.4°

(ii) normalised navicular height truncated [18]. These 'ratio' measurements have moderate to high correlations with angular measurements derived from radiographs [11,14,19], which provide the most valid representation of skeletal foot alignment [12]. Although the arch index and normalised navicular height measurements have comparable reliability to other measures of arch height, these were selected because of their ease of use and demonstrated validity with skeletal alignment measured via radiographs [12]. Additionally, the arch index is sensitive to age-related changes in foot posture [7] and is strongly associated with both maximum force and peak pressure in the midfoot during walking [20]. The primary purpose of using the clinical tests in this study was to avoid unnecessary referral of participants for radiographic assessment.

The arch index was calculated as the ratio of area of the middle third of the footprint to the entire footprint area not including the toes, with a higher ratio indicating a flatter foot [9] (Figure 3). The footprint was taken using carbon paper and a graphics tablet was used to calculate the surface area in each third of the foot.

Normalised navicular height truncated is the ratio of navicular height relative to the truncated length of the foot. Navicular height is the distance measured from the most medial prominence of the navicular tuberosity to the supporting surface. Foot length is truncated by measuring the perpendicular distance from the first metatarsophalangeal joint to the most posterior aspect of the heel [18], with a lower normalised navicular height ratio indicating a flatter foot (Figure 4).

To determine normal values for the arch index and normalised navicular height, we requested and were provided with raw foot posture measurements from Scott and colleagues [7] comprising data from 50 healthy young adults (26 female and 24 male with a mean age ± SD of 20.9 ± 2.6 years). The participants reported on by Scott and colleagues [7] were of similar age to the target participants for our study (Figure 1).

For the normal-arched foot study, participants qualified for the second stage of the screening assessment involving radiographic evaluation when either the arch index and normalised navicular height scores fell within ± 1 standard deviation (SD) of the mean values adapted from Scott and colleagues [7] (Figure 1). A threshold of ± 1 SD was selected as the 'normal limits' of several human physiological and anthropometric characteristics are frequently

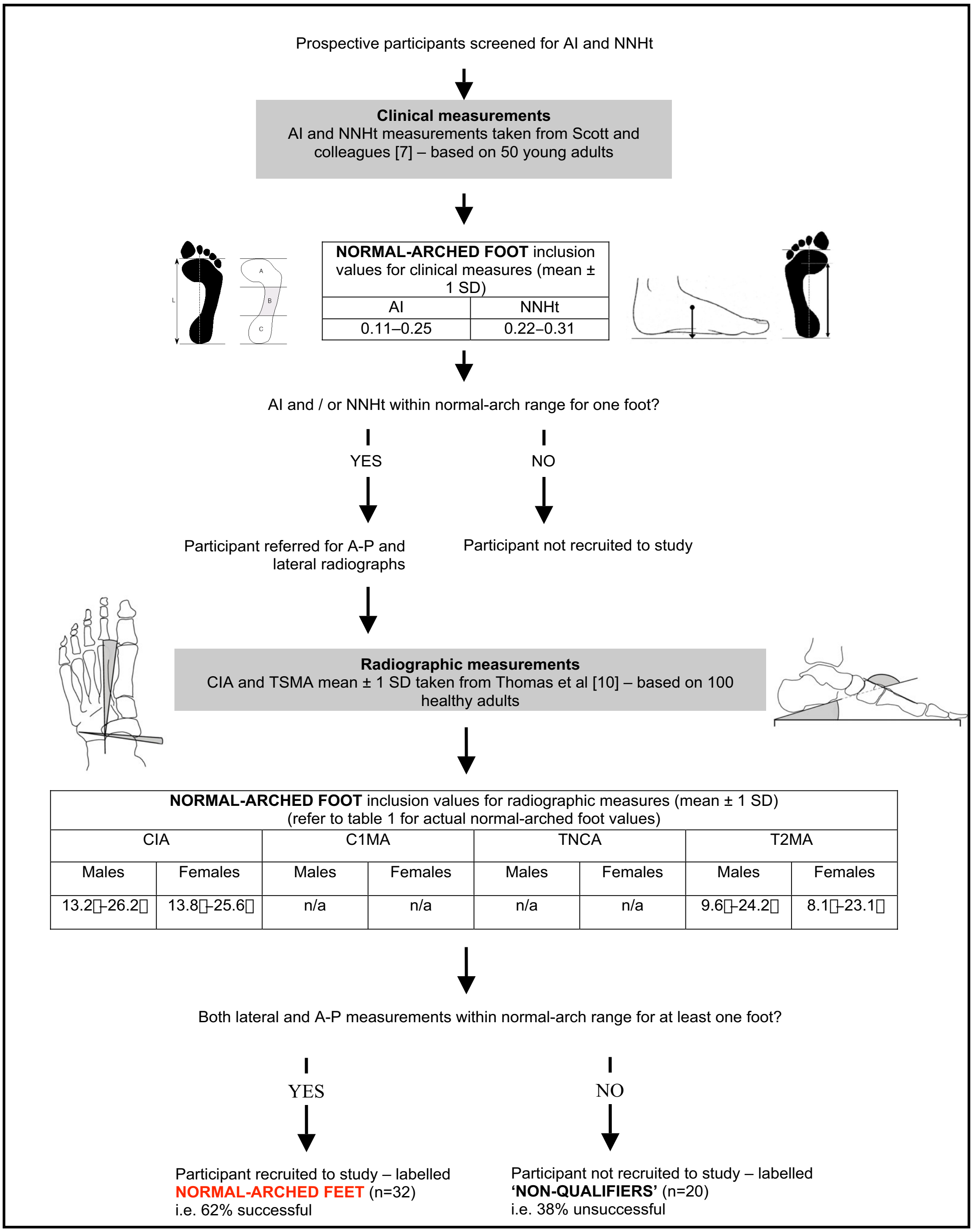

Figure 1

Screening protocol for normal-arched foot posture. Flow chart shows how the foot posture screening protocol was derived from normative data. * Values derived from Scott and colleagues [7]. CIA – calcaneal inclination angle, C1MA – calcaneal-first metatarsal angle, TNCA – talo-navicular coverage angle, T2MA – talus-second metatarsal angle.

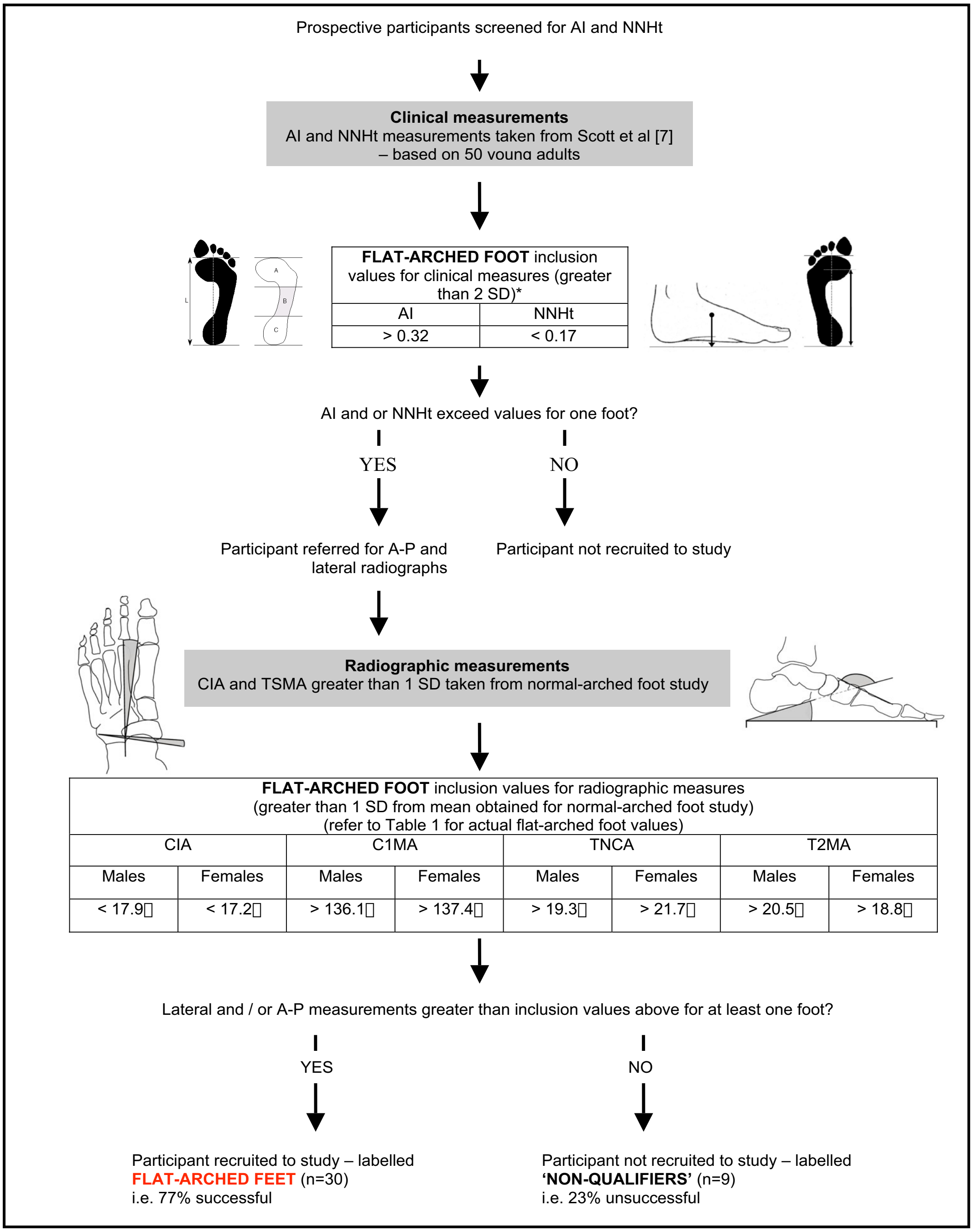

Figure 2

Screening protocol for flat-arched foot posture. Flow chart shows how the foot posture screening protocol was derived from normative data. * Values derived from Scott and colleagues [7]. The rationale for using 2 SD standard deviations was to increase the likelihood of participants with flat-arched feet qualifying for inclusion via radiographic appraisal. CIA – calcaneal inclination angle, C1MA – calcaneal-first metatarsal angle, TNCA – talo-navicular coverage angle, T2MA – talus-second metatarsal angle.

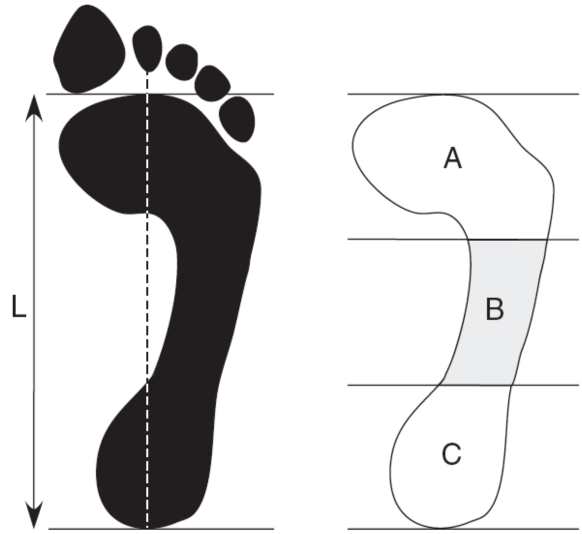

Figure 3
Arch index. Footprint with reference lines for calculating the arch index. The length of the foot (excluding the toes) is divided into equal thirds to give three regions: A – forefoot; B – midfoot; and C – heel. The arch index is then calculated by dividing the midfoot region (B) by the entire footprint area (i.e. Arch index = B/[A+B+C]).

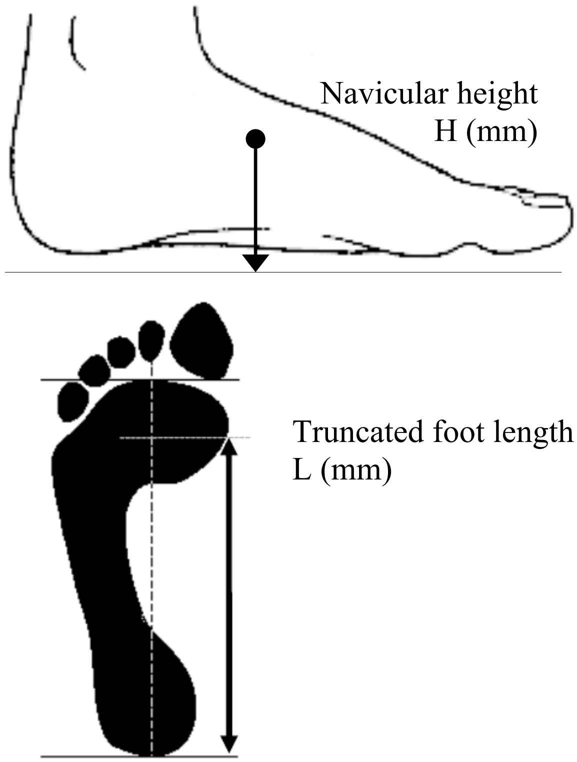

Figure 4
Normalised navicular height (truncated). Calculating normalised navicular height truncated. The distance between the supporting surface and the navicular tuberosity is measured. Foot length is truncated by measuring the perpendicular distance from the 1st metatarsophalangeal joint to the most posterior aspect of the heel. Normalised navicular height truncated is calculated by dividing the height of the navicular tuberosity from the ground (H) by the truncated foot length (L) (i.e. Normalised navicular height truncated = H/L).

defined to lie within 1–2 standard deviations of the population mean [21].

Stage 2: Radiographic measurements
The second screening stage involved two bilateral radiographs comprising: (i) antero-posterior (A-P) and (ii) lateral views obtained with the subject weight-bearing in a relaxed bipedal stance position. From the A-P view, the talus-second metatarsal angle and the talo-navicular coverage angle were assessed (Figure 5). From the lateral view, the calcaneal inclination angle and the calcaneal-first metatarsal angle were assessed (Figure 5). These angles were chosen to represent foot posture based on: (i) ease of measurement and good reliability; and (ii) degree by which they reflect foot posture in both the sagittal and transverse planes.

Anterior-posterior radiographic angles
The talo-navicular coverage angle is formed by the bisection of the anterior-medial and the anterior-lateral extremes of the talar head and the bisection of the proximal articular surface of the navicular [22] (Figure 5). The talus-second metatarsal angle is formed by the bisection of the second metatarsal and a line perpendicular to a line connecting the anterior-medial and the anterior-lateral extremes of the talar head [10] (Figure 5). Angles measured from the A-P view reflect transverse plane alignment of the midfoot and forefoot, with larger angles for the talo-navicular coverage angle and talus-second metatarsal angles indicating a flatter foot.

Lateral radiographic angles
The calcaneal inclination angle is the angle between the inferior surface of the calcaneus and the supporting surface [14] (Figure 5). The calcaneal-first metatarsal angle is the angle formed by the inferior surface of the calcaneus and a line parallel to the dorsum of the mid-shaft of the first metatarsal. Angles measured from the lateral view reflect sagittal plane alignment of the hindfoot and forefoot, with a lower calcaneal inclination angle and a greater calcaneal-first metatarsal angle indicating a flatter foot (Figure 5).

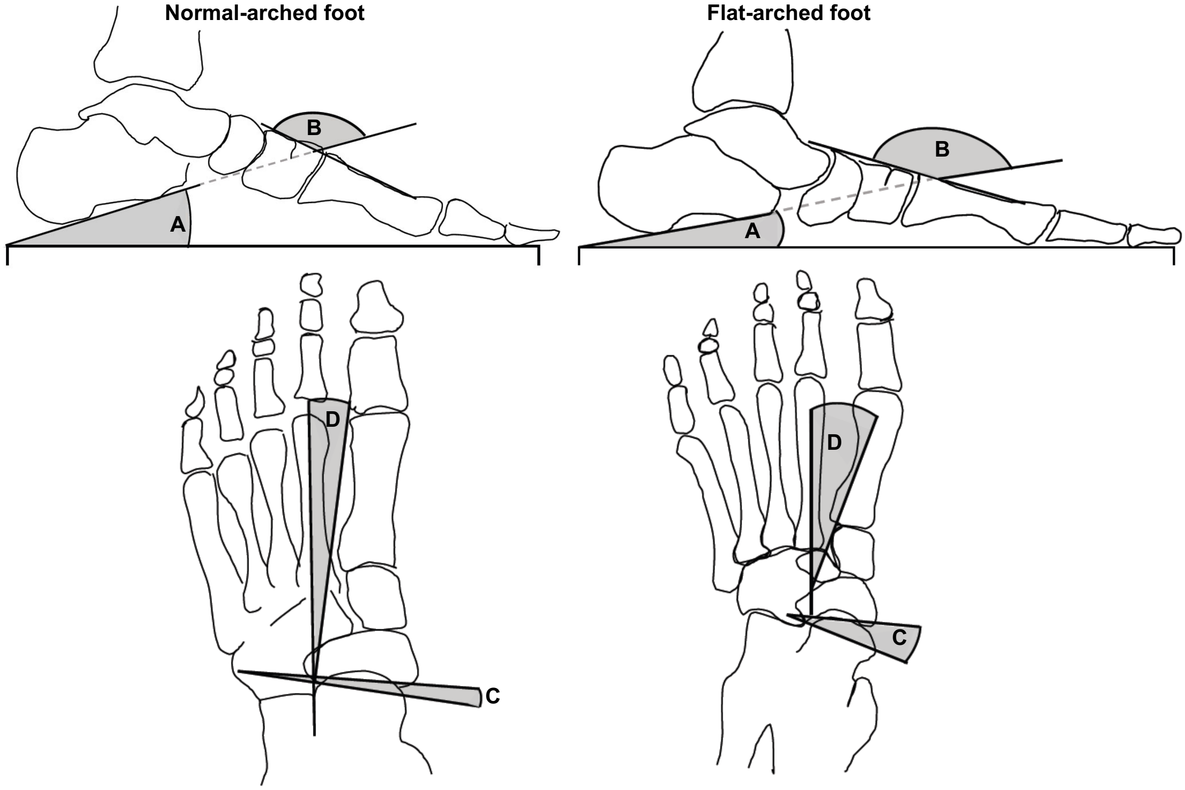

Figure 5
Radiographic measurements. Traces from two representative participants illustrate x-ray angular measurements from normal (left) and flat-arched (right) foot posture. Lateral views (top) show: calcaneal inclination angle; calcaneal-first metatarsal angle; anterior posterior views (bottom) show: talonavicular coverage angle; talus second metatarsal angle. A – calcaneal inclination angle, B – calcaneal-first metatarsal angle, C – talo-navicular coverage angle, D – talus-second metatarsal angle. Angle A *decreases* with flat-arched foot posture; angle B, C and D *increase* with flat-arched foot posture, compared to the normal-arched foot posture.

Normal values for the calcaneal-inclination angle were derived from a study by Thomas and colleagues [10] comprising 100 adults (50 females and 50 males with a mean age of 34.7 years for females and 34.3 years for males), which represents a slightly older population to that included in our study.

As shown in Figure 2, the talo-navicular coverage angle and calcaneal-first metatarsal angle taken from the initial normal-arched foot radiographs were used to calculate reference values for the flat-arched foot study. Participants qualified for the flat-arched study when both measures from the lateral and/or anterior-posterior views exceeded 1 SD from the actual mean values reported for the normal study. The decision to accept either the lateral or antero-posterior measurements was based on the lack of consensus regarding which plane best represents the 'flat-arched foot'.

Reliability of clinical and radiological measures

The reliability of the clinical measurements has been reported to be moderate to excellent, with intra-class correlation coefficients (ICCs) of 0.67 and 0.99 for normalised navicular height [23] and the arch index [12], respectively. For radiographic measures, the ICCs are reported to be excellent for the calcaneal inclination angle (0.98), calcaneal-first metatarsal angle (0.99) [12] and good for the talo-navicular coverage angle (0.79) [24]. As the reliability of the talus-second metatarsal angle is unknown, we evaluated intra- and inter-tester reliability for this angle. Intra-tester reliability was evaluated by a podiatrist with seven years of post-graduate experience. Inter-tester reliability was evaluated between the same tester and one other tester with four years of undergraduate podiatry training. The x-ray measurements were marked onto clear-plastic overhead transparencies placed over the x-ray using a permanent fine-point marker. For intra-tester reliability, the tester was blinded from the initial measurements when they performed their re-test session approximately two-weeks later. For inter-tester reliability, the examiners evaluated the x-rays independently, were blinded to each other's assessment and the data for each angle was recorded from single measurements. Testers were not blinded from the participants' anthropometric measurements (e.g. clinical measures of foot posture) for either the intra-tester or intra-tester components of the study.

Statistical analysis

To satisfy the assumption of independence with statistical analysis, only measurements from a single foot were analysed [25]. All data were explored for normal distribution by evaluating skewness and kurtosis. The relative reliability of the talo-navicular coverage angle was assessed using type (3,1) intra-class correlation coefficients and absolute limits of agreement [26]. To evaluate the anthropometric-related differences between the normal-arched and flat-arched groups, a series of independent-samples *t*-tests were used. To determine the degree of association between clinical and radiographic measurements, data from the normal-arched, flat-arched and non-qualifying groups were pooled and Pearson *r* correlation coefficients were calculated. For both the *t*-tests and correlation coefficients, the level of significance was set at 0.05. All statistical tests were conducted using SPSS version 13 for Windows (SPSS Inc, Chicago, IL).

Results

Participant characteristics

The mean ± SD age, height and body mass of the study sample were 23.2 ± 5.6 years, 1.70 ± 0.10 m, and 71.6 ± 14.6 kg, respectively. Following the radiographic assessment, 32 participants were recruited to the normal-arched study, 31 qualified for the flat-arched foot study and 28 participants were classified as having neither normal- or flat-arched feet and were not suitable for either study. Anthropometric data for the normal-arched, flat-arched and non-qualifying participants are summarised in Table 1. Scatter plots of the distributions of all participants' clinical and radiological measurements are shown in Figure 6 and 7.

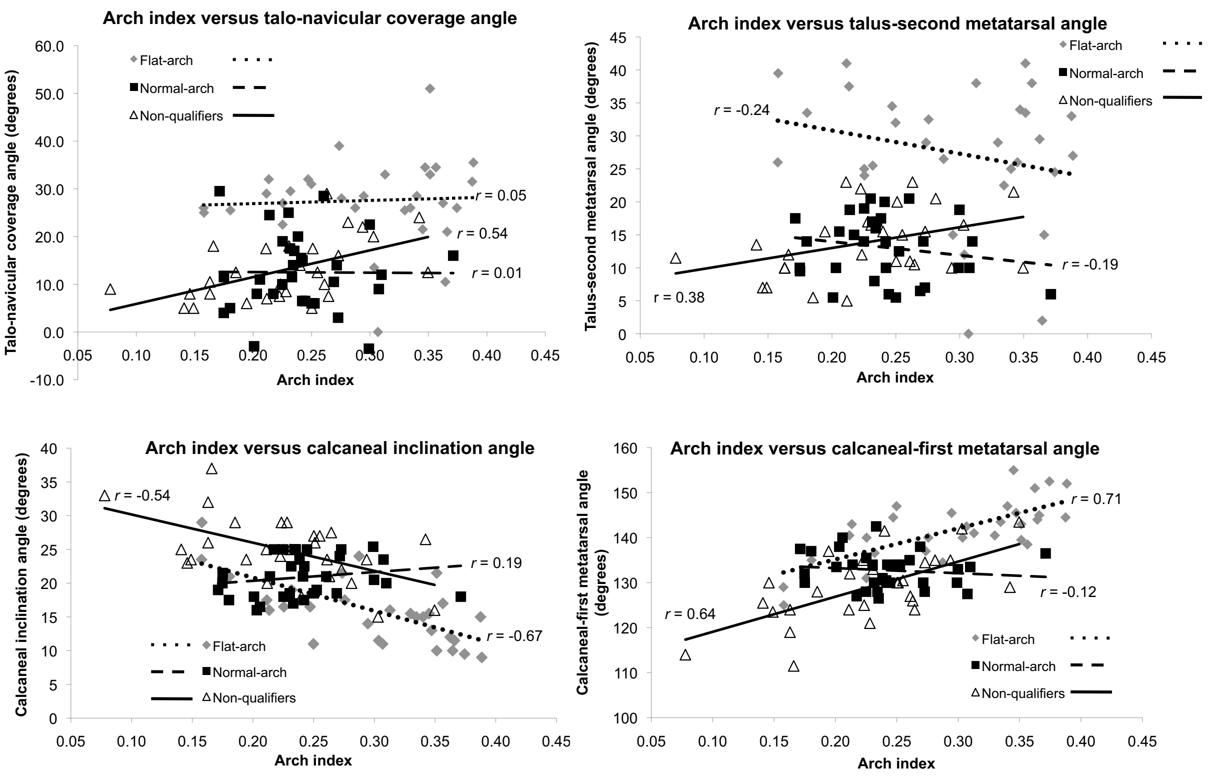

Figure 6
Arch index versus radiographic measures for each foot posture group. Scatter plots with trend lines for the arch index and radiographic measures of foot posture show the distribution of values for normal-arch, flat-arch and non-qualifying foot postures.

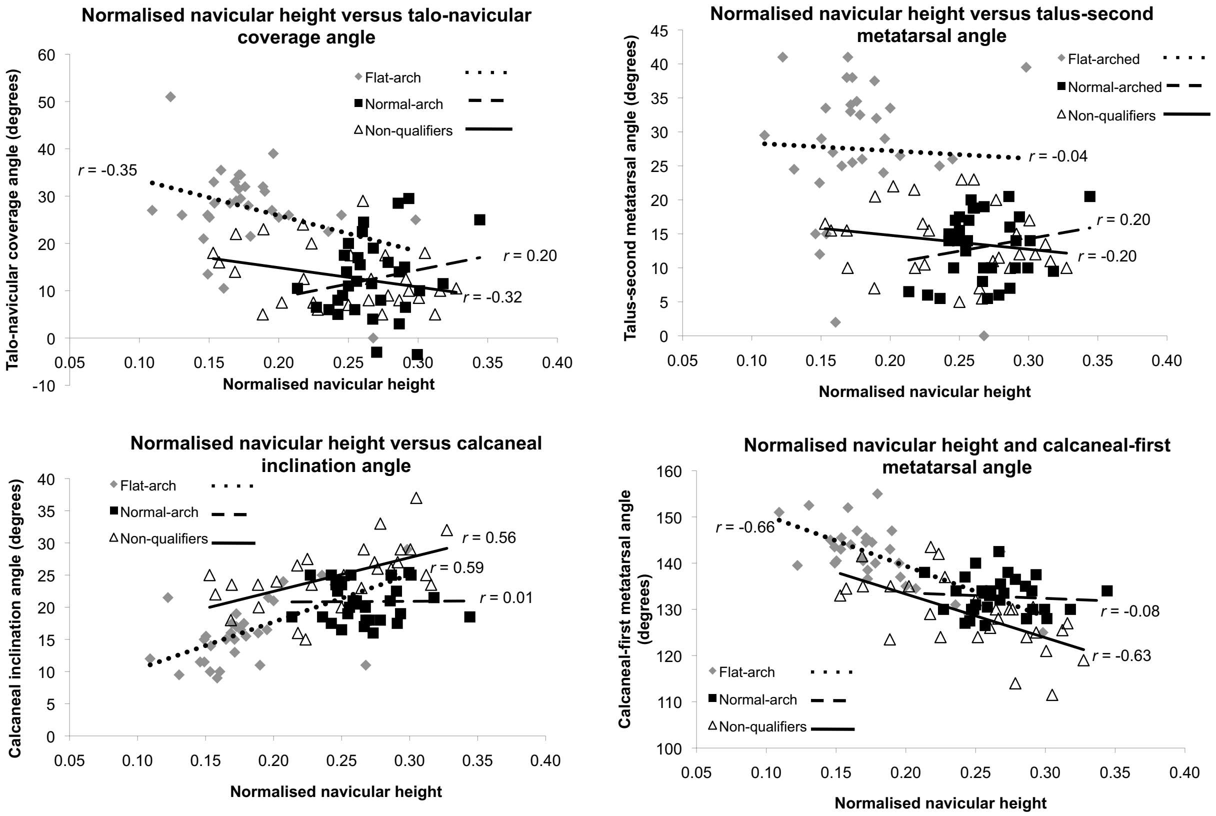

Figure 7
Normalised navicular height versus radiographic measures for each foot posture group. Scatter plots with trend lines for the normalised navicular height and radiographic measures of foot posture show the distribution of values for normal-arch, flat-arch and non-qualifying foot postures.

Reliability of the talus-second metatarsal angle

The within- and between-tester reliability of measuring the talus-second metatarsal angle is shown in Table 2. The talus-second metatarsal angle demonstrated good to excellent intra-rater reliability with left and right foot ICCs ranging from 0.71 to 0.91 and absolute random error ranging from 7.1 to 12.2°. Inter-rater reliability for the talus-second metatarsal angle was moderate to very good with left and right foot ICCs ranging from 0.68 to 0.78 and absolute random error ranging from 5.6 to 7.1° (Table 2).

Anthropometric differences between normal and flat-arched groups

General anthropometric characteristics including age, height and weight were not significantly different between the normal and flat-arched groups. However, all clinical and radiological differences were statistically different between groups ($p < 0.001$) (Table 1).

Associations between clinical and radiological measures of foot posture

The relationships among the clinical and radiological measures (for the entire group n = 91) are shown in Table 3. Both clinical measures were significantly correlated with all radiographic angles, with r values ranging from 0.24 to 0.70. The clinical measurements displayed a moderate to strong relationship with radiographic measurement from the lateral view, with r values ranging from 0.59 to 0.70. However, the clinical measurements displayed only a weak to moderate relationship with radiographic measurement from the antero-posterior view, with r values ranging from 0.24 to 0.56. The strongest association between clinical and radiological measures occurred for the normalised navicular height and calcaneal first metatarsal inclination angle (r = 0.70). For the clinical measures, arch index and normalised navicular height displayed a significant negative correlation to each other (r = -0.58). For the radiographic measures, the lat-

Table 2: Relative and absolute reliability of measuring the talus-second metatarsal angle (T2MA)

	RELATIVE RELIABILITY	ABSOLUTE RELIABILITY	
	Type (3,1) ICC (95% CI)	Systematic bias (% mean difference)	Random error (95% LoA)
Within-rater			
left feet (n = 51)	0.91 (0.85 – 0.95)*	- 0.5°	7.1°
right feet (n = 51)	0.71 (0.55 – 0.83)*	- 0.3°	12.2°
Between-rater			
left feet (n = 41)	0.78 (0.62 – 0.88)*	- 1.0°	5.6°
right feet (n = 41)	0.68 (0.47 – 0.82)*	1.5°	7.1°

*Significant at $p < 0.05$

eral view angles were significantly correlated with angles obtained from the antero-posterior view, with r values ranging from 0.25 to 0.47. Figure 6 and 7 show scatter plots and associations between clinical and radiographic measures for each foot posture group.

Discussion

The purpose of developing this screening protocol was to assist with the recruitment of participants into a series of laboratory-based gait studies investigating functional differences between normal-arched and flat-arched feet. For the normal-arched study, the clinical and radiographic values were derived from two published sources [7,10], which describe normative foot posture in healthy and asymptomatic adult populations. Radiographic values obtained from the normal-arched foot study were subsequently used to calculate inclusion values for the flat-arched foot study. This resulted in normal and flat-arched groups with significantly different foot posture characteristics without systematic bias for age, height or weight between the groups.

Participants with normal-arched feet in this study displayed a similar mean arch index value (0.24 ± 0.04) to those reported by Cavanagh and Rodgers [9] (0.23 ± 0.05) for 107 subjects (mean age, 30 years). Interestingly, our study found a higher mean arch index value (0.24 ± 0.04) compared to Scott and colleagues [7] (0.18 ± 0.07), from which our normative reference values were derived. This difference may be due to our study reporting arch index values from only participants who satisfied the radiographic inclusion criteria and not the full range of participants who underwent clinical screening. Accordingly, we recommend using the values from our study tabulated in Figure 8, as our normative arch index values were validated with radiographs.

It is difficult to compare the arch index values used to define the participants with flat-arched feet in our study (0.30 ± 0.07) to those of Cavanagh and Rodgers [9] (≥ 0.26), as they defined the 'flat-arched foot' to lie within the top 25% of the distribution of arch index scores obtained from the 107 subjects. In contrast, we defined the flat-arched foot as greater than two standard deviations from the normative mean (as reported by Scott and colleagues [7]). The rationale for using two standard deviations was to increase the likelihood of participants with flat-arched feet qualifying for inclusion via radiographic

Table 3: Pearson *r* values comparing the radiographic and clinical measures

	Radiographic measures				**Clinical measurements**	
	Lateral view		Anterior-posterior view			
	CIA	C1MA	TNCA	T2MA	AI	NNHt
Clinical measurements						
AI	- 0.59**	0.66**	0.40**	0.24*	-	- 0.58**
NNHt	0.60**	- 0.70**	- 0.56**	- 0.47**	-	-
Radiographic measurements						
Anterior-posterior view T2MA	- 0.25*	0.38**	-	-	-	-
TNCA	- 0.36**	0.47**	-	-	-	-

AI – arch index, NNHt – normalised navicular height truncated, CIA – calcaneal inclination angle, C1MA – calcaneal first metatarsal angle, TNCA – talo-navicular coverage angle, T2MA – talus-second metatarsal angle.
*Significant at $p < 0.05$, **Significant at $p < 0.01$

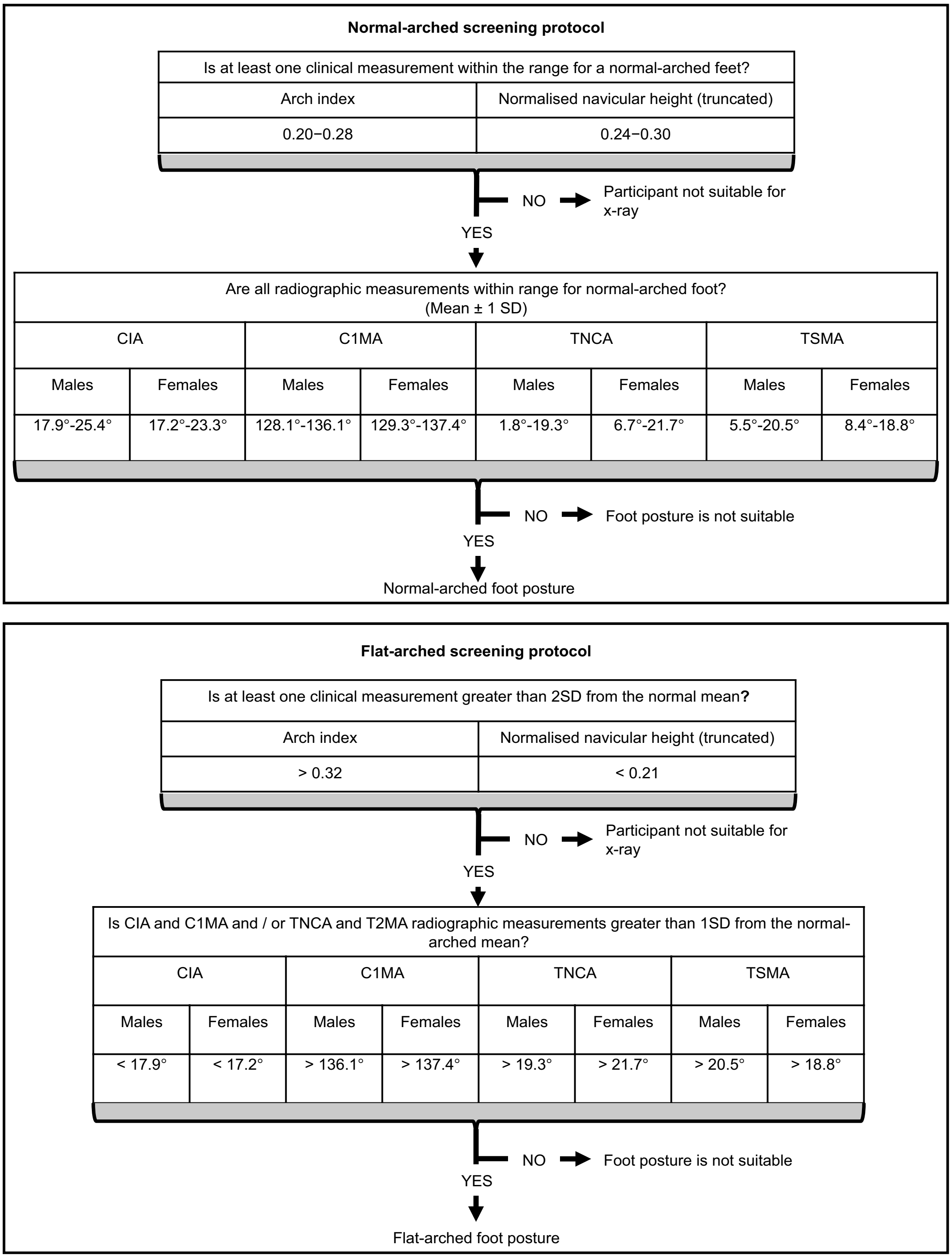

Figure 8

Screening protocol for normal- and flat-arched foot posture. Flow chart shows how the foot posture screening protocol can be applied to future studies recruiting participants with normal- and flat-arched foot posture. CIA – calcaneal inclination angle, C1MA – calcaneal-first metatarsal angle, TNCA – talo-navicular coverage angle, T2MA – talus-second metatarsal angle.

appraisal. Therefore, it is important to highlight that the arch index reference values that defined flat-arched feet in our study were stricter, which resulted in the recruitment of flatter-arched feet compared to those reported by Cavanagh and Rodgers [9].

From the normal-arched feet, we report the first normative values published for the calcaneal-first metatarsal angle and talo-navicular coverage angle from a young adult population (Table 1). The actual values obtained for the calcaneal inclination angle and talus-second metatarsal angle from normal-arched feet in this study were within 1.4° to 2.9°, respectively, of those reported by Thomas and colleagues [10] for 100 subjects (mean age, 35 years).

With respect to the relationship between clinical and radiographic measures, all correlations were statistically significant, with the associations ranging from moderate to strong (r = 0.24 to 0.70). Of the two clinical measures, normalised navicular height provided the strongest association with all radiographic angles measured from both the A-P and lateral views. These findings are different to the associations reported by Menz and Munteanu [12] who reported the arch index to provide the strongest correlation for the calcaneal inclination angle and calcaneal-first metatarsal angle from 95 older participants (mean age, 79 years). This discrepancy may be due to age-related differences in body mass of younger compared to older adult populations, as the arch index is confounded by variations in soft tissue composition of the foot between different individuals [27].

Furthermore, while both clinical measures were significantly correlated with all radiographic angles, the arch index and normalised navicular height were most strongly associated with the calcaneal inclination angle and calcaneal-first metatarsal angle obtained from the lateral view. Therefore, we found the arch index and normalised navicular height measurements were more sensitive to detecting flat-arched feet associated with angles measured from the lateral view, which better represents sagittal plane alignment. Consequently, using the arch index and normalised navicular height measurements in the current study may have lead to a bias when recruiting participants with flat arches characterised by a low calcaneal inclination angle and high calcaneal-first metatarsal angle. Further research is required to validate a reliable clinical test that is sensitive to radiographic variations with transverse plane deformity, such as the recently reported foot mobility magnitude test [28]. It is also not clear whether foot posture variations in the sagittal, transverse or both planes provide the best descriptor of the flat-arched foot. For example, loss of the tibialis posterior tendon function with disease is associated with abnormal joint moments in both the sagittal and transverse midfoot planes [29,30]. Ness et al. [29] reported significantly less forefoot plantarflexion and less abduction during walking in 34 patients with tibialis posterior tendon dysfunction compared to 25 healthy controls. This would indicate that an acquired flatfoot deformity is characterised by altered foot posture in multiple planes. However, the variants of foot posture investigated in our study present a different set of considerations because pain and dysfunction were not present.

The protocol for screening foot posture described here could be applied to future research studies specifically recruiting participants with normal- and flat-arched foot posture. With the moderate correlation between clinical and radiographic measures of foot posture, we recommend the arch index and normalised navicular height measurements be used during initial foot screening to identify potentially suitable participants, followed by radiographic evaluation including lateral and antero-posterior views.

This foot screening protocol needs to be viewed in light of some limitations. The intra- and inter-tester reliability of the talus-second metatarsal angle ranged from moderate to excellent with ICCs between 0.68 and 0.91 and limits of agreement ranging from 5.6° to 12.1°, respectively. Another drawback from this study is that the homogeneity of the participant group in this investigation limits the generalization of our findings to a young adult population.

Further research is required to provide validation of radiographic measures of foot posture by investigating whether the radiographic angles are related to functional differences during gait. Moreover, large prospective studies investigating the relationship between radiographic measures of foot posture and injury could provide further validation of the radiographic measures.

Conclusion

The foot screening protocol presented here provides a strategy for recruiting participants with normal- and flat-arched foot posture, including reference values for clinical and radiographic measurement. The arch index and normalised navicular height ratios provide valid and reliable measures of foot posture. Normalised navicular height displayed the strongest association with radiographic angles, especially the calcaneal inclination angle. Further research is required to determine whether foot posture variations in the sagittal, transverse or both planes provide the best descriptor of the flat-arched foot. In the absence of this research, we recommend the protocol outlined in this article to classify foot posture in research.

Competing interests

HBM and KBL are Editor-in-Chief and Deputy Editor-in-Chief, respectively, of *Journal of Foot and Ankle Research*. It is journal policy that editors are removed from the peer review and editorial decision-making processes for papers they have co-authored.

Authors' contributions

GSM, HBM and KBL conceived the idea and obtained funding for the study. GSM, HBM and KBL designed the study protocol. GSM recruited/screened participants' foot posture and evaluated the radiographs. GSM, HBM and KBL drafted the manuscript. All authors have read and approved the final manuscript.

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4.0 Electromyography methods

4.1 Introduction

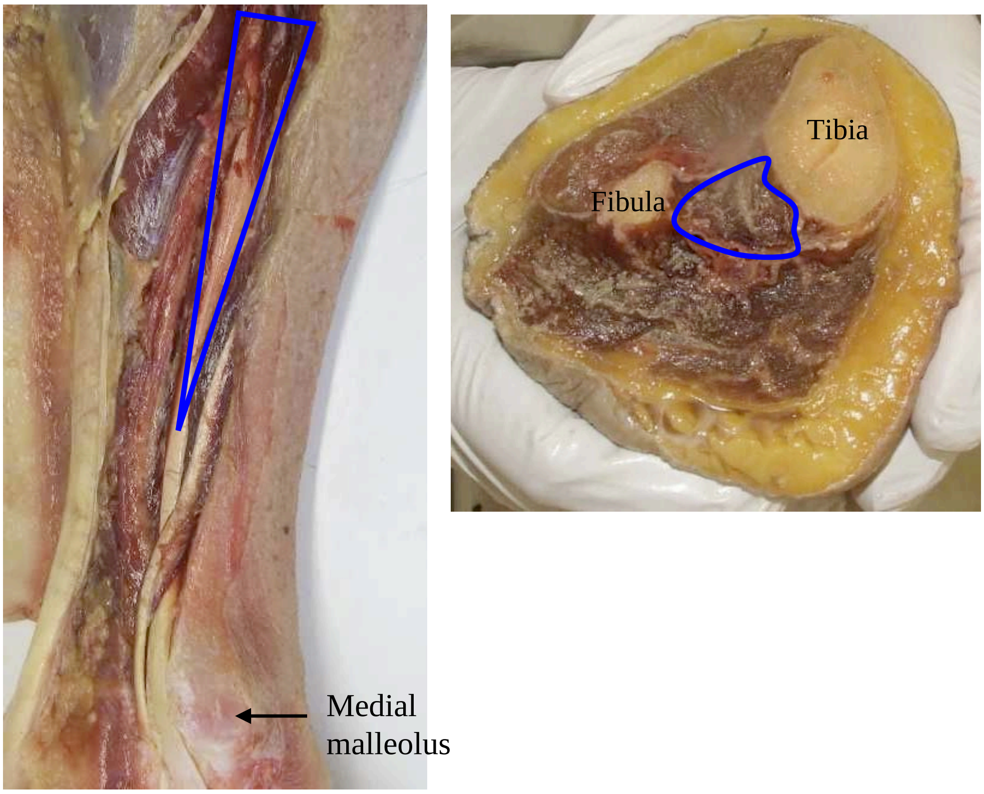
The rationale for selecting specific muscle was based on those that were likely to be strongly influenced by foot posture. Therefore, the following muscles were chosen for investigation in this thesis; tibialis posterior, peroneus longus, tibialis anterior and medial gastrocnemius. The following section presents an overview of the anatomy and function of these muscles.

4.1.1 Tibialis posterior anatomy and function

The tibialis posterior muscle is located within the deep posterior compartment of the leg, arising from the adjacent posterior surfaces of the tibia, fibula and interosseus membrane (Figure 4.1) [132]. The tendon of tibialis posterior forms in the distal third of the leg and changes direction to enter the foot where it passes behind the medial malleolus. The location of the tibialis posterior tendon relative to the axes of the subtalar and ankle joints facilitates inversion and plantarflexion, respectively. Tibialis posterior is described as the most powerful supinator of the rearfoot as a result of the large inverter moment arm acting on the subtalar joint [133,134].

Figure 4.1. Gross anatomy of the lower leg from a non-embalmed cadaveric limb. (a) medial supra-malleolar leg region; and (b) cross-section of lower leg. Delineations of the tibialis posterior musculotendinous section (a) and tibialis posterior muscle (b) are shown in blue.

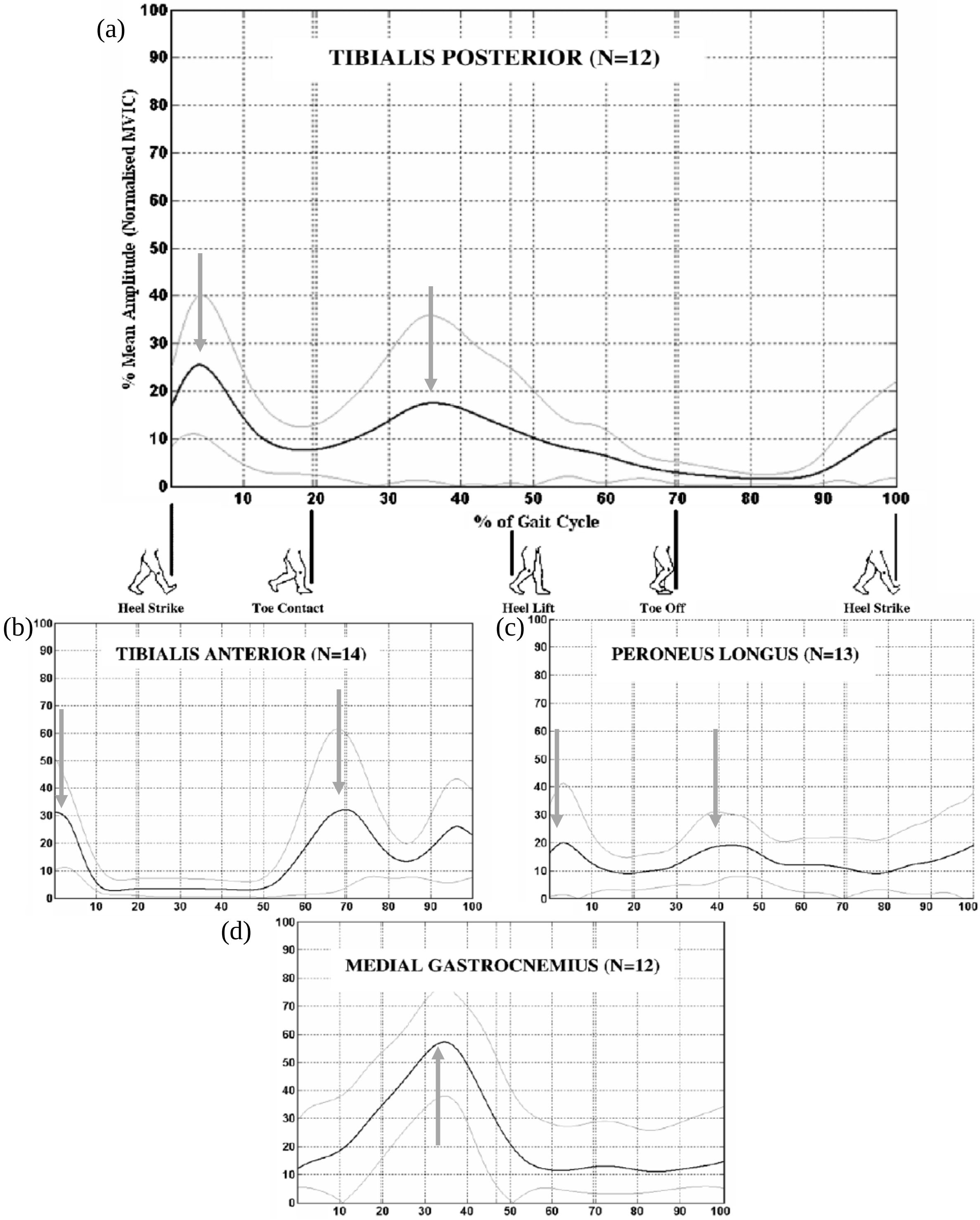
(a) (b)



Pictures taken from Semple R, Murley GS, Woodburn J, Turner DE. Tibialis posterior in health and disease: a review of structure and function with specific reference to electromyographic studies. *Journal of Foot and Ankle Research* 2009; 2(24):1-8

The tibialis posterior tendon has multiple insertions within the foot [135,136], with the largest component extending to the navicular tuberosity. Other components attach to the remaining tarsal bones, the middle three metatarsals and the flexor hallucis brevis muscle. The complex anatomy of the insertion sites function to stabilise the medial longitudinal arch. While variations of the insertion have been reported in the literature [135,136], the structural and functional significance of these variations is unknown. Dynamic electromyography (EMG) studies indicate that tibialis posterior demonstrates two bursts of activity during walking; the first burst occurs during contact phase and the second burst occurs during the midstance to propulsion phase of gait [21] (Figure 4.2).

Figure 4.2. Ensemble-averaged EMG curves for (a) tibialis posterior, (b) tibialis anterior, (c) peroneus longus and (d) medial gastrocnemius from participants with normal-arched feet. Vertical arrows indicate the main bursts that occur during walking gait.



Picture taken from Murley GS, Buldt AK, Trump PJ, Wickham JB. Tibialis posterior EMG activity during barefoot walking in people with neutral foot posture. *Journal of Electromyography and Kinesiology* 2009;19:e69-e77.

4.1.2 Peroneus longus anatomy and function

Peroneus longus occupies most of the superior lateral compartment of the leg and arises from the anterior head of the fibula and upper two thirds of the lateral surface of the fibula shaft. The tendon of peroneus longus arises approximately midway along the lateral compartment [137]. The tendon passes posterior to the lateral malleolus where it shares a synovial shealth with peroneus brevis. Peroneus longus continues further distally until it changes direction as it courses medially around the lateral aspect of the cuboid [137]. It inserts onto the plantar postero-lateral aspect of the cuneiform and the lateral base of the first metatarsal [138,139].

Peroneus longus functions to evert the foot and plantarflex the ankle joint during gait [137]. Peroneus longus is the second strongest evertor after peroneus brevis [140], although these muscles have nearly identical evertor moment arms acting on the talocrural and subtalar joints [134]. It has also been suggested that during gait, peroneus longus supports the medial and lateral arches of the foot, because the tendon passes along the plantar aspect of the foot [141]. Dynamic EMG indicates peroneus longus displays two bursts of activity during walking; the first burst occurs during contact phase and the second burst occurs during the midstance to propulsion phase [21] (Figure 4.2).

4.1.3 Tibialis anterior anatomy and function

Located in the anterior compartment of the leg, tibialis anterior is palpable around the antero-lateral edge of the tibia. It originates from multiple sites including; the lateral condyle of the tibia, the proximal two-thirds of the lateral tibial shaft, the anterior aspect of the interosseous membrane and the deep surface of the fascia cruris [137,142]. The tendon of tibialis anterior passes anterior to the talocrural joint beneath the superior and inferior bands of the extensor retinaculum [138]. A study of 290 feet indicated that tibialis anterior inserts onto the base of the first metatarsal and first cuneiform in approximately 90% of cadaveric specimens [143].

Tibialis anterior is a dorsiflexor of the ankle joint and inverter of the foot. Cadaveric studies of tibialis anterior have shown that the inversion moment arm acting on the talocrural and subtalar joint is 20 to 70% that of tibialis posterior [133,134], however in some specimens tibialis anterior has an eversion moment arm on these joints [134]. Dynamic EMG investigations indicate that tibialis anterior displays two bursts of activity during walking; the first burst occurs just prior to toe contact and the second burst peaks during toe off and remains active throughout the swing phase [21]. The role of the first burst is to decelerate ankle joint plantarflexion after heel strike to avoid the forefoot slapping the ground. The second burst occurs late in propulsion to enable toe clearance in swing phase and during heel strike [144] (Figure 4.2).

4.1.4 Medial gastrocnemius anatomy and function

Medial gastrocnemius is the larger of the two gastrocnemii and is located in the posterior compartment of the leg. The origin arises from the posterior condyle and popliteal surface of the femur and from the area adjacent to the knee joint capsule [137]. The medial and lateral heads are joined by a broad aponeurosis approximately half way along the leg with the larger medial head extending slightly further distally. [137]. The deeper tendon of soleus merges with the gastrocnemii aponeurosis to form the Achilles tendon, which inserts onto the posterior aspect of the calcaneus. The gastrocnemii and soleus, collectively known as triceps surae, share a common action during gait. Electromyographic investigations show these muscles display a single burst during the propulsive phase of gait, which plantarflexes the talocrural joint and propels the body forward [21,144] (Figure 4.2). Cadaveric research indicates that triceps surae also has an inversion moment arm on the talocrural joint when the foot is in an everted position; however this becomes an eversion moment arm when the foot is in an inverted position [134].

4.2 Electromyography procedure

To evaluate dynamic muscle activity for the projects contained in this thesis, EMG was chosen as the most appropriate modality. The advantage of using EMG over other modalities for assessing muscle activation, such as magnetic resonance imaging (MRI) and ultrasound, is the ability to investigate muscle activity simultaneously with dynamic weightbearing tasks such as walking [145]. This allows interpretation of muscle activity specific to different phases of the gait cycle.

In the investigations described in this thesis, tibialis posterior and peroneus longus were recorded with bipolar fine wire intramuscular electrodes, as first described by Basmajian and Stecko [146]. Surface electrodes cannot be used to record elctromyographic activity from tibialis posterior and peroneus longus. In the case of tibialias posterior this is due to the deep location of this muscle within the leg. Whereas the elctromyographic signal from peroneus longus is susceptible to receiving cross-talk from surrounding muscles such as lateral gastrocnemius and tibialis anterior [147].

The wire electrodes were fabricated from 75 μm Teflon® coated stainless steel wire (A-M Systems, Washington, USA) with 1 mm of insulation stripped from the recording surface of the two wires. The electrode wires were inserted into a 23 gauge sterilised single-use hypodermic needle with the exposed electrode tips bent 3 mm and 5 mm to prevent the contact areas from touching during recording. The wire and electrode ensemble were packed into a surgical sterilisation pouch and sterilised in an autoclave prior to EMG testing.

Tibialis anterior and medial gastrocnemius EMG signals were recorded with surface electrodes (DE 3.1, Delsys Inc., Boston, USA). The electrodes used feature a double differential three-bar type configuration with 99.9% silver contact material and an inter-electrode distance of 10 mm. The application of surface electrodes followed the recommendations of SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) [147]. Generally, each muscle belly was marked with a permanent marker, swabbed with 70% alcohol solution and the overlying skin was shaved. The skin was then abraded with skin sandpaper to reduce the electrical impedance of the skin.

4.2.1 Anatomical positioning of electrodes

Optimal positioning of recording electrodes within or overlying muscles is critical in EMG [148]. Numerous studies have demonstrated that the neuromuscular characteristics surrounding musculotendonous junctions can significantly alter EMG signal amplitude estimations [147-151]. For example, when surface electrodes are placed over an innervation zone (i.e. where motor nerve terminations and muscle fibers are connected), the amplitude signal will be minimal [152]. There are established standards in the literature that designate specific regions of muscle that provide optimal signal acquisition [147-151]. Accordingly, electrodes were positioned away from the location of innervation zones based on published recommendations [147-151]. The following list outlines the locations for the muscles that were evaluated:

(i) Tibialis posterior – the intramuscular electrode was inserted at a distance of approximately 50% between the popliteus cavity to the medial malleolus [153] (Figure 4.3)

(ii) Peroneus longus – the intramuscular electrode was inserted at approximately 20% of the distance from the head of fibula to the lateral malleolus, starting from the head of fibula [153]

(iii) Tibialis anterior – the surface electrode was placed at approximately 20% of the distance from the tuberosity of the tibia to the inter-malleoli line, starting from the tuberosity of the tibia [147]

(iv) Medial gastrocnemius – the surface electrode was placed at approximately 25% of the distance from the medial side of the popliteus cavity to the calcaneal tubercle [147]

Figure 4.3. Medial view of the right leg with the needle in position.

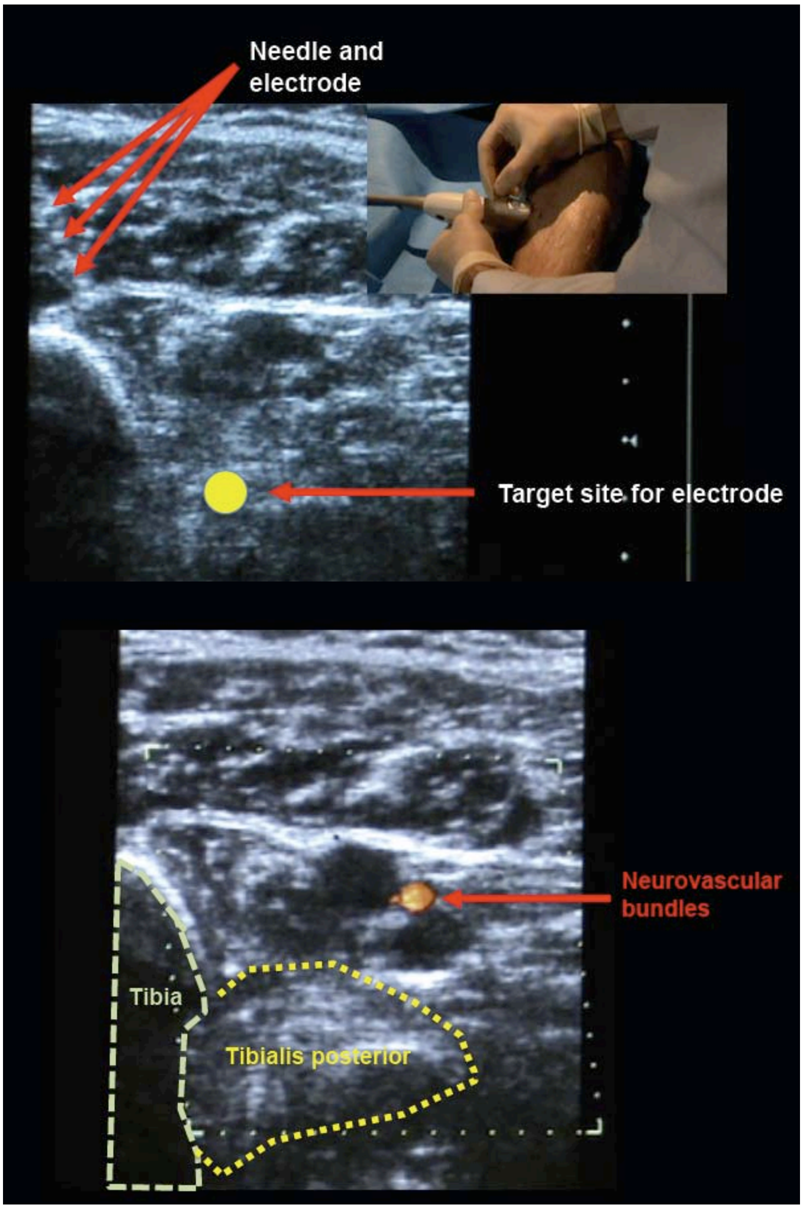
4.2.2 Intramuscular insertion procedure for tibialis posterior

There are two anatomical approaches for inserting intramuscular EMG electrodes into the tibialis posterior muscle belly; (i) posterior-medial, and (ii) anterior. A video demonstration of both approaches can be viewed via downloadable media (http://www.jfootankleres.com/imedia/3898488712749379/supp1.mov) and (http://www.jfootankleres.com/imedia/1348843031274937/supp2.mov) [132]. The posterior insertion involves guiding the electrode posterior to the tibia at a distance mid-way between the popliteus cavity to the medial malleolus. Caution is required not to penetrate the great saphenous vein or posterior neurovascular bundle. The anterior insertion involves guiding the electrode through tibialis anterior and the interosseous membrane avoiding the deep anterior neurovascular bundle.

When selecting the anterior or posterior insertion approach, two key issues to consider are safety and dynamic stability of the electrode. Cadaveric and MRI studies have shown the anterior approach provides a larger safety window when inserting electrodes, as there is a larger distance between osseous structures and neurovascular bundles compared to the posterior approach [25,26]. However, during piloting and preparation for previous tibialis posterior EMG work [12], we have found the anterior approach to be unstable during walking. The most frequent problem is retraction of the electrode tips from tibialis posterior through the interosseous membrane into tibialis anterior. Therefore, the posterior approach was used for the experiments in this thesis.

Historically, intramuscular insertion procedures were undertaken blindly without the aid of assisted imaging techniques. Recent advances in diagnostic ultrasonography have improved the accuracy of intramuscular electrode insertion visualisation of the target zone and key structures. Ultrasound imaging facilitates real-time observation of the insertion and identification of neurovascular bundles and anatomical variants (Figure 4.4).

Figure 4.4. Two ultrasound screenshots show a cross-sectional view of the lower leg from the same person. Key anatomical landmarks to consider with the needle insertion for tibialis posterior are shown. The top of each screenshot represents the level of the skin, and the needle has entered the leg in an approximately anterior posterior direction.



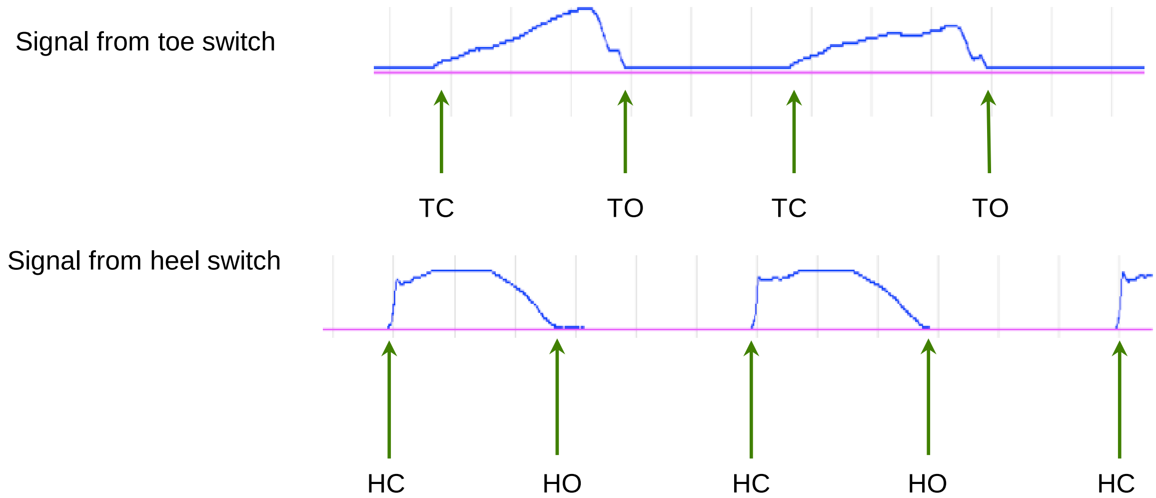
Accordingly, the insertion of intramuscular electrodes for the EMG experiments in this thesis was undertaken with ultrasound guidance (Sonoline Elegra 256 Advanced, Siemens, Germany). The position of the electrodes within the muscles (tibialis posterior and peroneus longus) was confirmed using a muscle stimulator

(ES-320, ITO Company Ltd, Tokyo, Japan). When flexion of the lesser digits occurred, the electrodes were considered to have moved into the flexor digitorum longus muscle, subsequently the electrodes were retracted and the insertion procedure was repeated with another needle and electrode.

4.2.3 Reference for temporal characteristics of the gait cycle

The temporal characteristics of the walking cycle were recorded using circular force sensitive resistors (footswitches) with a diameter of 13 mm (Model: 402, Interlink Electronics, California, USA). These were placed on the plantar surface of the interphalangeal joint of the hallux and the most posterior plantar aspect of the calcaneus to record the timing of heel contact, toe contact, heel off and toe off (Figure 4.5).

Figure 4.5. Screen-shot showing the heel and toe switch signals (EMGworks® 3.7 analysis, Delsys Inc., Boston, USA). HC – heel contact, HO – heel off, TC – toe contact, TO – toe off.



4.2.4 Electromyographic processing

The highest frequency components of the EMG signal are reported to be around 400-500 Hz [154]. It has been shown that EMG parameters such as waveform shape may require greater sampling rates than twice the Nyquist rate[3] for accurate interpretation [155]. Furthermore, EMG sampled below half the Nyquist rate is likely to result in a condition called ‘aliasing’ (i.e. poor temporal and amplitude representation of the signal) [156]. Therefore, raw EMG signal was passed through a differential amplifier with a gain of 1000 and a sampling frequency of 2 kHz. A band pass filter (built into the amplifier; Delsys Inc., Boston, USA) of 20–2000 Hz was applied to the raw electrical input of the intramuscular electrodes and 20–450 Hz for the surface electrodes (Figure 4.6)

All EMG data from the MVICs and walking trials were full wave rectified and low pass filtered at a cut off frequency of 6 Hz through a 4th order Butterworth filter with phase lag (Figure 4.6 and 4.7).

The footswitch data were used to calculate and time-normalise each gait cycle starting with heel strike (0%) and ending with ipsilateral heel strike (100%) (Figure 4.5 and 4.7).

The procedures undertaken to normalise EMG amplitude are outlined in Chapters 5, 6 and 7.

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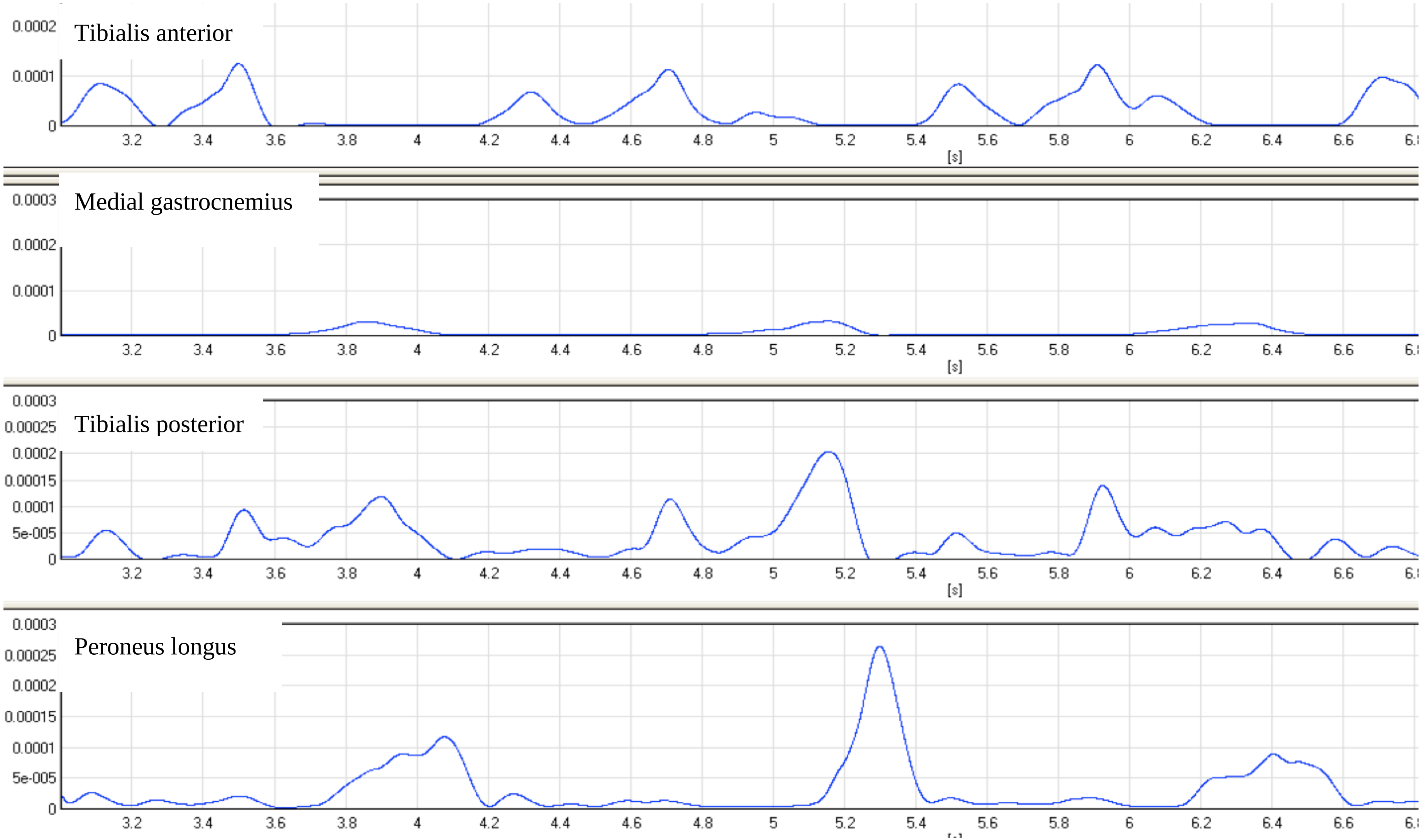
[3] The Nyquist rate is the minimum sampling rate required to avoid aliasing. It is equal to twice the highest frequency contained within the signal.

Figure 4.6. Screenshot showing rectified and filtered EMG signals for a single participant walking over a 9m walkway (the footswitch signals have been removed for clarity) (EMGworks® 3.7 analysis, Delsys Inc., Boston, USA). The vertical axes indicate EMG amplitude (volts), the horizontal axes indicate time (seconds).

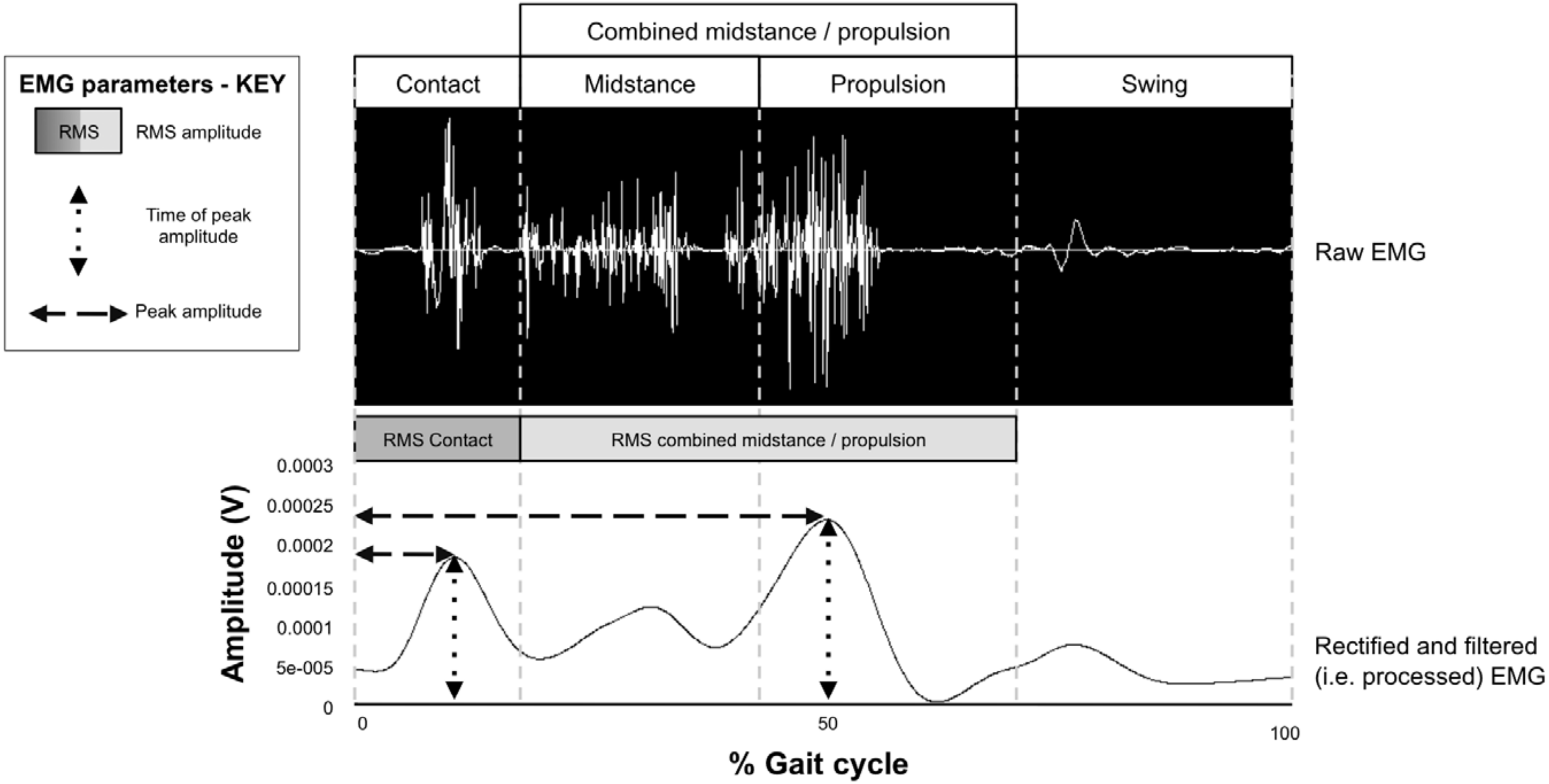


4.3 Electromyographic analysis

Two consecutive strides (i.e. comprising three consecutive heel contacts from the ipsilateral limb) were analysed for each trial and averaged from the last four of six trials for each speed (i.e. four average gait cycles derived from eight ipsilateral steps). These strides were selected for analysis to ensure the participants' walking velocity were not accelerating or decelerating.

Three EMG parameters were analysed for each muscle, including: (i) time of peak amplitude; (ii) root mean square (RMS); and (iii) peak amplitude. These parameters have been utilised in previous single-session investigations (Figure 4.6). The following phases of the gait cycle were assessed and based on when these muscles are most active in normal-arched feet: *tibialis posterior* and *peroneus longus* – contact and combined midstance/propulsion phase; *tibialis anterior* – contact phase; *medial gastrocnemius* – combined midstance/propulsion phase [21]. Contact phase was defined as the period between heel contact and toe contact.
Midstance/propulsion phase was defined as the period between toe contact and toe off. These events were determined from the footswitches placed on the heel and interphalangeal joint of the great toe (Figure 4.7).

Figure 4.7. Raw and processed EMG for tibialis posterior from a single representative participant. MVIC – maximum voluntary isometric contraction.



Reliability of lower limb EMG during overground walking: a comparison of maximal- and sub-maximal normalisation techniques

Preface

Although there is consensus among several studies that *within-session* stability of intramuscular EMG signals from lower limb muscles during walking is very good [157-159], results from studies investigating *between-session* reliability are not convincing.

A literature search and recursive search of published reference lists identified only three studies that had investigated the between-session reliability of gait-related intramuscular EMG. Kadaba and colleagues [159] used bipolar wires to investigate lower limb muscles from 10 participants on four test days separated by one week. Variance ratios for between-session reliability were 0.561 to 0.671 among the five muscles, which the authors described as 'poor constancy' between-sessions. Bogey and colleagues [160] later investigated soleus EMG via bipolar intramuscular electrodes in 18 young adults. They were critical of previous work by Kadaba and colleagues [159] for not normalising the EMG signal to a reference contraction. Accordingly, Bogey and colleagues [160] analysed EMG amplitude measures at 1% epochs of the gait cycle and normalised the signal using a dynamic heel-rise test. They reported a mean variance ratio of 0.187 and concluded that wire EMG 'meets acceptable standards for use in clinical and research trials'. More recently, Murley and colleagues [21] reported substantial changes in normalised amplitude of tibialis posterior in three participants, tested two weeks apart, indicating high variability between sessions.

Therefore, the aim of this study was to determine the magnitude of error (i.e. absolute reliability) between two EMG sessions when assessing tibialis posterior and peroneus longus via intramuscular electrodes, and tibialis anterior and medial gastrocnemius via surface electrodes during walking. This would indicate the feasibility of conducting subsequent studies that required a repeated-test design (e.g. to evaluate the effect of foot orthoses over a time-series – where the foot orthoses are introduced and removed repetitively over time). A secondary aim was to explore the effect of different normalisation techniques on between-session reliability and between-participant variability.

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Reliability of lower limb electromyography during overground walking: A comparison of maximal- and sub-maximal normalisation techniques

George S. Murley [a,b,*], Hylton B. Menz [b], Karl B. Landorf [a,b], Adam R. Bird [a,b]

[a] Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, Victoria 3086, Australia
[b] Musculoskeletal Research Centre, Faculty of Health Sciences, La Trobe University, Bundoora, Victoria 3086, Australia

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ABSTRACT

The purpose of this study was to determine the reliability of investigating electromyography (EMG) of selected leg muscles during walking. Tibialis posterior and peroneus longus EMG activity were recorded via intramuscular electrodes. Tibialis anterior and medial gastrocnemius EMG activity were recorded with surface electrodes. Twenty-eight young adults attended two test-sessions approximately 15 days apart. Relative and absolute measures of reliability were calculated for EMG timing and amplitude parameters during specific phases of the gait cycle. Maximum contractions and sub-maximal contractions were obtained via maximum isometric voluntary contractions and a very fast walking speed, respectively. Time of peak EMG amplitude for all muscles displayed relatively narrow limits of random error. However, reliability of peak and root mean square amplitude parameters for tibialis posterior and peroneus longus displayed unacceptably wide limits of random error, regardless of the normalisation reference technique. Whilst some amplitude parameters for tibialis anterior and medial gastrocnemius displayed good to excellent relative reliability, the corresponding values for absolute error were generally large.

Timing and amplitude EMG parameters for all muscles displayed low to moderate coefficient of variation within each test session (range: 7–25%). Overall, between-participant variability was minimised with sub-maximal normalisation values. These results demonstrate that re-application of electrodes results in large random error between sessions, particularly with tibialis posterior and peroneus longus. Researchers planning studies of these muscles with a repeated-test design (e.g. to evaluate the effect of an intervention) must consider whether this level of error is acceptable.

1. Introduction

The systematic investigation of muscle function in gait has expanded rapidly since Duchenne and Trendelenburg linked the absence of functional hip abductors with specific gait abnormalities (Duchenne, 1883; Trendelenberg, 1895). However, two centuries later, and despite modern digital computerised technology, there remains conjecture regarding the reliability (and therefore validity) of electromyography (EMG) in gait research.

The main source of measurement error is frequently attributed to the process of recording the EMG signal—either via surface or intramuscular electrodes. While surface electrodes are reported to provide adequate reliability for assessing within- and between-session lower limb EMG activity (Kadaba et al., 1985, 1989; Jacobson et al., 1995; Bogey et al., 2003), their major shortcoming is they cannot record from deep muscles, which requires the use of intramuscular needles or wire electrodes. However, needle and fine-wire electrodes are not without their problems; they can retract into neighbouring muscles during contraction (Jonsson and Komi, 1973), and there is concern about the reliability of the EMG signal.

Although there is consensus among several studies that *within-session* stability of intramuscular EMG signals from lower limb muscles during walking is very good (Kadaba et al., 1985; Jacobson et al., 1995; Franettovich et al., 2008), results from studies investigating *between-session* reliability are not convincing (Kadaba et al., 1989; Bogey et al., 2003; Murley et al., 2009b). Accordingly, there is a lack of evidence to demonstrate acceptable standards for between-session reliability of intramuscular-derived EMG recordings of various lower limb muscles. As such, the validity of investigating changes in muscle activity over time using intramuscular EMG measurements is questionable—this is a fundamental issue for EMG-related research.

In addition, several review articles over the last decade have called for researchers to report more rigorous reliability statistics

* Corresponding author at: Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, Victoria 3086, Australia. Tel.: +61 3 9479 5776; fax: +61 3 9479 5768.
E-mail address: g.murley@latrobe.edu.au (G.S. Murley).

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(Nevill and Atkinson, 1997; Atkinson and Nevill, 1998; Hopkins, 2000). Relative measures of reliability dominate gait-related EMG literature, but few studies report absolute measures of reliability (e.g. 95% limits of agreement). Therefore, there is also a lack of evidence relating to the magnitude of systematic bias and random error from intramuscular EMG recorded during gait on different test days. This represents a significant obstacle for study planning, particularly for studies assessing the effect of an intervention on deep muscle activity over repeated sessions.

With these issues in mind, the primary aim of this study was to determine the magnitude of error (i.e. absolute reliability) between two EMG sessions when assessing: tibialis posterior and peroneus longus (via intramuscular electrodes), and tibialis anterior and medial gastrocnemius (via surface electrodes) during walking. The secondary aim was to explore the effect of different normalisation techniques on between-session reliability and between-participant variability.

2. Methods

2.1. Participants

Twenty-eight young adults were recruited into the study after providing informed consent (Table 1). They had no recent musculoskeletal symptoms and were without neuromuscular disease. To recruit participants with only normal-arched feet, a foot screening protocol was developed that included both clinical and radiographic measures of foot posture (Murley et al., 2009a). This foot screening protocol was derived from normative foot posture values for the arch index, navicular height and two angular measures obtained from radiographs. Ethical approval was obtained from the La Trobe University Human Ethics Committee (Ethics ID: FHEC06/205).

Participants attended two EMG testing sessions separated by approximately 2 weeks (Table 1)—this period allowed sufficient time for residual muscle soreness to resolve prior to the re-test session. We acknowledge that intramuscular haematoma formation may still be present well beyond 2 weeks (Lynch et al., 2008), however the effect of haematoma on muscle function and signal contamination is unclear. Significant haematoma formation following intramuscular needle examination is rare, however, even when participants are taking anticoagulant or antiplatelet therapy (Lynch et al., 2008).

2.2. Instrumentation

Bipolar fine-wire intramuscular electrodes were used to record the EMG signals from tibialis posterior and peroneus longus. The electrodes were fabricated from 75 μm Teflon® coated stainless steel wire (A-M Systems, Washington, USA) with 1 mm of insulation stripped to form the recording surface of the two wires. The electrode wires were inserted into a single-use 23-gauge hypodermic needle with the exposed electrode tips bent 3 and 5 mm (to prevent the contact areas from touching during recording) and were sterilised prior to use. The process of fine-wire electrode construction and positioning of wires *in vivo* was undertaken in accordance with previous work (Murley et al., 2009b).

Tibialis anterior and medial gastrocnemius EMG signals were recorded with the use of DE-3.1 surface electrodes (Delsys Inc., Boston, USA). The electrodes featured a double differential 3-bar type configuration with a 99.9% silver contact material and an inter-electrode distance of 10 mm. The application of surface electrodes followed the recommendations of SENIAM—Surface Electromyography for the Non-Invasive Assessment of Muscles (Hermens et al., 1999).

The temporal characteristics of the walking cycle were measured using circular force-sensitive resistors (footswitches), with a diameter of 13 mm (Model: 402, Interlink Electronics, California, USA).

2.3. Data collection protocol

During testing, participants were instructed to walk at two self-selected walking speeds whilst barefoot: (i) at their usual comfortable walking speed and (ii) as fast as possible without running (i.e. as though they were late for an appointment), referred to as the "very fast" walking speed (Latt et al., 2008). In our study, the participants' self-selected speeds for the 'normal' and 'very fast' walking conditions were established during a warm-up period from two trials along a 9 m walkway. Six trials were recorded during testing and any trial exceeding $\pm 5\%$ of the average warm-up speed was excluded, with the trial being repeated.

2.3.1. EMG normalisation techniques

Two reference contractions were adopted for normalising EMG amplitude parameters across each data collection session: (i) maximum isometric voluntary contractions (MVICs) and (ii) dynamic normalisation comprising the 'very fast' self-selected walking speed. At the completion of each testing session, three MVICs for each muscle were undertaken as described in earlier work (Murley et al., 2009b). These comprised a gradual and continuous 2-s build-up followed by a maximum 2-s effort. Three consecutive maximum efforts were separated by a 1-min recovery period. A 600 ms window in the middle of the 2-s recording period was used to calculate average root mean square (RMS) from three trials (similar to the 500 ms window length assessed by a previous reliability study assessing intramuscular EMG) (Bogey et al., 2003). The normalisation value taken from the very fast walking condition was derived from the equivalent parameter of the normal walking speed. For example, the peak EMG amplitude for tibialis posterior during stance phase was normalised by the peak EMG amplitude obtained from the stance phase of the very fast walking speed (Fig. 1).

2.3.2. EMG processing

Two consecutive strides (i.e. comprising three consecutive heel contacts from the ipsilateral limb) were analysed for each trial and averaged from the last four of six trials for each speed (i.e. four average gait cycles derived from 8 ipsilateral steps). Three EMG parameters were analysed for each muscle: (i) time of peak amplitude; (ii) root mean square (RMS); and (iii) peak amplitude. These parameters have been utilised in previous single-session investigations (Fig. 1) (Murley and Bird, 2006; Murley et al., 2009b). The following phases of the gait cycle were assessed based on when each muscle is most active in normal-arched feet (Murley et al., 2009b): contact and combined midstance/propulsion phase for tibialis posterior and peroneus longus; contact phase for tibialis anterior; and combined midstance/propulsion phase for medial gastrocnemius.

2.4. Statistical analysis

Skewness and kurtosis values were used to evaluate the distribution of data and log-transformation was performed when either test- or re-test parameters were not normally distributed. Paired *t*-tests were performed to assess for systematic differences between test- and re-test sessions, with p values less than 0.05 considered significant. Between-session reliability was evaluated with relative measures of reliability including intra-class correlation coefficients (ICC), coefficient of variation (CoV); and absolute measures of reliability including limits of agreement (LoA) or ratio LoAs. Test and retest values were assessed for heteroscedasticity by correlating the absolute differences (i.e. measurement error) with the measurements for each variable. Heteroscedasticity is present when measurement error increases with the size of the measured variable (Atkinson and Nevill, 1998). Ratio LoAs were calculated when heteroscedasticity of the difference scores was reduced with log-transformation (Nevill and Atkinson, 1997). In this case, the ratios were derived from the 'antilog' of the log-transformed LoA (Nevill and Atkinson, 1997). Within-session variability of the different normalisation techniques was assessed using CoV calculations. Reliability increases with higher ICCs (i.e. >0.9), lower CoV values (i.e. $<10\%$), and narrower LoAs (Atkinson and Nevill, 1998).

Table 1
Participant anthropometric characteristics and duration between test and re-test sessions.

	N	Participant anthropometric characteristics		
		Age (years) mean ± SD	Height (cm) mean ± SD	Weight (kg) mean ± SD
Males	13	26.2 ± 7.4	177.5 ± 5.9	77.7 ± 10.5
Females	15	20.6 ± 2.3	162.9 ± 7.8	62.7 ± 13.0
Total	28	23.2 ± 5.9	169.7 ± 10.0	69.7 ± 14.0
Duration between test sessions (days) mean ± SD	15.1 ± 10.7; median 14; range (7–56)			

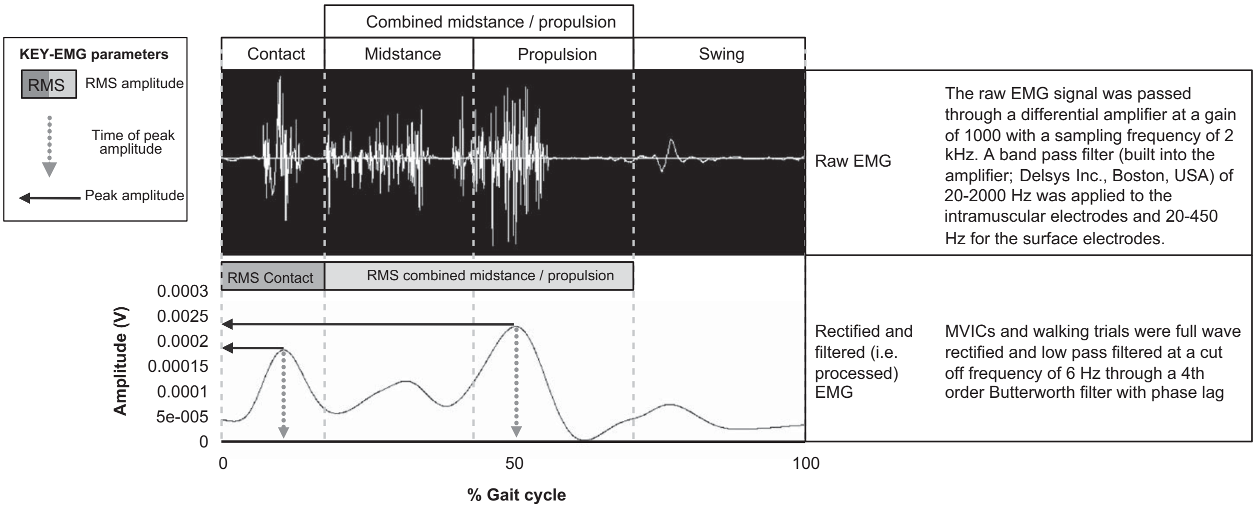

Fig. 1. A single gait cycle showing raw and processed EMG for tibialis posterior from a single participant. Time of peak amplitude, peak amplitude and RMS amplitude (root mean square) were derived from the linear envelope (processed curve).

3. Results

Out of the initial 30 participants recruited, two were not included in the final analysis. One participant was excluded prior to data processing as they expressed significant discomfort caused by the intramuscular electrodes during walking. Another participant did not attend the re-test session due to a sporting injury. Thus, the statistical analysis was based on twenty-eight participants' test and re-test values.

3.1. Walking speed within- and between-sessions

Random error for differences between test- and re-test normal walking speed was very low and there was no significant systematic difference in normal walking speed (Table 2).

3.2. Within-session trial-to-trial variability and effect of normalisation on between-participant variability

Table 3 shows the EMG parameter that displayed lowest trial-to-trial variability was the time of peak amplitude, with CoVs among the four muscles ranging from 6 to 17%. When comparing the within-session gait and MVIC amplitude values (Table 3 and Table 4), lower variation was found for the raw MVICs followed by the raw very fast walking speed and raw normal speed, respectively. The range of CoV values among individual participants and all EMG parameters varied substantially (1 – 107%).

Fig. 2 shows that the sub-maximal normalisation values consistently displayed the lowest variability (relative to the overall mean) between-participants for RMS and peak amplitude. Tibialis posterior and peroneus longus EMG amplitude displayed less or equivalent between-participant variability with the un-normalised values compared to MVIC normalisation values within each session. The opposite was observed for tibialis anterior and medial gastrocnemius EMG amplitude, which displayed less between-participant variability with the MVIC normalisation values compared to the un-normalised values.

3.3. Reliability of EMG muscle activity between test- and re-test sessions

From the 72 EMG parameters evaluated for test and re-test reliability, one parameter displayed significant systematic bias from test to re-test (peak amplitude of peroneus during midstance/propulsion using a sub-maximum normalisation value [mean difference: 7.7%; ratio between test and re-test: 1.15]) (Table 5a).

3.3.1. Time of peak amplitude

For tibialis posterior, time of peak amplitude displayed moderate relative reliability with ICCs of 0.55 and 0.50 during contact and midstance, respectively. For absolute reliability, the ratio LoA indicated very good reliability for time of peak amplitude in contact phase (0.77–1.32) and moderate to good reliability during midstance (0.71–1.36) (Table 5b). The contextual meaning of the ratio LoA is explained in Section 4.

For peroneus longus, time of peak amplitude displayed poor to moderate relative reliability with ICCs between 0.23 and 0.58 during contact and midstance, respectively. For absolute reliability, the ratio LoA indicated very good reliability for contact phase (0.77–1.25) and poor to moderate reliability during midstance phase (mean difference $\pm$95% LoA: -0.10 ± 7.49) (Table 5b).

For tibialis anterior, time of peak amplitude displayed moderate relative reliability with an ICC of 0.56 during contact phase. For absolute reliability, the ratio LoA indicated moderate reliability for contact phase (0.64–1.57) (Table 5b).

For medial gastrocnemous, time of peak amplitude displayed poor relative reliability with an ICC of 0.20 during midstance phase. For absolute reliability, the mean difference $\pm$95% LoA indicated moderate reliability for midstance phase (0.14 ± 10.36) (Table 5b).

3.3.2. Peak and RMS amplitude

For tibialis posterior, the reliability of both amplitude parameters was poor to moderate, irrespective of the normalisation technique or phase of the gait cycle. The most reliable amplitude

Table 2
Trial-to-trial walking speed variability and between session reliability for walking speed.

N	Within-session variability CoV (%) for trial-to-trial walking speed		*N*	Between-session reliability (normal walking speed)[a] Mean ± SD walking speed for test and re-test sessions	
	Test session (range)	Re-test session (range)		Test mean ± SD	Re-test mean ± SD
30	2.3(3.7–4.7)	2.3(1.1–4.1)	28	1.10 ms ± 0.11	1.08 ms ± 0.10

Trial-to-trial walking speed variability and between session reliability for walking speed.

[a] Type (2.1) ICC 0.97 (95% CI: 0.93–0.99).

Table 3
Within-session variability of raw EMG for normal and very fast walking.

EMG variables **Gait phase** *Parameter*	Normal walking speed—raw EMG						Very fast walking speed—raw EMG					
	Test session 1 (CoV)			Test session 2 (CoV)			Test session 1 (CoV)			Test session 2 (CoV)		
	Mean %	SD %	Range %	Mean %	SD %	Range %	Mean %	SD %	Range %	Mean %	SD %	Range %
Tibialis posterior												
Contact												
TimePeak	**9**	8	2–26	**12**	8	3–31	**9**	7	3–28	**9**	5	1–23
PeakAmp	**20**	10	4–45	**20**	9	7–45	**19**	9	4–37	**21**	13	3–60
RMS	**19**	10	5–39	**23**	9	9–45	**18**	8	3–35	**20**	12	7–61
Mid/Prop												
TimePeak	**9**	7	1–21	**8**	6	1–20	**6**	6	1–23	**6**	5	1–17
PeakAmp	**25**	17	6–80	**25**	13	6–53	**19**	9	3–39	**22**	13	8–65
RMS	**18**	9	6–37	**20**	11	4–53	**18**	9	3–43	**18**	12	2–62
Peroneus longus												
Contact												
TimePeak	**17**	11	3–39	**21**	22	3–107	**16**	13	3–61	**16**	17	1–63
PeakAmp	**25**	14	4–58	**23**	13	9–57	**23**	14	5–66	**22**	13	7–55
RMS	**21**	13	5–51	**24**	11	5–58	**19**	8	3–33	**23**	12	5–53
Mid/Prop												
TimePeak	**7**	6	1–25	**7**	6	1–17	**5**	6	1–25	**5**	4	1–15
PeakAmp	**19**	11	3–43	**20**	15	2–50	**15**	8	4–38	**16**	10	3–37
RMS	**19**	12	6–58	**20**	12	4–45	**14**	8	1–36	**15**	9	4–44
Tibialis anterior												
Contact												
TimePeak	**15**	15	1–57	**18**	13	1–59	**14**	9	1–35	**12**	7	4–37
PeakAmp	**13**	7	2–30	**13**	7	3–39	**10**	6	1–29	**11**	5	5–26
RMS	**13**	7	2–30	**12**	6	4–30	**10**	5	1–25	**11**	6	3–24
Medial gastrocmenius												
Mid/Prop												
TimePeak	**10**	11	1–56	**7**	5	1–18	**6**	6	1–26	**6**	4	1–21
PeakAmp	**13**	8	2–34	**12**	6	1–30	**12**	6	2–27	**12**	8	1–29
RMS	**10**	9	2–53	**10**	6	1–27	**11**	9	2–52	**12**	9	2–36

CoV—coefficient of variation; Contact—contact period of gait cycle; Mid/Prop—combined midstance and propulsion period of gait cycle; TimePeak—time of peak amplitude; PeakAmp—peak EMG amplitude; RMS—root mean square amplitude.

Table 4
Trial-to-trial variability of MVICs across test and re-test sessions.

Muscles	**Test session 1 (CoV)**			**Test session 2 (CoV)**		
	Mean (%)	SD (%)	Range (%)	Mean (%)	SD (%)	Range (%)
Tibialis posterior	14	10	1–47	11	9	1–33
Peroneus longus	13	9	2–40	11	9	1–31
Tibialis anterior	7	5	1–24	8	6	1–27
Medial gastrocnemius	7	4	0–17	10	8	0–34

Coefficient of variation (CoV) for all participants and muscles. Each CoV is derived from 3 MVIC trials per test session.

parameter/phase occurred with sub-maximum normalisation of peak amplitude during contact phase, with an ICC of 0.46 and ratio LoA between 0.44 and 2.77 (Table 5a).

For peroneus longus, the reliability of both amplitude parameters was poor to moderate, irrespective of the normalisation technique or phase of the gait cycle. The most reliable amplitude parameter/phase occurred with sub-maximum normalisation of RMS amplitude during contact phase, with an ICC 0.56 and ratio LoA between 0.39 and 2.45. Systematic error for peak amplitude during the midstance/propulsion phase was statistically significant between-sessions (7.7% greater in re-test session, $p<0.05$) when normalised by the sub-maximum reference value (Table 5a).

For tibialis anterior, the reliability of both amplitude parameters was dependant on the normalisation techniques applied. Relative reliability was moderate to good for MVIC normalised values (ICC: 0.56–0.65), moderate for sub-maximum values (ICC: 0.34–0.56) and very good to excellent for un-normalised values (ICC: 0.85–0.88). The most reliable amplitude parameter/phase occurred with un-normalised RMS

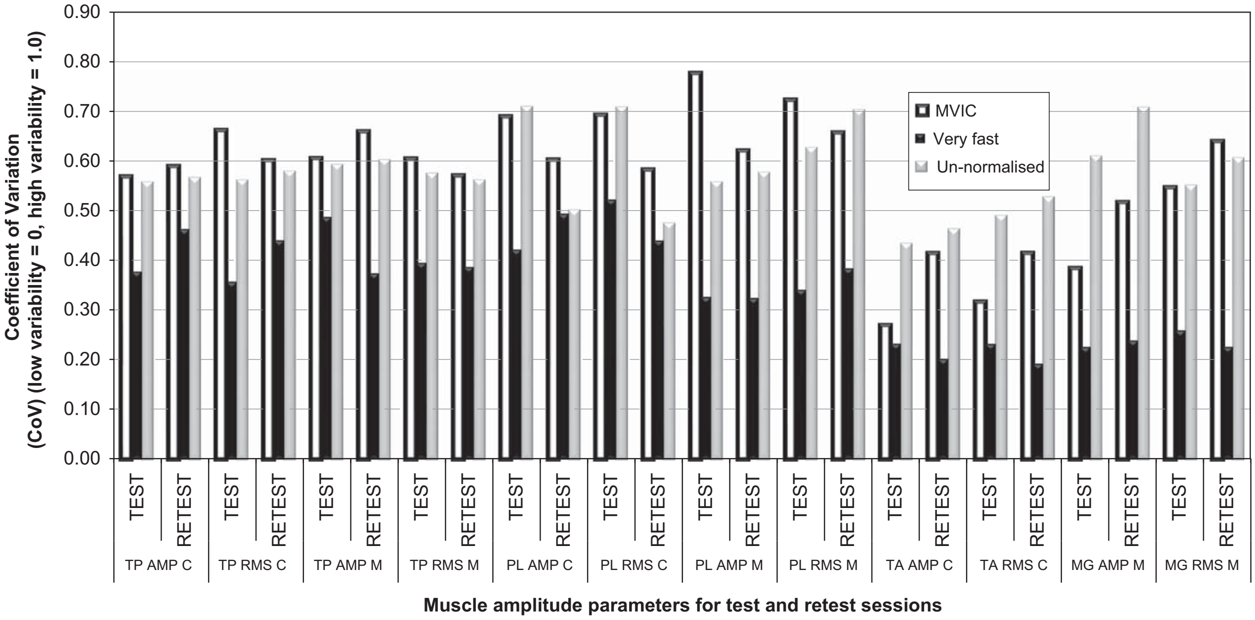

Fig. 2. Coefficient of variation for amplitude parameters among normalised and un-normalised EMG data for test- and re-test sessions. CoV—coefficient of variation; TP—tibialis posterior; PL—peroneus longus; TA—tibialis anterior; MG—medial gastrocnemius; AMP—peak amplitude; RMS—root mean square; C—contact phase; M—midstance/propulsion phase; MVIC—maximum voluntary isometric contraction.

amplitude, with an ICC 0.88 and ratio LoA between 0.61 and 1.47 (Table 5a).

For medial gastrocnemius, the reliability of both amplitude parameters was dependant on the normalisation values applied. Relative reliability was good to very good for MVIC normalised values (ICC: 0.61–0.84), poor for sub-maximum values (ICC: 0.08–0.19) and very good to excellent for un-normalised values (ICC: 0.78–0.86). The most reliable amplitude parameter/phase occurred with un-normalised RMS amplitude, with an ICC 0.86 and ratio LoA between 0.55 and 2.09 (Table 5a).

4. Discussion

The aim of this study was to determine the magnitude of between-session error of EMG in gait assessment. This was necessary in order to explore the feasibility of using a repeated-measures design for future studies. The absolute measures of reliability have been tabulated (Tables 5a and 5b) to provide researchers planning studies with pertinent data to determine whether intervention effects are larger than random error.

4.1. Within-session trial-to-trial variability

Variability between gait trials among the temporal and amplitude characteristic was low to moderate, with averaged CoV values between 5 and 25%. Some participants displayed a variation of 100% between gait trials—although this usually related to a small average measurement with an associated high standard deviation value. For example, one participant's mean time of peak amplitude for tibialis anterior occurred at 2% of the gait cycle with a standard deviation of 2%, which indicates relatively small variation in clinical terms. This is further supported by a recent study, despite only assessing five participants, that reported the mean change in tibialis posterior EMG amplitude with ankle taping was larger than the random standard error of measurement within a single session (Franettovich et al., 2008).

4.2. Between-session reliability—time of peak amplitude

The time of peak amplitude was the most reproducible EMG parameter among the four muscles, particularly for bursts during the contact phase of the gait cycle. However, it is essential that the reader interpret the absolute measures of reliability in the context of the EMG parameter and phase of the gait cycle. For example, the ratio LoA for tibialis posterior time of peak amplitude in contact phase was 0.77–1.32. This can be interpreted as meaning: if a participant's time of peak amplitude occurred at 10% of gait during initial testing, then on a subsequent testing approximately 2 weeks later, the time of peak amplitude could occur as early as 7.7% or as late as 13.2% of the gait cycle. This ratio LoA (i.e. 7.7–13.2%) suggests that to attribute a 'true change' (e.g. in response to an intervention) in tibialis posterior time of peak amplitude, the observed change must occur earlier than 7.7% or later than 13.2% of the gait cycle. Therefore, this parameter for assessing tibialis posterior may be appropriate for investigating the effect of interventions with neurogenic conditions that cause large changes in phasic activation of tibialis posterior, such as cerebral palsy (Michlitsch et al., 2006).

4.3. Between-session reliability—peak and RMS amplitude

The amplitude parameters for tibialis posterior and peroneus longus displayed unacceptable levels of error among the different methods of normalisation. For example, the ratio LoA for tibialis posterior peak amplitude during contact phase was 0.44–2.77 when normalised by the sub-maximum reference values. This can be interpreted as meaning: if a participant's peak amplitude was 90% of the sub-maximum reference value on initial testing, then on a subsequent testing session the true value for peak amplitude would lie between 39.6 and 249.3% of the sub-maximum reference value.

Table 5a
Relative and absolute reliability for all muscles and amplitude parameters among normalised and un-normalisation values.

	Maximum voluntary contraction normalisation					**Sub-maximum contraction normalisation**					**Un-normalised raw EMG**				
	Relative reliability		**Absolute reliability**			**Relative reliability**		**Absolute reliability**			**Relative reliability**		**Absolute reliability**		
EMG variables **Gait phase** *Parameter*	Type (2.1) ICC (95% CI)	Mean CoV %	Systematic bias (% mean difference)	Random error (95% LoA)	Ratio 95% LoA	Type (2.1) ICC (95% CI)	Mean CoV %	Systematic bias (% mean difference)	Random error (95% LoA)	Ratio 95% LoA	Type (2.1) ICC (95% CI)	Mean CoV %	Systematic bias (% mean difference)	Random error (95% LoA)	Ratio 95% LoA
Tibialis posterior															
Contact															
PeakAmp	0.13 (−0.26 to 0.48)	58.60	**	**	0.60–6.00	0.46[*1] (0.12–0.71)	41.68	**	**	0.44–2.77	0.21 (−0.19 to 0.54)	55.66	**	**	0.24–4.62
RMS	0.20 (−0.19 to 0.54)	78.82	**	**	0.26–6.23	0.42[*1], [#] (0.07 – 0.68)	*	−2.46	68.90	*	0.14 (−0.26 to 0.49)	54.35	**	**	0.22–4.45
Mid/Prop															
PeakAmp	−0.05 (−0.34 to 0.42)	64.27	**	**	0.19–9.82	0.34 (−0.03 to 0.65)	44.02	**	**	0.37 – 3.28	0.33 (−0.07 to 0.64)	59.44	**	**	0.25–4.87
RMS	0.09 (−0.30 to 0.46)	61.56	**	**	0.23–7.59	0.32[*1#] (−0.06 to 0.62)	*	3.27	52.87	*	0.24 (−0.16 to 0.58)	56.21	**	**	0.26–4.71
Peroneus longus															
Contact															
PeakAmp	0.53[*1] (0.17–0.77)	64.63	**	**	0.26–4.08	0.52[*1] (0.15–0.76)	45.34	**	**	0.38–3.16	0.08 (−0.32 to 0.46)	62.73	**	**	0.20–5.30
RMS	0.48[*1] (0.10–0.74)	78.34	**	**	0.20–3.48	0.56[*1] (0.21–0.78)	47.36	**	**	0.39–2.45	0.04 (−0.36 to 0.44)	59.97	**	**	0.20–4.88
Mid/Prop															
PeakAmp	0.62[*1] (0.31–0.80)	73.99	**	**	0.34–4.64	0.40[*1] (0.02–0.67)	33.05	**	**	[*2]0.52–2.57	0.32 (−0.11 to 0.65)	57.18	**	**	0.29–5.28
RMS	0.62[*1] (0.32–0.81)	70.67	**	**	0.31–4.77	0.33 (−0.06 to 0.62)	36.30	**	**	0.45–2.99	0.37[*1] (0.01–0.65)	66.20	**	**	0.26–5.83
Tibialis anterior															
Contact															
PeakAmp	0.56[*1] (0.22–0.78)	35.11	**	**	0.55–1.76	0.34[*1#] (−0.02 to 0.65)	*	2.78	33.12	*	0.85[*1] (0.70–0.93)	44.45	**	**	0.60–1.54
RMS	0.65[*1] (0.34–0.83)	37.02	**	**	0.55–1.70	0.56[*1] (0.21–0.78)	21.55	––	––	0.70–1.63	0.88[*1] (0.76–0.95)	50.52	**	**	0.61–1.47
Medial gastroccmenius															
Mid/Prop															
PeakAmp	0.61[*1] (0.30–0.80)	44.89	**	**	0.50–2.39	0.19 (−0.22 to 0.51)	23.07	**	**	0.58–1.85	0.78[*1] (0.58–0.90)	65.17	**	**	0.49–2.62
RMS	0.84[*1] (0.68–0.93)	57.47	**	**	0.47–2.12	0.08 (−0.30 to 0.44)	24.83	**	**	0.56–2.17	0.86[*1] (0.72–0.94)	57.42	**	**	0.55 – 2.09

CoV—coefficient of variation; LoA—limits of agreement;

** Ratio limits of agreement calculated for parameters with heteroscedasticity (i.e. correlation between the mean value and measurement error reduced with log-transformation).
* Heteroscedasticity not present, refer to adjacent values for systematic bias and random error.
*1 Statistically significant correlation ($p < 0.05$).
ICC calculated from raw data (i.e. log-transformation not required).
*2 Significant systematic bias detected ($p < 0.05$).

Table 5b
Between-session relative and absolute reliability for time of peak amplitude for all muscles.

EMG variables **Muscle_Gait phase** *Parameter*	**Time of peak amplitude** **Relative reliability** Type (2.1) ICC (95% CI)	Mean CoV %	**Absolute reliability** Systematic bias (% mean difference)	Random error (95% LoA)	Ratio LoA
TP_Contact	0.55*1 (0.23–0.76)	15.07	**	**	0.77–1.32
TP_MidProp	0.50*1 (0.17–0.74)	16.72	**	**	0.71–1.36
PL_Contact#	0.23 (−0.15 to 0.55)	*	−0.10	7.49	*
PL_MidProp	0.58*1 (0.27–0.78)	14.02	**	**	0.77–1.25
TA_Contact	0.56*1 (0.24–0.77)	22.66	**	**	0.64–1.57
MG_MidProp#	0.20 (−0.19 to 0.53)	*	0.14	10.36	*

CoV—coefficient of variation; LoA—limits of agreement.

** Ratio limits of agreement calculated for parameters with heteroscedasticity (i.e. correlation between the mean value and measurement error reduced with log-transformation).

* Heteroscedasticity not present, refer to adjacent values for systematic bias and random error.

*1 Statistically significant correlation ($p < 0.05$).

Tibialis posterior and peroneus longus EMG amplitude measures have emerged as potentially important biomechanical parameters from studies with a single-session design (Ringleb et al., 2007; Murley et al., 2009c). For example, in the presence of tibialis posterior tendon dysfunction and flat-arched foot posture, tibialis posterior EMG amplitude is significantly greater compared to normal controls (Gray and Basmajian, 1968; Keenan et al., 1991; Ringleb et al., 2007). However, our findings suggest that the magnitude of between-session error for tibialis posterior and peroneus longus RMS and peak amplitude is too large and renders these amplitude parameters unusable in a study adopting a repeated-measures design (i.e. where the intra-muscular electrodes are removed and inserted again on another test day). Therefore, RMS and peak EMG amplitude measures for tibialis posterior and peroneus longus cannot be reliably assessed following the implementation of a rehabilitation program or intervention; that is, there is too much between-session error.

Tibialis anterior and medial gastrocnemius EMG amplitude displayed significantly less error compared to tibialis posterior and peroneus longus. The ratio LoA for tibialis anterior RMS amplitude during contact phase was 0.61–1.47 when the signal was un-normalised. Again, this can be interpreted as meaning: if a participant's un-normalised RMS amplitude was 6.1 mV on initial testing, then on a subsequent testing session the true value for RMS amplitude would lie between 3.7 and 8.9 mV. Previous studies have reported very good to excellent between-session reliability for tibialis anterior and medial gastrocnemius based on relative measures of reliability (Winter and Yack, 1987; Kadaba et al., 1989). Indeed, we have shown in this study that ICCs of 0.86–0.88 can be obtained for RMS and peak amplitude parameters for these muscles. However, when absolute measures of reliability are applied to these muscles and parameters, there is evidence of unacceptably large error to the extent of rendering these parameters unusable in gait research involving repeated measures.

Several factors may confound the between-session reliability of intramuscular electrodes, including movement of the wire recording ends following muscle contraction; variations in intramuscular bleeding (causing impedance); and variation in the location of the insertion site with repeated insertions (Jonsson and Komi, 1973). Further development of intramuscular EMG assessment of tibialis posterior and peroneus longus is required to allow future studies to reliably assess amplitude characteristics with treatment interventions.

4.4. *Within-session between-participant variability*

EMG amplitude data derived from normal walking were normalised using the MVIC and sub-maximum reference values to explore the effect of normalisation on between-participant variability. The sub-maximum normalisation values produced consistently less variability between participants among all EMG amplitude variables, compared to MVIC and un-normalised values, respectively. One possible explanation for the greater variability with MVICs is that it is difficult to control and monitor the participants' effort or output. Although this finding of greater variability with MVICs has been reported elsewhere (Dankaerts et al., 2004; Bolgla and Uhl, 2007), there is a lack of data specific to intramuscular-derived EMG from deep muscles such as tibialis posterior. Normalisation methods that reduce inter-subject variability are important to EMG gait research, as small and potentially important changes from interventions can go undetected because of wide overlapping confidence intervals (Murley et al., 2009c).

The findings for this study need to be viewed in light of some limitations. Although the 'overall' reliability of the experimental protocol was assessed in this study, we cannot be certain what magnitude of error could be attributed to isolated components of the protocol, such as the precision of manufacturing the electrode wires or the accuracy of placing electrode wires *in vivo*. Furthermore, as the MVICs were conducted manually (i.e. without using an isokinetic dynamometer), some random variability may be attributed to the tester resisting the MVICs.

In summary, our results demonstrate that although EMG from the muscles tested is stable within a single session, the re-application of electrodes causes substantial absolute random error between-sessions for amplitude measures. Researchers planning studies of these muscles with a repeated-test design (i.e. to evaluate the effect of an intervention) must consider whether this level of error is acceptable in the context of the pathology and intervention under investigation.

Conflict of interest

There are no conflicts of interest.

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Foot posture influences the electromyographic activity of selected lower limb muscles during gait

Preface

The systematic review (Chapter 2) found limited evidence that having pronated foot posture influences EMG activity of tibialis posterior, tibialis anterior, flexor hallucis longus and peroneus longus compared to normal or supinated foot posture. However, it was difficult to generalise the findings from the six studies reviewed – largely due to significant clinical heterogeneity between the included studies. For example, Hunt and Smith [15] compared adults without a history of musculoskeletal disease to another symptomatic group with pronated foot posture. Whereas, Keenan and colleagues [8] investigated two groups of participants aged 40–71 years with moderate to long standing rheumatoid arthritis.

Another limitation with the studies included in the systematic review was the lack of reliable and valid methods used to classify participants' foot posture. For example, two studies [15,161] conducted a subjective visual assessment of foot posture, while only one study [8] used valid radiographic measurements to differentiate between different foot types. To address some of the shortcomings in foot posture classification in the biomechanical literature, the foot screening protocol (Chapter 3) was designed to provide a clear pathway for recruiting participants with normal- and flat-arched feet. We hypothesised that adopting a more systematic approach to classifying foot posture would assist in the identification of functional differences in EMG activity between foot types.

Therefore, the objective of the study in this chapter was to investigate the effect of foot posture on lower limb muscle activity in people during walking.

The study in this chapter has been published:

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Research

Foot posture influences the electromyographic activity of selected lower limb muscles during gait

George S Murley*[1,2], Hylton B Menz[2] and Karl B Landorf[1,2]

Address: [1]Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, Australia and [2]Musculoskeletal Research Centre, Faculty of Health Sciences, La Trobe University, Bundoora, Australia

Email: George S Murley* - g.murley@latrobe.edu.au; Hylton B Menz - h.menz@latrobe.edu.au; Karl B Landorf - k.landorf@latrobe.edu.au

* Corresponding author

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Abstract

Background: Some studies have found that flat-arched foot posture is related to altered lower limb muscle function compared to normal- or high-arched feet. However, the results from these studies were based on highly selected populations such as those with rheumatoid arthritis. Therefore, the objective of this study was to compare lower limb muscle function of normal and flat-arched feet in people without pain or disease.

Methods: Sixty adults aged 18 to 47 years were recruited to this study. Of these, 30 had normal-arched feet (15 male and 15 female) and 30 had flat-arched feet (15 male and 15 female). Foot posture was classified using two clinical measurements (the arch index and navicular height) and four skeletal alignment measurements from weightbearing foot x-rays. Intramuscular fine-wire electrodes were inserted into tibialis posterior and peroneus longus under ultrasound guidance, and surface EMG activity was recorded from tibialis anterior and medial gastrocnemius while participants walked barefoot at their self-selected comfortable walking speed. Time of peak amplitude, peak and root mean square (RMS) amplitude were assessed from stance phase EMG data. Independent samples *t*-tests were performed to assess for significant differences between the normal- and flat-arched foot posture groups.

Results: During contact phase, the flat-arched group exhibited increased activity of tibialis anterior (peak amplitude; 65 versus 46% of maximum voluntary isometric contraction) and decreased activity of peroneus longus (peak amplitude; 24 versus 37% of maximum voluntary isometric contraction). During midstance/propulsion, the flat-arched group exhibited increased activity of tibialis posterior (peak amplitude; 86 versus 60% of maximum voluntary isometric contraction) and decreased activity of peroneus longus (RMS amplitude; 25 versus 39% of maximum voluntary isometric contraction). Effect sizes for these significant findings ranged from 0.48 to 1.3, representing moderate to large differences in muscle activity between normal-arched and flat-arched feet.

Conclusion: Differences in muscle activity in people with flat-arched feet may reflect neuromuscular compensation to reduce overload of the medial longitudinal arch. Further research is required to determine whether these differences in muscle function are associated with injury.

Background

Human foot posture is highly variable among healthy individuals and ranges from flat- to high-arched [1]. While foot posture is strongly influenced by some systemic conditions, such as neurological and rheumatological diseases, there is emerging evidence that variations in foot posture among healthy individuals are associated with changes in lower limb motion [2,3], and in some cases, increased risk of lower limb injury [4,5]. The link between variations in foot posture and increased risk of lower limb injury may arise from abnormal muscle activity. For example, it has been suggested that the flat-arched foot relies on additional muscular support during gait [2], and that fatigue of these controlling muscles with exercise can result in the development of various injuries such as tibial stress fractures [6].

With this mind, we recently conducted a systematic review of studies that investigated the effect of foot posture on lower limb muscle activity during walking or running [7]. The review concluded that there is some evidence to indicate that pronated foot posture is associated with greater electromyography (EMG) amplitude for invertor muscles, such as tibialis posterior, and lower EMG amplitude for evertor muscles, such as peroneus longus, when compared to normal or supinated foot posture. However, these findings may not be generaliseable to the wider population because of highly selected samples. For instance, the best evidence to date that indicates tibialis posterior muscle activation is greater in flat-arched foot posture was reported by a study comprising older adults with long-standing rheumatoid arthritis [8]. Therefore, other than the early descriptive work of Gray and Basmajian in 1968 [9], it is unknown whether foot posture influences tibialis posterior muscle activation in adults without pain or dysfunction.

Another issue with previous studies is that strategies for classifying foot posture have infrequently included valid and reliable measurements. Several methods of classifying foot posture have been employed, including: the arch index [10], the arch ratio [11], radiographic alignment [8], two-dimensional video analysis [12] and subjective clinical observation [2,9]. Furthermore, only in the last decade has normative foot posture data for various clinical and radiological measurements been published [3,13-16]. Utilising these data, we recently developed a protocol for classifying foot posture based on both clinical and radiographic measurements [16]. We hypothesised that adopting a more systematic approach to classifying foot posture would assist in the identification of functional differences in EMG activity between foot types.

With these issues in mind, the objective of this study was to investigate EMG activity of tibialis posterior, peroneus longus, tibialis anterior and medial gastrocnemius in healthy adults with normal- and flat-arched foot posture.

Methods

Participants

Sixty adults aged 18 to 47 years were recruited to this study. Of these, 30 had normal-arched feet (15 male and 15 female) and 30 had flat-arched feet (15 male and 15 female). Participant characteristics are presented in Table 1. A foot screening protocol that included both clinical and radiographic measures to classify foot posture was used to recruit participants with normal- and flat-arched feet [16]. This protocol was derived from normative foot posture values for two clinical measurements (the arch index and navicular height) and four angular measurements obtained from antero-posterior and lateral x-rays (talus-second metatarsal angle, talonavicular coverage angle, calcaneal inclination angle and calcaneal-first metatarsal angle) [16]. To qualify for the normal-arched foot group, participants had either a normal arch index or navicular height measurement, and their four radio-

Table 1: Participant characteristics

	Foot posture groups	
	Flat-arch n = 30	**Normal-arch n = 30**
General anthropometric		
Gender ratio (female/male)	15/15	15/15
Age mean ± SD (years)	21.8 ± 4.3	23.6 ± 5.9
Height mean ± SD (cm)	171.0 ± 10.0	169.7 ± 9.7
Weight mean ± SD (Kg)	73.3 ± 15.50	69.9 ± 13.6
Left or right foot count[FC]	13 right/17 left	13 right/17 left
Clinical measurements		
AI mean ± SD	0.30 ± 0.07*	0.24 ± 0.04*
NNHt mean ± SD	0.18 ± 0.04†	0.27 ± 0.03†
Radiographic measurements		
CIA mean ± SD (degrees)	15.7 ± 4.5#	20.8 ± 3.5#
C1MA mean ± SD (degrees)	142.3 ± 6.0‡	132.8 ± 4.1‡
TNCA mean ± SD (degrees)	27.6 ± 9.0^	11.9 ± 8.1^
T2MA mean ± SD (degrees)	27.1 ± 10.1¥	13.0 ± 6.4¥
Walking velocity	1.21 ± 0.13**	1.10 ± 0.11**

AI -- arch index, NNHt -- normalised navicular height truncated, CIA -- calcaneal inclination angle, C1MA -- calcaneal first metatarsal angle, TNCA -- talo-navicular coverage angle, T2MA -- talus-second metatarsal angle. [FC]denotes the number of participants whose left or right foot was suitable for inclusion in their respective group (i.e. normal-arch or flat-arch).
Mean differences and 95% confidence interval (CI) expressed relative to normal-arch.
Statistically significant findings for comparisons listed below ($p < 0.001$):
* AI: mean difference 0.06, 95% CI 0.03 to 0.09
† NNHt: mean difference -0.09, 95% CI -0.11 to -0.08
#CIA: mean difference -5.13°, 95% CI -7.21 to -3.05°
‡C1MA: mean difference 9.47°, 95% CI 6.8 to 12.14°
^TNCA: mean difference 15.70°, 95% CI 11.28 to 20.12°
¥ T2MA: mean difference 14.08°, 95% CI 9.73 to 18.44°
** Walking speed: mean difference 0.11 ms, 95% CI 0.05 to 0.17 ms

graphic measurements were within a normal range. To qualify for the flat-arched group, participants had an arch index or navicular height measurement greater than two standard deviations from mean values obtained for the normal-arched group. Furthermore, their radiographic measurements were greater than 1 standard deviation from the mean values obtained for the normal-arched group for either the sagittal and or transverse plane measurements. Figures 1, 2 and 3 illustrate the clinical and radiographic measurements.

The participants were without symptoms of macrovascular disease (e.g. angina, stroke, peripheral vascular disease), neuromuscular disease, or any biomechanical abnormalities that affected their ability to walk. Ethical approval was obtained for the study from the La Trobe University Human Ethics Committee (Ethics ID: FHEC06/205) and the study was registered with the Radiation Safety Committee of the Victorian Department of Human Services. The x-rays were performed in accordance with the Australian Radiation Protection and Nuclear Safety Agency Code of Practice for the Exposure of Humans to Ionizing Radiation for Research Purposes (2005) [17].

Experimental protocol

Bipolar fine-wire intramuscular electrodes were used to record the EMG signal from tibialis posterior and per-

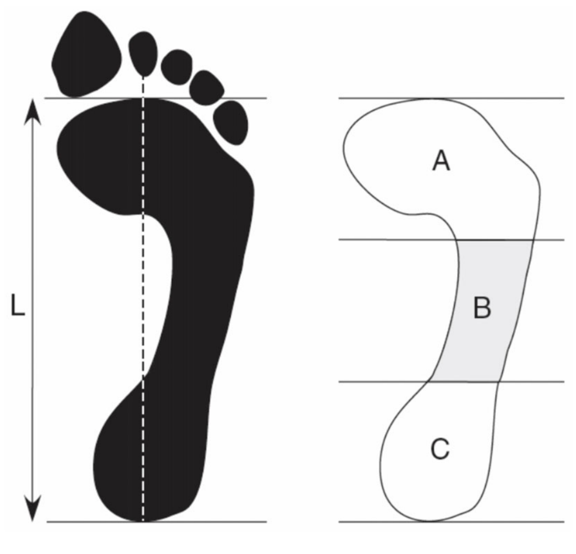

Figure 1
Footprint with reference lines for calculating the arch index. The length of the foot (excluding the toes) is divided into equal thirds to give three regions: A -- forefoot; B -- midfoot; and C -- heel. The arch index is then calculated by dividing the midfoot region (B) by the entire footprint area (i.e. Arch index = B/[A+B+C]).

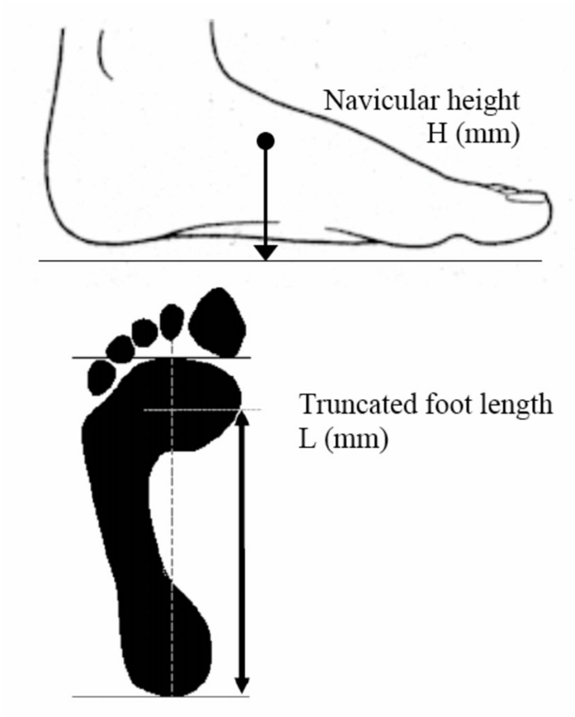

Figure 2
Calculating normalised navicular height truncated. The distance between the supporting surface and the navicular tuberosity is measured. Foot length is truncated by measuring the perpendicular distance from the 1st metatarsophalangeal joint to the most posterior aspect of the heel. Normalised navicular height truncated is calculated by dividing the height of the navicular tuberosity from the ground (H) by the truncated foot length (L) (i.e. Normalised navicular height truncated = H/L).

oneus longus. The electrodes were fabricated from 75 μm Teflon® coated stainless steel wire (A-M Systems, Washington, USA) with 1 mm of insulation stripped to form the recording surface of the two wires. The electrode wires were inserted into a 23 gauge sterilized single use hypodermic needle with the exposed electrode tips bent 3 mm and 5 mm to prevent the contact areas from touching during recording. The process of fine-wire electrode construction and positioning of wires in vivo was undertaken in accordance with previous work [14] (Additional file 1).

Tibialis anterior and medial gastrocnemius EMG was recorded with the use of DE-3.1 surface electrodes (Delsys Inc., Boston, USA). The electrodes featured a double differential 3-bar type configuration with a 99.9% silver contact material and an inter-electrode distance of 10 mm. The application of surface electrodes followed the recommendations of SENIAM [18].

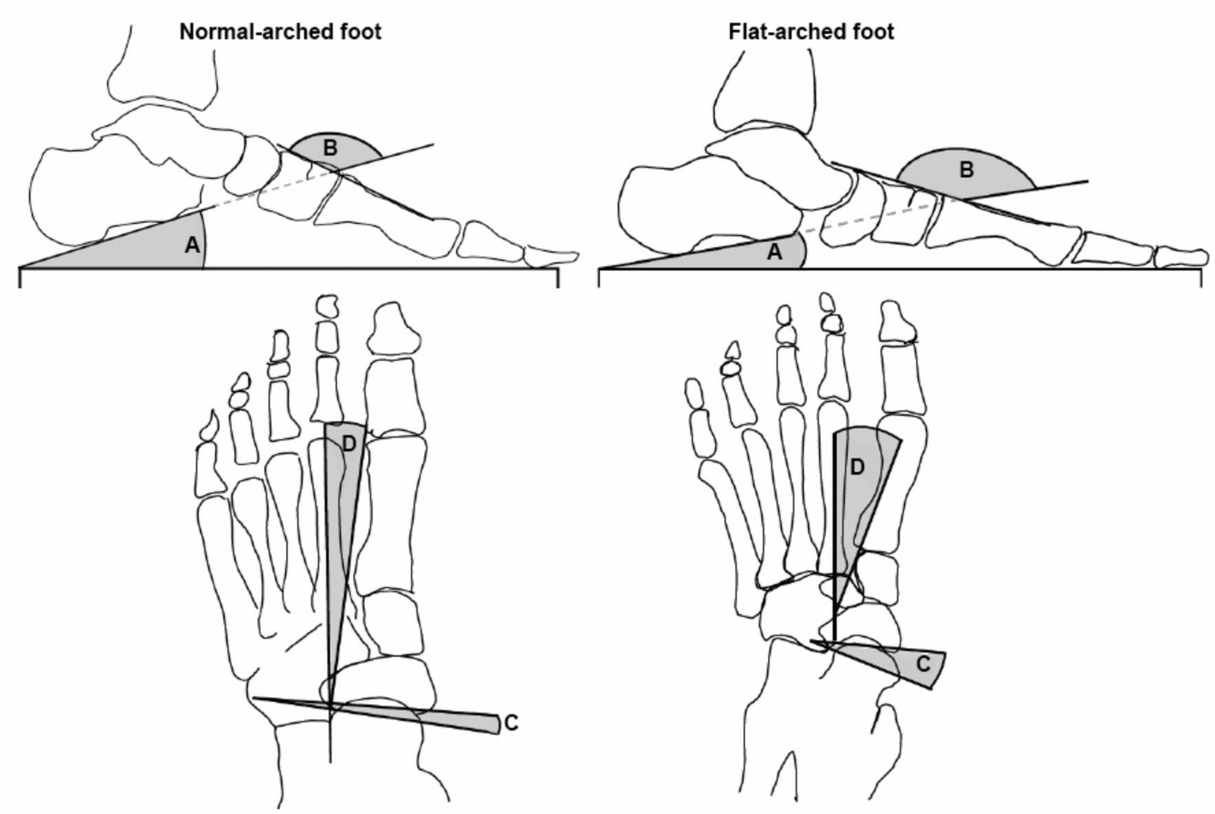

Figure 3
Traces from two representative participants illustrate x-ray angular measurements from normal (left) and flat-arched (right) foot posture. Lateral views (top) show: calcaneal inclination angle; calcaneal-first metatarsal angle; anterior posterior views (bottom) show: talonavicular coverage angle; talus second metatarsal angle. A - calcaneal inclination angle, B - calcaneal-first metatarsal angle, C - talo-navicular coverage angle, D - talus-second metatarsal angle. Angle A *decreases* with flat-arched foot posture; angle B, C and D *increase* with flat-arched foot posture, compared to the normal-arched foot posture.

The temporal characteristics of the walking cycle were measured using circular force sensitive resistors (footswitches) with a diameter of 13 mm (Model: 402, Interlink Electronics, California, USA). These were placed on the plantar surface of the interphalangeal joint of the hallux and the most posterior plantar aspect of the calcaneus to record the timing of heel contact, toe contact, heel off and toe off.

During testing, participants were instructed to walk at their self-selected walking speed, which was established following a warm-up period from two trials along a 9 m walkway. Six trials were recorded during testing, with any trial exceeding ± 5% of the average warm-up speed excluded and subsequently repeated.

Maximum voluntary isometric contractions (MVIC) were used for normalising EMG amplitude parameters. At the completion of each testing session, three MVICs for each muscle were undertaken comprised of a gradual and continuous 2 s build-up followed by a maximum 2 s effort. Each participant was instructed to perform a maximum contraction against the resistance of the tester and was given verbal encouragement while doing so. The resisted movements included; supination - tibialis posterior, pronation - peroneus longus, dorsiflexion - tibialis anterior, plantarflexion (knee extended) - medial gastrocnemius. The participant sat on a bench while performing the MVICs for tibialis posterior, tibialis anterior and the peroneal muscles. For the medial gastrocnemius MVICs, the participant sat on the floor with their back against a wall, to ensure the participant did not slide backward during the contraction.

Three consecutive maximum efforts were separated by a 1 min recovery period. A 600 ms window in the middle of the 2 s recording period was used to calculate average root mean square (RMS) from three trials.

Data processing

During the gait trials, the raw EMG signal was passed through a differential amplifier at a gain of 1000 with a sampling frequency of 2 kHz. A band pass filter (built into the amplifier; Delsys Inc., Boston, USA) of 20-2000 Hz was applied to the intramuscular electrodes and 20-450 Hz for the surface electrodes.

EMG data from the MVICs and walking trials were full wave rectified and low pass filtered at a cut off frequency of 6 Hz through a 4^{th} order Butterworth filter with phase lag. Data were analysed from the third or fourth stride depending on the quality of the footswitch signal. Two consecutive strides were analysed for each trial and averaged from the last four of six trials for each speed (i.e. four average gait cycles derived from eight ipsilateral steps). Three EMG parameters were analysed for each muscle, including: (i) time of peak amplitude; (ii) root mean square (RMS); and (iii) peak amplitude (Figure 4). These parameters have been utilised in previous single-session investigations [14,19,20]. The following phases of the gait cycle were assessed based on when these muscles are most active in normal-arched feet [14]: tibialis posterior and peroneus longus - contact and combined midstance/propulsion phase; tibialis anterior - contact phase; and medial gastrocnemius - combined midstance/propulsion phase.

Statistical analysis

The distribution of data was evaluated from skewness and kurtosis values and Levene's test for equality of variances. Independent samples *t*-tests were performed to assess for significant differences between the normal- and flat-arched groups for anthropometric characteristics, walking speed and EMG parameters with p values less than 0.05 considered significant.

Results

Participant characteristics

The normal- and flat-arched foot posture groups were matched for age, gender, height and weight, with no significant differences for any of these characteristics except for the clinical and radiographic measures of foot posture (Table 1). However, the self-selected comfortable walking speed of the flat-arched group was slightly greater than the normal-arched group (mean difference: 0.11 ms, 95% CI: 0.05 to 0.17, $p < 0.001$).

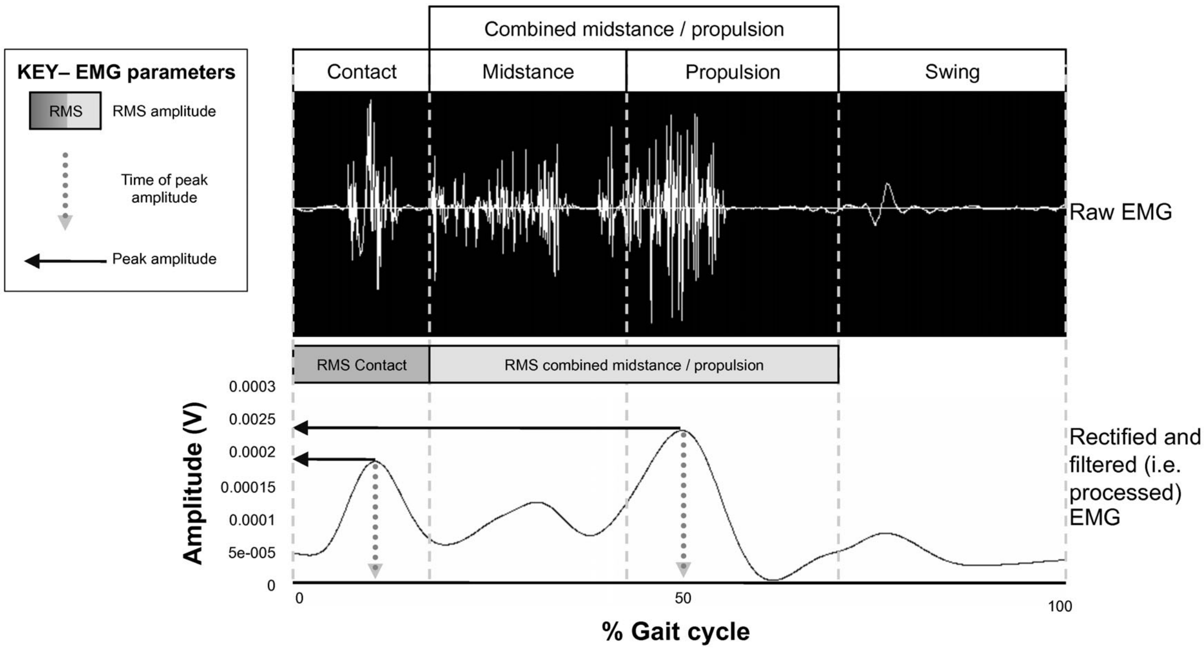

Figure 4
A single gait cycle showing raw and processed EMG for tibialis posterior from a single participant. Time of peak amplitude, peak amplitude and RMS amplitude (root mean square) were derived from the linear envelope (processed curve).

Effect of foot posture on muscle EMG activation

Comparisons of EMG variables between the normal- and flat-arched foot groups are presented in Table 2. Statistically significant differences in peak and RMS EMG amplitude were detected for tibialis posterior, peroneus longus and tibialis anterior. There were no significant differences in EMG time of peak amplitude.

Contact phase - heel contact to toe contact

For tibialis anterior, the flat-arched group exhibited increased peak EMG amplitude (mean difference: 19.0%; 95% CI: 11.2 to 26.9; $d = 1.3$; $p < 0.001$) and RMS amplitude (mean difference: 10.4%; 95% CI: 4.0 to 16.8; $d = 0.87$; $p = 0.002$), compared to the normal-arched group. For peroneus longus, the flat-arched foot group exhibited decreased peak EMG amplitude (mean difference: -12.8%; 95% CI: -25.1 to -0.5; $d = 0.48$; $p = 0.041$), compared to the normal-arched group (Figure 5). For tibialis posterior, the flat-arched foot group exhibited decreased peak EMG amplitude (mean difference: -14.3%; 95% CI: -29.1 to 0.4; $d = 0.51$; $p = 0.058$) compared to the normal arched group, although this finding did not reach statistical significance (Figure 5).

Midstance/propulsion phase - toe contact to toe-off

For peroneus longus, the flat-arched foot group exhibited decreased peak EMG (mean difference: -13.7%; 95% CI: -26.1 to -1.4; $d = 0.58$; $p = 0.030$), compared to the normal-arched group (Figure 5). For tibialis posterior, the flat-arched group exhibited increased peak EMG amplitude (mean difference: 26.5%; 95% CI: 4.2 to 48.7; $d = 0.69$; $p = 0.021$) and RMS amplitude (mean difference: 16.4%; 95% CI: 3.6 to 29.1; $d = 0.68$; $p = 0.013$), compared to the normal-arched group (Figure 5). No significant differences between groups were detected for medial gastrocnemius.

Discussion

The objective of this study was to investigate the effect of flat-arched foot posture on the EMG activity of selected leg muscles. During comfortable walking, participants in the flat-arched foot group functioned at a significantly greater percentage of their maximum amplitude for tibialis posterior during midstance/propulsion phase, compared to participants in the normal-arched group (peak amplitude, 86 versus 60% of MVIC; RMS amplitude, 48 versus 31% of MVIC). Similar trends have been reported by earlier studies comparing these foot types [8,9], however these studies did not report 95% confidence intervals for the percentage difference or effect size calculations, making it difficult to assess the precision and the magnitude of the differences observed [7]. Effect sizes for the differences observed in peak and RMS for tibialis posterior amplitude were 0.68 and 0.69 respectively, representing moderate

Table 2: Effect of foot posture on all EMG variables

Muscle	Phase of gait cycle	EMG parameter	% mean difference ^	95% CI	Effect size #	p value (2-tailed)
Tibialis posterior	Contact	TimePeak	0.1	0.0 to 1.7	0.52	0.051
		PeakAmp	-14.3	-29.1 to 0.4	0.51	0.058
		RMS	-7.8	-18.4 to 2.7	0.39	0.144
	Mid/Prop	TimePeak	0.0	-3.8 to 3.7	0.01	0.980
		PeakAmp	26.5*	4.2 to 48.7	0.69	0.021*
		RMS	16.4*	3.6 to 29.1	0.68	0.013*
Peroneus longus	Contact	TimePeak	1.6	0.0 to 3.2	0.51	0.057
		PeakAmp	-12.8*	-25.1 to -0.5	0.48	0.041*
		RMS	-6.6	-13.9 to 0.6	0.48	0.075
	Mid/Prop	TimePeak	3.3	-0.3 to 6.9	0.47	0.079
		PeakAmp	-20.0	-42.9 to 2.9	0.46	0.086
		RMS	-13.7*	-26.1 to -1.4	0.58	0.030*
Tibialis anterior	Contact	TimePeak	0.1	-0.7 to 0.9	0.09	0.737
		PeakAmp	19.0*	11.2 to 26.9	1.3	<0.001*
		RMS	10.4*	4.0 to 16.8	0.87	0.002*
Medial oastrocnemius	Mid/Prop	TimePeak	0.4	-1.8 to 2.7	0.10	0.715
		PeakAmp	2.7	-15.4 to 20.7	0.12	0.766
		RMS	7.2	-12.3 to 16.9	0.22	0.753

Contact -- contact period of gait cycle; Mid/Prop -- combined midstance and propulsion period of gait cycle; TimePeak -- time of peak amplitude; PeakAmp -- peak EMG amplitude; RMS -- root mean square amplitude; ^ relative to normal-arch foot group; CI -- confidence interval;# Cohen's d calculation;
* statistically significant independent sample t-test ($p < 0.05$)

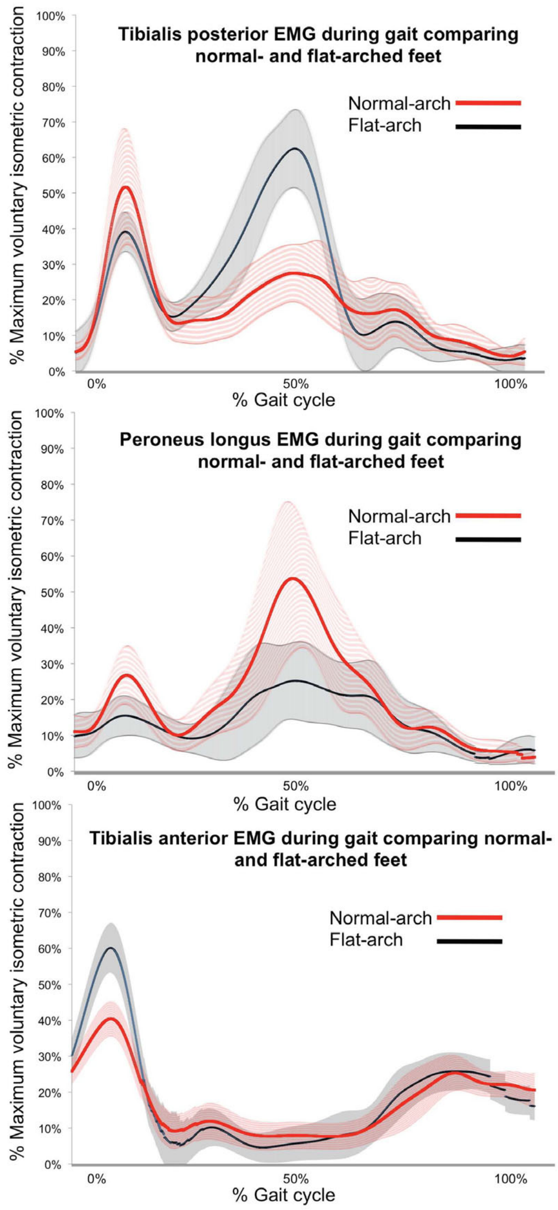

Figure 5
Ensemble averaged EMG curves for tibialis posterior, peroneus longus and tibialis anterior for 30 participants with normal-arch and 30 participants with flat-arch feet. The curves differ slightly to the actual results (Table 2), as these curves are derived from a single gait cycle for each participant to illustrate the main findings. Solid lines -- mean amplitude; shaded area surrounding solid line -- 95% confidence interval. Significant differences are generally indicated where 95% confidence intervals separate between groups. HC - heel contact.

differences in muscle activity. Despite the issue of random variability for tibialis posterior EMG amplitude during gait [14,20], our results provide strong evidence to indicate that tibialis posterior is working harder (i.e. as a percentage of a maximum contraction) during midstance/propulsion in participants with flat-arched feet, compared to those with normal-arched feet.

One explanation for our findings is that the medial longitudinal arch and supportive structures (e.g. ligaments) of a flat-arched foot may undergo greater loading during walking, compared to the normal-arched foot. Greater loading of the medial arch would require greater work from tibialis posterior to protect the arch structures from excessive tissue stress and injury. While cadaveric research has shown an increased loading of the foot's medial structures with simulated tibialis posterior tendon dysfunction [21], it is also possible that these events can occur in reverse, that is, the flat-arched foot may place a greater demand on tibialis posterior. This mechanism is further supported by our findings for peroneus longus.

In contrast to tibialis posterior, participants in the flat-arched group functioned at a significantly lower percentage of their maximum amplitude for peroneus longus during contact phase and midstance/propulsion phase, compared to participants in the normal-arched group (peak amplitude - contact phase, 24 versus 37% MVIC; RMS amplitude - midstance/propulsion, 25 versus 39% MVIC). These findings indicate that peroneus longus is working less during the contact and midstance/propulsion phases in participants with flat-arched feet, compared to those with normal-arched feet. Effect sizes for these differences were 0.48 and 0.58 for peak amplitude (contact phase) and RMS (midstance/propulsion phase) amplitude respectively, representing moderate differences in muscle activity. These functional differences between foot types may reflect a compensatory lack of activity in peroneus longus to avoid further overloading the medial arch. Alternatively, this finding may occur as a result of flat-arched feet being less laterally unstable, therefore requiring less peroneus longus activity.

A further significant finding was that participants in the flat-arched group functioned at a significantly greater percentage of their maximum amplitude for tibialis anterior during contact phase, compared to participants in the normal-arched group (peak amplitude, 65 versus 46% MVIC; RMS amplitude, 43 versus 32% MVIC). Effect sizes for these differences were 1.3 and 0.87 for peak and RMS amplitude respectively, representing large differences in muscle activity. During contact phase of the gait cycle, tibialis anterior is thought to decelerate ankle joint plantarflexion and resist foot pronation [22]. Interestingly, the role of tibialis anterior in resisting pronation of the foot during the contact phase was not assisted via strong co-activation of tibialis posterior. In fact, tibialis posterior functioned at a lower percentage amplitude during contact phase compared to the normal arched group, although this finding did not reach statistical significance ($p = 0.058$).

There were no differences in medial gastrocnemius timing or amplitude EMG parameters comparing normal- and flat-ached feet. This finding adds to the growing body of evidence that medial gastrocnemius muscle activation is not affected by differences in foot posture [7]. Furthermore, this indicates that medial gastrocnemius is unlikely to have a significant function as an inverter of the hindfoot, since deviations in hindfoot alignment have not been shown to cause changes in the activity of this muscle [7].

The finding that participants in the flat-arched foot group walked slightly faster than those in the normal-arched group (mean difference, 0.11 ms) was unexpected and may have influenced some results in this study. It should be noted that both foot posture groups were instructed to walk at their normal comfortable walking speed and data collection was carried out under identical conditions. This difference in walking speed required some consideration, as numerous studies investigating the influence of walking speed on EMG amplitude have indicated that EMG amplitude increases linearly with walking speed [23-25]. There may be a biological or compensatory reason why participants with flat-arched feet walked faster than those with normal-arched feet, such as a means of increasing stability of the foot and lower limb during walking. In this case, the independent variable (flat-arch foot posture) may have influenced the covariate (walking speed), and this poses a conceptual issue preventing us from adopting an analysis of co-variance approach to adjust for walking speed [26]. However, we believe that the differences in muscle activity observed between the groups are unlikely to have been caused by differences in walking speed. Participants in the flat-arch group functioned at a significantly *lower* percentage of their maximum amplitude for peroneus longus during contact phase and midstance/ propulsion phase, despite walking faster. Furthermore, den Otter and colleagues [23] have shown that negative amplitude gains (i.e. increased amplitude with reduced walking speed) of peroneus longus only occur at very slow speeds. Therefore, it is unlikely that the normal-arched group displayed a relative 'negative gain' compared to the flat-arched group.

The results presented here may have implications for the management of lower extremity overuse conditions. Although it is still unknown whether these functional differences in muscle activation are beneficial or detrimental in relation to injury, preliminary evidence indicates that these differences may be reversible with intervention [27]. In a recent study, Franettovich and colleagues [27] investigated the effect of an anti-pronation taping technique on lower limb EMG muscle activation in four adults with pronated foot posture. They reported that the anti-pronation tape significantly reduced the EMG amplitude of the tibialis posterior and tibialis anterior muscles during walking. While this indicates that anti-pronation tape may bring muscle function in a flat-arched foot closer to that observed in a normal-arched foot, further research is required to ascertain whether these changes are associated with clinical outcomes.

This study has several strengths, including the use of a rigorous protocol to classify foot posture, the use of in-dwelling needle electrodes to assess tibialis posterior and peronus longus, and a relatively large sample size (n = 60 compared to 17 to 43 in previous studies [2,7-10,12]). However, the results of this study also need to be interpreted in light of two limitations. Firstly, we did not simultaneously record other kinematic and kinetic variables, thus we can only speculate as to the mechanical effects of the EMG differences. Secondly, the participants in this study were relatively homogenous as they were mostly young, healthy and without musculoskeletal injury. Therefore, caution should be taken in generalising these results to symptomatic or clinical populations. A further limitation was that we used MVICs to normalise the EMG amplitude parameters. It is difficult to control and monitor the participants' effort or output with MVICs which may be a factor that leads to greater between-participant variability compared to other normalisation protocols [20].

Conclusion

Lower limb muscle function is affected by foot posture. The flat-arched group functioned at a greater percentage of their maximum EMG amplitude during contact phase for tibialis anterior and during midstance/propulsion for tibialis posterior, compared to normal-arched feet. The flat-arched foot group also functioned at a lower percentage of their maximum EMG amplitude throughout stance phase for peroneus longus, compared to normal-arched feet. These differences in muscle activity may reflect neuromuscular compensation to reduce overload of the medial longitudinal arch in people with flat-arched feet. Further research is required to determine whether these differences in muscle function are associated with injury.

Competing interests

HBM and KBL are Editor-in-Chief and Deputy Editor-in-Chief, respectively, of *Journal of Foot and Ankle Research*. It is journal policy that editors are removed from the peer review and editorial decision-making processes for papers they have co-authored.

Authors' contributions

GSM, HBM and KBL conceived the idea and obtained funding for the study. GSM, HBM and KBL designed the study protocol. GSM recruited participants, conducted the laboratory testing and processed data. GSM, HBM and KBL drafted the manuscript. All authors have read and approved the final manuscript.

Additional material

Additional file 1
A video demonstration of the insertion of an intramuscular electrode into tibialis posterior via the posterior approach
Click here for file
[http://www.biomedcentral.com/content/supplementary/1757-1146-2-35-S1.m4v]

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Do foot orthoses change muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet?

Preface

The systematic review (Chapter 2) included 12 studies that investigated the effect of foot orthoses on lower limb muscle function during walking or running. The review indicated that irrespective of the foot orthosis (FO) material, there was some evidence that peroneus longus and tibialis anterior EMG amplitude, and tibialis anterior EMG duration are significantly greater when wearing FOs. However, it was unclear whether increasing or decreasing EMG amplitude or duration was beneficial relative to 'normal' EMG activity. While it makes intuitive sense that FOs would be beneficial if they can bring muscle activity (measured via EMG) closer to that seen in a normal or non-pathological population, no previous EMG studies investigating FOs have investigated this issue.

In Chapter 6, participants with normal- and flat-arched feet were investigated to determine whether foot posture influences lower limb muscle activity during walking. The findings of this study indicated that during contact phase, people with flat-arched feet exhibit increased activity of tibialis anterior and decreased activity of peroneus longus. During midstance/propulsion, they also exhibited increased activity of tibialis posterior and decreased activity of peroneus longus, compared to those with normal-arched feet. To investigate whether these differences in EMG muscle activity could be brought closer to that seen in those with normal-arched feet, the participants with flat-arched feet were issued with two types of FOs two weeks prior to the EMG testing session.

Therefore, the aim of this study was to investigate whether FOs change muscle activity in the participants with flat-arched feet closer to a pattern observed in participants with normal-arched feet.

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Do foot orthoses change lower limb muscle activity in flat-arched feet towards a pattern observed in normal-arched feet?

George S. Murley [a,b,*], Karl B. Landorf [a,b], Hylton B. Menz [b]

[a] Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, 3086, Australia
[b] Musculoskeletal Research Centre, Faculty of Health Sciences, La Trobe University, Bundoora, 3086, Australia

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ABSTRACT

Background: One of the hypothesised mechanisms by which foot orthoses obtain their clinical effect is by influencing muscle activity, however previous studies have reported highly variable findings. The aim of this study was to determine whether orthoses change muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet.

Methods: Thirty young asymptomatic adults with flat-arched feet were recruited. Foot posture was classified using two clinical measurements and four skeletal alignment measurements from weight-bearing foot x-rays. Electromyographic activity was recorded while walking from tibialis posterior and peroneus longus via in-dwelling wire electrodes, and from tibialis anterior and medial gastrocnemius via surface electrodes. Four experimental conditions were assessed: (i) barefoot, (ii) shoe only, (iii) a heat-moulded (modified) prefabricated foot orthosis, and (iv) a 20-degree inverted-style customised foot orthosis.

Findings: During the contact phase of gait, tibialis posterior electromyographic amplitude decreased significantly with the prefabricated orthosis (peak amplitude — 19% decrease, $P=0.007$; RMS amplitude — 22% decrease, $P=0.002$) and the customised orthosis (peak amplitude — 12% decrease, $P<0.001$, RMS amplitude — 13% decrease, $P=0.001$), compared with the shoe-only condition. During the midstance/propulsive phase, peroneus longus electromyographic amplitude increased significantly with the prefabricated orthosis, compared with the shoe-only (peak amplitude — 21% increase, $P=0.024$; RMS amplitude — 24% increase, $P=0.019$) and customised orthosis conditions (peak amplitude — 16% increase, $P=0.028$).

Interpretation: The foot orthoses significantly altered tibialis posterior and peroneus longus electromyographic amplitude. However, only the modified prefabricated orthosis changed peroneus longus electromyographic amplitude towards a pattern observed with normal-arched feet. Otherwise, few differences were found between the modified prefabricated and customised orthoses. Further research is required to determine whether these changes in muscle function are associated with clinical outcomes.

1. Introduction

Foot orthoses (FOs) are commonly used in the conservative management of a range of lower limb overuse conditions (Landorf and Keenan, 2000). Although there is no universally adopted classification for different types of FOs, one key distinction is between prefabricated 'off the shelf' FOs and more expensive, customised FOs (Landorf et al., 2001). Irrespective of the variety of materials and manufacturing processes available, FOs generally aim to realign skeletal structures, alter movement patterns of the lower extremity during gait and most importantly, reduce symptoms associated with lower limb conditions (Collins et al., 2007; Landorf and Keenan, 2007; McMillan and Payne, 2008).

In light of the proposed effects of FOs on lower limb biomechanics, we recently conducted a systematic review of studies that investigated the effect of FOs on lower limb muscle activity during walking or running (Murley et al., 2009b). The review concluded that there is some evidence that various styles of FOs increase electromyographic (EMG) amplitude of tibialis anterior and peroneus longus (Tomaro and Burdett, 1993; Nawoczenski and Ludewig, 1999; Mundermann et al., 2006; Murley and Bird, 2006). However, it is unclear whether these changes represent optimisation in muscle function; that is, whether FOs alter the pattern of muscle activity in flat-arched feet towards the pattern observed in 'normal' feet.

Aside from varus and valgus wedging, one of the most common features of FOs is the contour under the medial longitudinal arch of the foot. It is plausible that medial arch support provided by an FO will assist tibialis posterior in reducing pronation of the foot, particularly of the rearfoot and midfoot, although evidence of this relationship is lacking. To date, only one study has investigated the effect of FOs on

* Corresponding author. Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora, VIC, 3086, Australia.
E-mail address: g.murley@latrobe.edu.au (G.S. Murley).

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tibialis posterior EMG during gait (Stacoff et al., 2007). In this study, intramuscular electrodes were used to record tibialis posterior EMG activity from five participants (age range: 25–69 years) with flat-arched foot posture. No significant differences were found between the three styles of FOs tested. However, it has been found that there is high between-participant variability for tibialis posterior EMG during walking (Murley et al., 2009a). It is not surprising, therefore, that this study did not detect systematic changes in muscle activity when comparing the FOs in only five participants. The use of such small sample sizes within the EMG literature is widespread, and may be responsible for the conclusions reached by many authors that FOs have variable and non-systematic effects on lower limb EMG muscle activity during walking (Murley et al., 2009a).

To address some of these issues, we recently conducted a study comparing EMG muscle activity in 30 adults with flat-arched feet to 30 adults with normal-arched feet during walking (Murley et al., 2009c). The results of this study demonstrated that during the contact phase of gait, the flat-arched group exhibited increased activity of tibialis anterior and decreased activity of peroneus longus. During midstance/propulsion, the flat-arched group exhibited increased activity of tibialis posterior and decreased activity of peroneus longus, compared with those with normal-arched feet.

Accordingly, the aim of this study was to investigate whether FOs change lower limb muscle activity in people with flat-arched feet towards the pattern observed in people with normal-arched feet.

2. Methods

2.1. Participants

Thirty young adults with flat-arched feet (15 male and 15 female) aged 18 to 37 years were recruited to this study (Table 1). To categorise foot posture, we developed a foot screening protocol that included both clinical and radiographic measures of foot posture to recruit participants with flat-arched foot posture (Murley et al., 2009d). This protocol was derived from normative foot posture values for two clinical measurements (the arch index and normalised navicular height to truncated foot length) and four angular measurements obtained from antero-posterior and lateral x-rays (talus-second metatarsal angle, talonavicular coverage angle, calcaneal inclination angle and calcaneal-first metatarsal angle) (Table 1). To qualify for the flat-arched group, participants had to exhibit an arch index or normalised navicular height to truncated foot length measurement greater than two standard deviations from mean values obtained for people with normal-arched feet (Murley et al., 2009d). Furthermore, their radiographic measurements had to be greater than 1 standard deviation from the mean values obtained for people with normal-arched feet for either the sagittal and or transverse plane measurements (Murley et al., 2009d).

The participants were without symptoms of macrovascular (e.g. angina, stroke, peripheral vascular disease) and/or neuromuscular disease, or any biomechanical abnormalities that affected their ability to walk. Ethical approval was obtained for the study from the La Trobe University Human Ethics Committee (Ethics ID: FHEC06/205) and it was registered with the Radiation Safety Committee of the Victorian Department of Human Services. The x-rays were performed in accordance with the Australian Radiation Protection and Nuclear Safety Agency Code of Practice for the Exposure of Humans to Ionizing Radiation for Research Purposes (2005).

2.2. Foot orthoses (FOs)

Two different FOs commonly used in clinical practice were dispensed to participants: (i) a heat-moulded (modified) foam prefabricated foot orthosis and (ii) a 20-degree inverted-style customised foot orthosis (Fig. 1). Each participant received a pair of

Table 1
Participant anthropometric and foot posture characteristics.

General anthropometric	
Gender ratio (female/male)	15/15
Age mean (SD) years	21.8 (4.3)
Height mean (SD) cm	171.0 (10.0)
Weight mean ± (SD) kg	73.3 (15.5)
Left or right foot count	13 right 17 left
Clinical measurements	
Arch index^ (SD) [mean]	0.30 (0.07)
Normalised navicular height to truncated foot length** (SD) [mean]	0.18 (0.04)
Radiographic measurements	
CIA mean (SD) degrees	15.7 (4.5)
C1MA mean (SD) degrees	142.3 (6.0)
TNCA mean (SD) degrees	27.6 (9.0)
T2MA mean (SD) degrees	27.1 (10.1)
Walking velocity	
Metres per second (SD) ms^{-1}	1.21 (0.13)

CIA — calcaneal inclination angle.
C1MA — calcaneal-first metatarsal angle.
TNCA — talo-navicular coverage angle.
T2MA — talus-second metatarsal angle.
^ The arch index was calculated as the ratio of area of the middle third of the footprint to the entire footprint area not including the toes, with a higher ratio indicating a flatter foot.
** Normalised navicular height to truncated foot length is the ratio of navicular height relative to the truncated length of the foot. Navicular height is the distance measured from the most medial prominence of the navicular tuberosity to the supporting surface. Foot length is truncated by measuring the perpendicular distance from the first metatarsophalangeal joint to the most posterior aspect of the heel.

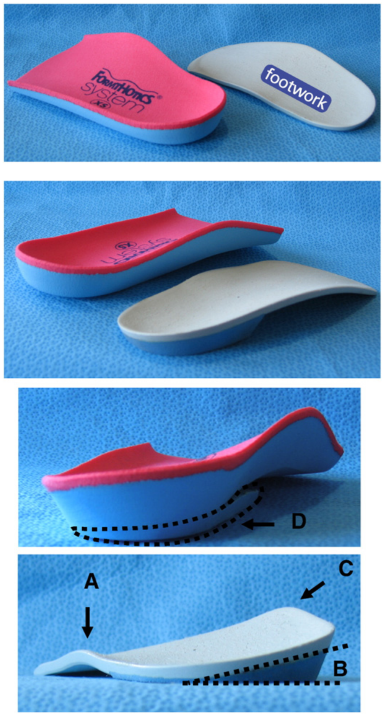

Fig. 1. Modified prefabricated (left) and customised (right) foot orthoses (left foot). Features of the customised orthosis: A — cuboid notch; B — 20° medial wedge; C — medial arch flare; and modified prefabricated orthoses: D — medial heel wedge.

Table 2
Characteristics of the prefabricated and customised foot orthoses.

Parameter	Modified prefabricated FO	Customised FO
Orthotic material	Dual-density polyethylene foam	Polypropylene plastic
Wedge	6 mm medial heel wedge (ethylene vinyl acetate) added under the heel region of the orthosis	Orthotic shell posted at 20 degree inverted. The heel region of the shell is supported by ethylene vinyl acetate wedge
Length	Three-quarter length	Three-quarter length
Arch support	Heat-moulded to individual participants' feet to enhance contour to the arch area of the foot	Plaster cast modifications are performed to contour the orthotic shell to the sustentaculum tali region of the arch

customised FOs and a pair of modified prefabricated FOs. The 20-degree inverted wedge was incorporated into the design of the customised FOs to provide greater supination force on the foot than would otherwise be exerted by a moulded shell alone (Blake, 1986). It has been hypothesised that this modification increases the supinatory force exerted by the orthosis at the rearfoot (i.e. increases the supination moment across the subtalar joint axis) compared with a standard FO (Blake and Ferguson, 1991). The rationale for including this feature was because the participants' foot posture was very flat from a clinical and radiographic prospective. The main features of the prefabricated and customised foot orthoses are summarised in Table 2.

The modified prefabricated FO was a three-quarter-length Formthotic™ made from dual-density polyethylene foam (Foot Science International, Christchurch, New Zealand). This device was heated with a heat gun and moulded to the individual participants' feet while they maintained a neutral subtalar joint position. Moulding was performed to enhance contour of the FO to the arch area of the foot. A 6 mm medial-heel wedge was adhered under the heel of the orthosis to provide additional resistance to rearfoot pronation during walking (Fig. 1). This modification process is consistent with the manufacturer's recommendations.

For the customised FO, a plaster cast impression was taken of each participant's feet in the subtalar joint neutral position using the suspension technique (Root et al., 1971). The plaster casts were taken by a podiatrist with 10 years of clinical experience and were sent to an orthotic laboratory (Footwork Podiatric Laboratory, Hallam, Australia). The laboratory custom-manufactured a single pair of 20-degree inverted-style FOs for each participant (Blake, 1986). The device was made from a semi-rigid 4 mm polypropylene thermoplastic shell and included features considered to minimise rearfoot pronation (Fig. 1).

Both pairs of FOs were dispensed to participants on average 12 days prior to EMG testing. To ensure the FOs were comfortable when the participant presented for EMG testing, they were advised to build up time in each pair and alternate the FOs each consecutive day (i.e. the prefabricated pair one day and the customised pair the next day).

Orthotic comfort was measured during the experimental period to; (i) evaluate orthotic comfort over time, and (ii) compare the prefabricated and customised FOs to determine if there were differences in comfort between the two devices. Participants rated the comfort of the FOs on a 150 mm visual analogue scale that has been utilised by similar studies to assess orthotic comfort (Mundermann et al., 2003) (Fig. 2). These comfort ratings were performed during the initial dispensing consultation and after the habituation period. During the initial dispensing consultation, the FOs were trialled in a pair of shoes comprising a flexible canvas upper and flat thin rubber sole (Dunlop Volley™, Pacific Dunlop Ltd, Melbourne, Australia). With the participant blinded, a randomly allocated pair of FOs were placed in the shoe and fitted to the participants' feet. The participant then walked for approximately 1 min before performing the comfort rating — this process was repeated for the second pair FOs. The comfort ratings were performed by placing a mark on the 150 mm visual analogue scale that represented the participants' comfort rating (Fig. 2). A second comfort rating was performed following the habituation period, however, on this occasion the participants were not blinded to the FOs.

2.3. Experimental protocol

Bipolar fine-wire intramuscular electrodes were used to record the EMG signal from tibialis posterior and peroneus longus. The electrodes were fabricated from 75 μm Teflon® coated stainless steel wire (A-M Systems, Washington, USA) with 1 mm of insulation stripped to form the recording surface of the two wires. The electrode wires were inserted into a 23 gauge sterilized single use hypodermic needle with the exposed electrode tips bent 3 mm and 5 mm to prevent the contact areas from touching during recording. For tibialis posterior, the intramuscular electrode was inserted at a distance of approximately 50% between the popliteus cavity to the medial malleolus (Leis and Trapani, 2000). For peroneus longus, the intramuscular electrode was inserted at approximately 20% of the distance from the head of fibula to the lateral malleolus, starting from the head of the fibula (Leis and Trapani, 2000). The process of fine-wire electrode construction and positioning of wires in vivo was undertaken in accordance with previous work (Murley et al., 2009a,c).

Tibialis anterior and medial gastrocnemius EMG was recorded with the use of DE-3.1 surface electrodes (Delsys Inc., Boston, USA). The electrodes featured a double differential 3-bar type configuration with a 99.9% silver contact material and an inter-electrode distance of 10 mm. The application of surface electrodes followed the recommendations of SENIAM (Hermens et al., 1999). For tibialis anterior,

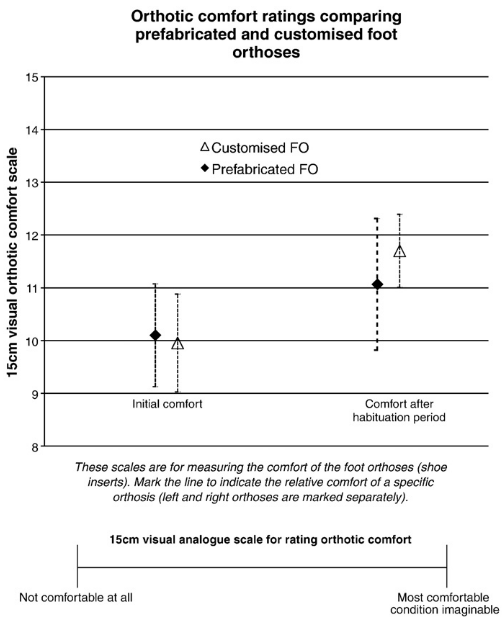

Fig. 2. Foot orthoses comfort scores for the left and right foot recorded during the initial dispensing and following the wear-in period.

the surface electrode was placed at approximately 20% of the distance from the tuberosity of tibia to the inter-malleoli line, starting from the tuberosity of tibia (Hermens et al., 1999). For medial gastrocnemius, the surface electrode was placed at approximately 25% of the distance from the medial side of the popliteus cavity to the calcaneal tubercle (Hermens et al., 1999).

Only muscles with key agonist/antagonist function producing dorsiflexion/plantarflexion and inversion/eversion of the foot were included in this study.

The temporal characteristics of the walking cycle were measured using circular force sensitive resistors (footswitches) with a diameter of 13 mm (Model: 402, Interlink Electronics, California, USA). These were placed on the plantar surface of the interphalangeal joint of the hallux and the most posterior plantar aspect of the calcaneus to record the timing of heel contact, toe contact, heel off and toe off.

During testing, participants walked under all four randomly allocated conditions: (i) barefoot, (ii) shoe only, (iii) shoe plus the modified prefabricated FO, and (iv) shoe plus the customised FO. The shoe used for testing was the same used during the initial comfort ratings. Participants were instructed to walk at their self-selected walking speed, which was established following a warm-up period from two trials along a 9 m walkway. Six trials were recorded for each condition. Any trial exceeding ±5% of the average warm-up speed was excluded and the trial was repeated.

EMG amplitude parameters for all conditions were normalised from the corresponding amplitude parameter recorded from the barefoot walking condition (i.e. dynamic and sub-maximal normalisation) (Murley et al., 2009e).

2.4. EMG data processing

The raw EMG signal was passed through a differential amplifier at a gain of 1000 with a sampling frequency of 2 kHz. A band pass filter (built into the amplifier; Delsys Inc., Boston, USA) of 20–2000 Hz was applied to the intramuscular electrodes and 20–450 Hz for the surface electrodes.

EMG and footswitch data were analysed from the 3rd or 4th stride depending on the quality of the footswitch signal. Two consecutive strides (i.e. comprising three consecutive heel contacts from the ipsilateral limb) were analysed for each trial and averaged from the last four of six trials for each speed (i.e. four average gait cycles derived from 8 ipsilateral steps). Three EMG parameters were analysed for each muscle, including: (i) time of peak amplitude; (ii) root mean square amplitude (RMS); and (iii) peak amplitude. These parameters have been utilised in previous single-session investigations (Fig. 3) (Murley and Bird, 2006; Murley et al., 2009a,c). The following phases of the gait cycle were assessed (based on when these muscles are most active in normal-arched feet): tibialis posterior and peroneus longus — contact and combined midstance/propulsion phase; tibialis anterior — contact phase; and medial gastrocnemius — combined midstance/propulsion phase (Murley et al., 2009a).

2.5. Statistical analysis

Skewness and kurtosis values were used to evaluate the distribution of data. To test for differences between conditions, a series of one-way repeated measure ANOVA tests were conducted. The within-subject factors for each muscle were as follows:

(i) Tibialis posterior — phases of gait (2) × EMG parameters (3) × walking conditions (3)
(ii) Peroneus longus — phases of gait (2) × EMG parameters (3) × walking conditions (3)
(iii) Tibialis anterior — phases of gait (1) × EMG parameters (3) × walking conditions (3)
(iv) Medial gastrocnemius — phases of gait (1) × EMG parameters (3) × walking conditions (3)

Where data violated the assumption for sphericity as determined by non-significant results ($P<0.05$) for the Mauchley's test, the F-ratio and degrees of freedom were taken from the Greenhouse–Geisser epsilon. To account for multiple comparisons, statistically significant univariate F-statistics were evaluated with Bonferroni post hoc analysis ($P=0.05$). The percentage mean difference, 98% confidence

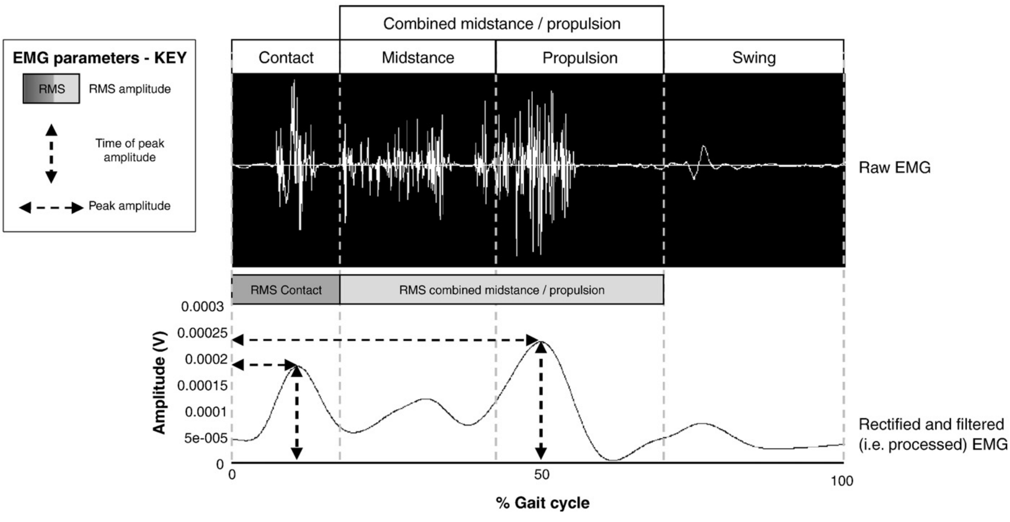

Fig. 3. A single gait cycle showing raw and processed EMG for tibialis posterior from a single participant. Time of peak amplitude, peak amplitude and RMS amplitude (root mean square) were derived from the linear envelope (processed curve).

intervals and effect sizes were calculated for significant *post hoc* findings. Effect size (d) was computed as a ratio of the mean change score divided by the standard deviation of the baseline scores. Cohen (Cohen, 1988) has suggested that an effect size of 0.20 or less represents a small change; 0.50 represents a moderate change; and 0.80 represents a large change.

A two-way repeated measures ANOVA was used to assess orthotic comfort ratings for each 'foot orthosis' (two levels: prefabricated and customised) and between each 'rating session' (two levels: session one when the orthoses were dispensed to participants and session two after the two-week habituation period).

3. Results

3.1. Effect of foot orthoses on lower limb muscle EMG activity

During the contact phase, significant within participant effects were detected for tibialis posterior peak amplitude ($F_{1.34,37.57}=7.58$, $P=0.005$) and RMS amplitude ($F_{1.43,40.04}=9.71$, $P=0.009$) [Greenhouse–Geisser adjusted F-statistic and degrees of freedom]. In addition, significant within participant effects were also detected for peroneus longus RMS amplitude ($F_{2,56}=3.55$, $P=0.035$) and tibialis anterior time of peak amplitude ($F_{2,58}=3.94$, $P=0.025$). During the midstance/propulsion phase, significant within participant effects were detected for peroneus longus peak amplitude ($F_{2,54}=5.16$, $P=0.009$). Multiple pair-wise comparisons between conditions revealed significant findings for tibialis posterior and peroneus longus. Fig. 4a–c present forest plots of pair-wise comparisons for all muscles and conditions with Bonferroni-adjusted 98% confidence intervals. Fig. 5 shows tibialis posterior and peroneus longus EMG ensemble averages derived from a single gait cycle for all participants.

3.2. Contact phase

Tibialis posterior EMG amplitude decreased significantly with the prefabricated orthosis (peak amplitude — 19% decrease, $P=0.007$; RMS amplitude — 22% decrease, $P=0.002$) and the customised orthosis (peak amplitude — 12% decrease, $P<0.001$, RMS amplitude — 13% decrease, $P=0.001$), compared to the shoe-only condition (Figs. 4 and 5).

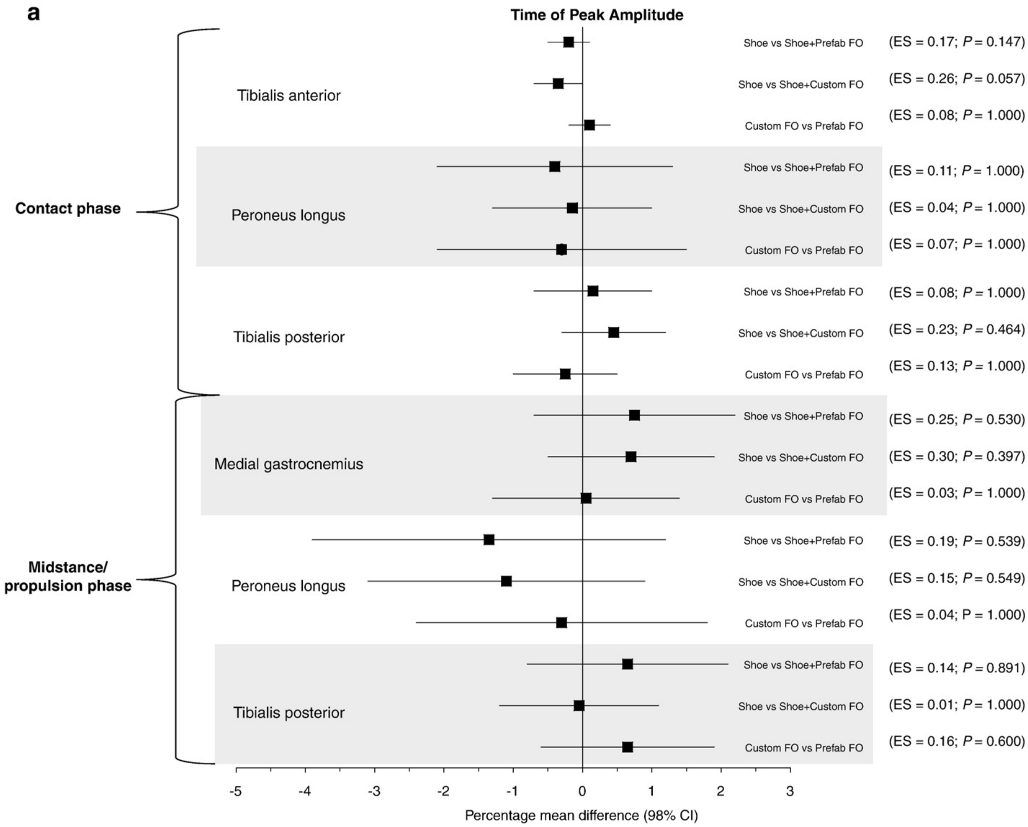

Fig. 4. a–c. Forest plots for *post hoc* comparison between shoe-only condition, customised foot orthoses and modified prefabricated foot orthoses. Bonferroni-adjusted (98%) confidence intervals for multiple comparisons. The change in direction (i.e. positive or negative) of each plot is relative to the condition listed first. TimePeak — time of peak amplitude; PeakAmp — peak amplitude; RMS — root mean square.

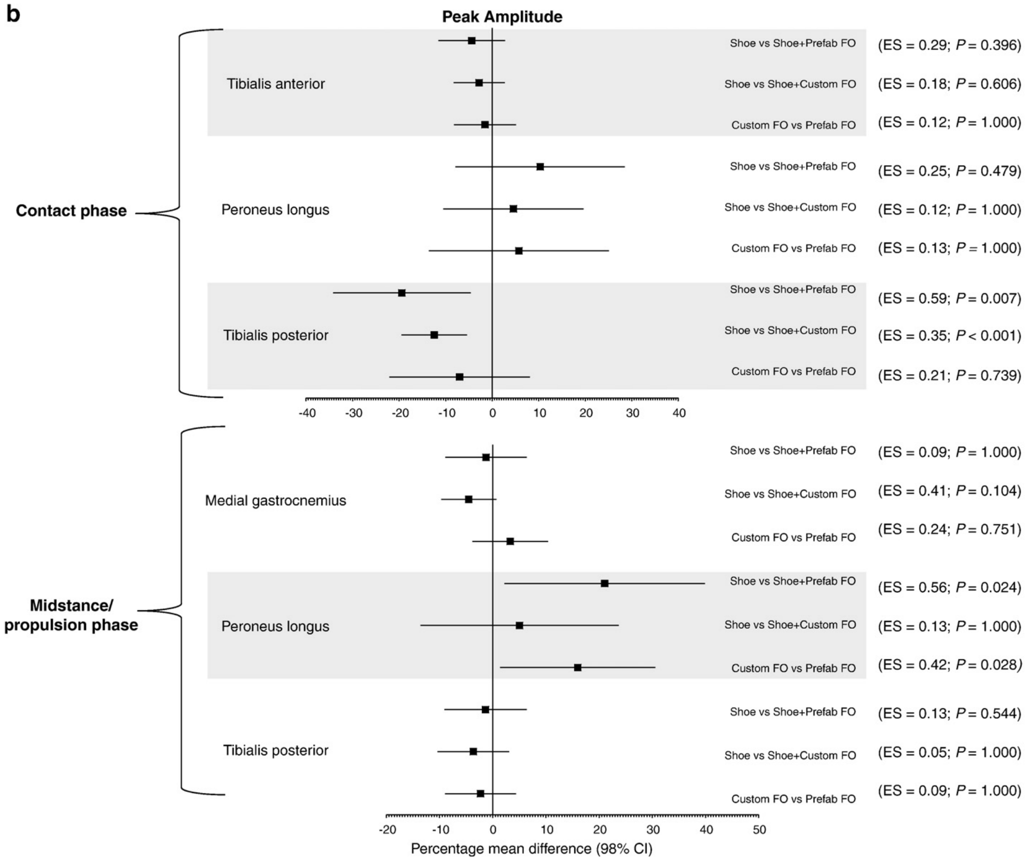

Fig. 4 (*continued*).

3.3. Midstance/propulsion phase

Peroneus longus EMG amplitude increased significantly with the prefabricated orthosis, compared with the shoe-only (peak amplitude — 21% increase, $P=0.024$; RMS amplitude — 24% increase, $P=0.019$) and customised orthosis conditions (peak amplitude — 16% increase, $P=0.028$) (Figs. 4 and 5).

3.4. Foot orthotic comfort ratings

Comfort ratings were available for 25 of the 30 participants. The mean comfort scores at the time of initial dispensing were 67% (range: 40–140 mm) for the modified prefabricated FO and 66% (range: 30–140 mm) for the customised FO. The mean comfort scores after the two-week habituation period were 74% (range: 30–150 mm) for the modified prefabricated FO and 78% (range: 80–150 mm) for the customised FO (Fig. 2). Significant effects for 'rating session' were detected ($F_{1,24}=13.99$, $P=0.001$) which indicated that orthotic comfort improved by 9% (95% CI, 4 to 14%; $P=0.001$) comparing the initial dispensing session to the second rating session following the two-week habituation period. There were no significant differences in comfort between the modified prefabricated and customised FOs at either the dispensing session or following two weeks of habituation.

4. Discussion

The aims of this study were to investigate whether modified prefabricated and customised FOs influence lower limb muscle activity, and if so, whether they optimise or 'reverse' the abnormal lower limb muscle activity previously observed in people with flat-arched feet (Murley et al., 2009c). The results revealed significant changes in tibialis posterior EMG amplitude with both styles of FOs, however only the prefabricated FO had a significant effect on peroneus longus EMG amplitude.

During the contact phase of gait, both styles of FOs significantly decreased tibialis posterior EMG amplitude compared with the shoe-only condition. Effect sizes for these significant findings ranged from 0.32 to 0.59, representing small to moderate differences in muscle activity. To our knowledge, this is the first study to detect significant gait-related changes in tibialis posterior using FOs. As tibialis posterior is thought to resist rearfoot eversion during the contact phase of gait,

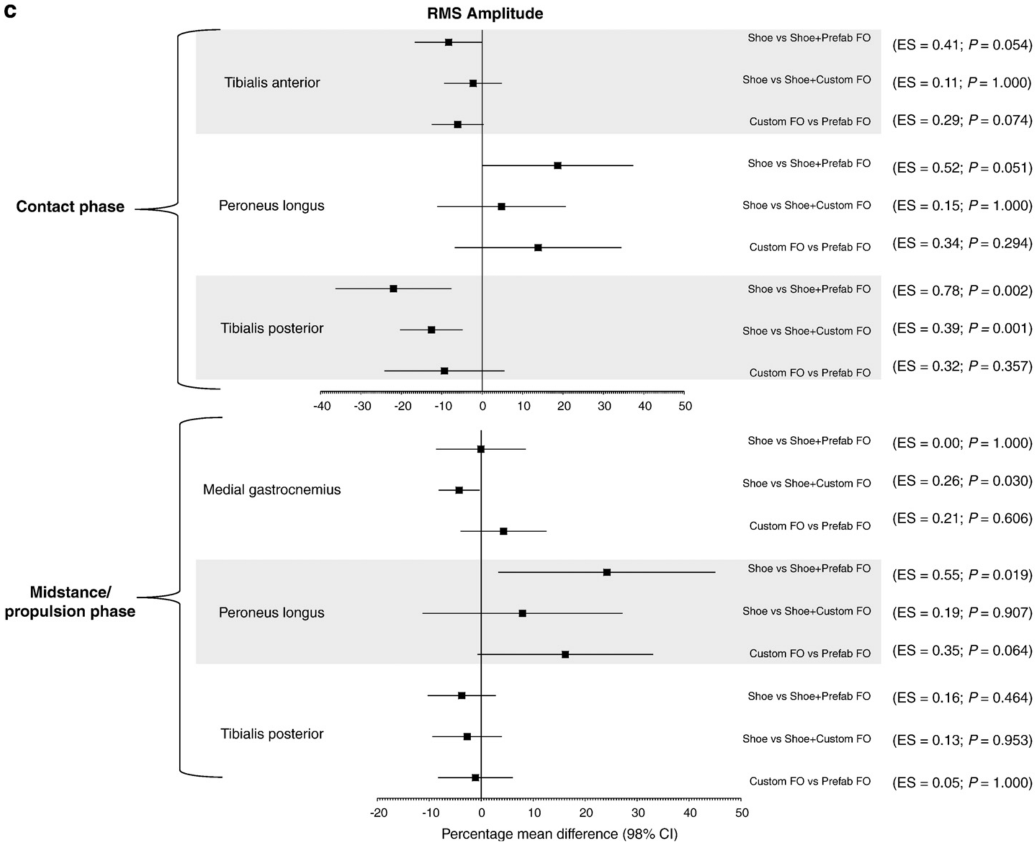

Fig. 4 (*continued*).

it could be suggested that the decrease in EMG amplitude during this phase may reflect a reduction in the kinematic demand for tibialis posterior when the foot is supported by FOs. This mechanism linking the FOs intervention and the changes in tibialis posterior EMG is supported by recent kinematic research which has found that 'semi-custom' and 'custom' FOs reduced rearfoot eversion during walking (Zifchock and Davis, 2008; Eslami et al., 2009; Mills et al., 2009).

While this finding is of interest, it is unclear whether reducing tibialis posterior EMG amplitude during contact phase is functionally beneficial in people with flat-arched feet. In previous research comparing tibialis posterior EMG activity of normal- and flat-arched feet we detected significant differences during only the midstance/ propulsion phase, with no significant differences detected during the contact phase (Murley et al., 2009c). However, another study which investigated participants with long-standing rheumatoid arthritis reported greater tibialis posterior amplitude during the contact phase in participants with valgus foot alignment, compared to normally-aligned feet (Keenan et al., 1991). Therefore, it is possible that this specific population (i.e. people with valgus foot deformity related to systemic disease) may benefit more from a reduction in tibialis posterior EMG amplitude during the contact phase than the asymptomatic population in our study. Further research should examine whether FOs reduce tibialis posterior EMG amplitude during the contact phase in this population and whether this is associated with a reduction in symptoms.

During the midstance/propulsion phase of gait, the modified prefabricated FO significantly increased peroneus longus EMG amplitude compared with the shoe only and customised FO. Effect sizes for these significant findings ranged from 0.35 to 0.56, representing small to moderate differences in muscle activity. This finding is consistent with previous studies which have reported that various styles of FOs increase peroneus longus EMG amplitude during gait (Tomaro and Burdett, 1993; Nawoczenski and Ludewig, 1999; Mundermann et al., 2006; Murley et al., 2009b).

We hypothesise that FOs increase peroneus longus EMG amplitude merely because the foot is made more laterally unstable. Increasing peroneus longus EMG amplitude with FOs during walking may secondarily assist with plantarflexion of the first ray (Murley and Bird, 2006). This may, in turn, assist dorsiflexion at the first metatarsophalangeal joint and help facilitate the windlass mechanism

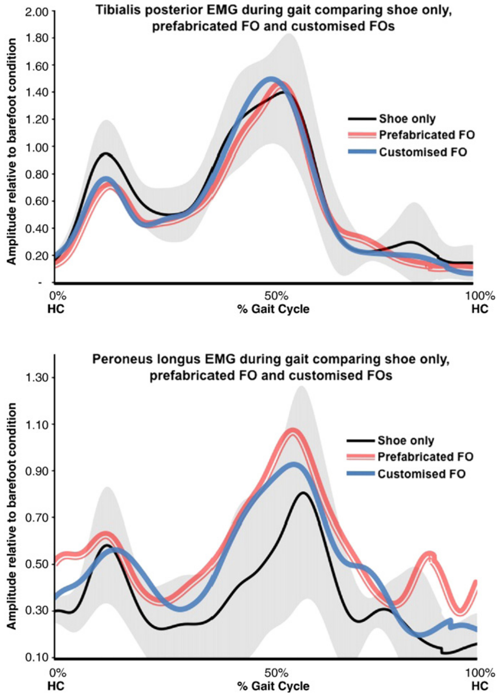

Fig. 5. EMG ensemble average for tibialis posterior and peroneus longus derived from a single gait cycle for all participants. Grey shading represents 95% confidence interval for the shoe-only condition. For clarity, error is not shown for the modified prefabricated and customised FO conditions. The curves differ slightly to the actual results (Fig. 4 — forest plots), as these curves are derived from a single gait cycle for each participant to illustrate the main findings. HC — heel contact.

during propulsion (Hicks, 1954). In addition, there may be some functional benefit in increasing peroneus longus EMG amplitude during the midstance/propulsion phase, as we have recently shown that people with flat-arched feet have significantly lower peroneus longus EMG amplitude compared with normal-arched feet (Murley et al., 2009c). While increasing peroneus longus EMG amplitude during midstance/propulsion alters the activity of this muscle closer to that observed in people with normal-arched feet, it remains uncertain what influence this has on lower limb function.

For tibialis anterior, the lack of significant findings was surprising given that the largest differences in muscle activity between normal- and flat-arched feet have previously been observed for tibialis anterior EMG amplitude during the contact phase (effect size = 1.3) (Murley et al., 2009c). There was a tendency for the FOs to *decrease* tibialis anterior EMG amplitude compared with the shoe-only condition, although this finding was not statistically significant. While other studies have reported a significant *increase* in tibialis anterior EMG amplitude with FOs, these studies all differ in the EMG normalisation and processing methods and some involved participants running (Tomaro and Burdett, 1993; Nawoczenski and Ludewig, 1999; Mundermann et al., 2006; Murley and Bird, 2006).

One of the reasons for the changes in muscle activity identified in this study may have been related to irritation from the FOs causing participants to walk differently. However, our results indicated that FO comfort improved significantly by 9% following the habituation period; and that the level of comfort was comparable to similar research reporting overall comfort of 'semi-custom' and 'custom' FOs (approximately 75% overall comfort after 2 weeks of habituation) (Zifchock and Davis, 2008). It is unclear, however, what level of comfort is biomechanically or clinically significant. Furthermore, while there was a significant difference between the modified prefabricated and customised FOs for peroneus longus EMG amplitude, there were no significant differences in comfort between these devices. Therefore, any differences observed between the two devices were unlikely to be related to discomfort.

4.1. Limitations

One of the strengths of this study was the use of a rigorous protocol to classify foot posture. Although this provided a reliable and valid method of identifying normal- and flat-arched feet, we acknowledge that the foot screening protocol was not designed to specifically identify people who would potentially benefit or respond to FOs. Several other factors, such as joint range of motion and dynamic gait observations may also be important variables for determining who will benefit most from FOs.

In regard to the orthotic comfort ratings, we recognise that the second rating method may have been affected by issues such as recall bias and that participants may have rated the comfort of the customised FOs more favourably knowing they were more expensive.

Another potential limitation of this study was the selection of the style of FOs, as there are no universally accepted guidelines for FO design or prescription (Petchell et al., 1998). Numerous other orthotic modifications designed to resist pronation, such as the medial heel skive technique (Kirby, 1992), could have been justifiably matched to the participants' flat-arched feet in this study and could possibly have led to further changes in muscle activity under similar experimental conditions. Furthermore, although the customized FOs were manufactured from a plaster cast of each participant's feet, we used the same degree of rearfoot posting. In the absence of any rigorous guidelines for orthoses prescription and given that the participants' foot posture was fairly homogenous (i.e. very flat-arched), we believe that the 20-degree cast correction was appropriate for the participants in this study.

5. Conclusion

Modified prefabricated and customised FOs are commonly used in clinical practice to treat a range of lower limb problems. While statistically significant changes were detected for tibialis posterior and peroneus longus, only the modified prefabricated FOs altered peroneus longus EMG amplitude in midstance/propulsion to a pattern closer to that observed in people with normal-arched feet. Overall, the FOs were perceived to provide equivalent comfort and they had a similar effect on muscle activity during walking. Further research is required to determine whether these changes in muscle function are associated with clinical outcomes.

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CHAPTER 8

8.0 Discussion

8.1 Summary of findings

This thesis presents the findings from five related studies for which the primary aim was to determine the effect of foot posture and foot orthoses on lower limb EMG activity during walking. The studies presented in Chapters 2 (systematic review), 3 (foot posture screening protocol) and 5 (EMG reliability) were undertaken to inform the design of the final two studies that investigated the effect of foot posture (Chapter 6) and foot orthoses (Chapter 7) on lower limb EMG activity. The results from the studies undertaken for this thesis satisfy the aims and objectives described in Chapter 1, which also provided the background that is relevant to this thesis.

The *first study* (Chapter 2) was a systematic review of the literature related to the effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running. The review concluded that there is some evidence to indicate that pronated foot posture and various styles of foot orthoses influence some lower limb EMG amplitude-related variables during gait. However, these studies were of only moderate methodological quality, with significant deficiencies in basic reporting of effect size and error. In addition, there were fundamental issues in the design of studies included in the systematic review, such as the use of inaccurate methods for classifying foot posture and the use of small sample sizes. To address these issues, a series of four studies were planned with the aim to systematically investigate the effect of foot posture and foot orthoses on lower limb muscle activity.

Accordingly, the *second study* (Chapter 3) was undertaken with the aim of designing a valid and reliable method of screening participants' foot posture. It was thought that this was a fundamental issue to address prior to undertaking the latter studies in this thesis. For this study, a combination of clinical and radiographic measurements was performed primarly because interpretation of clinical measurements is often confounded by soft tissue overlying the skeletal structures of the foot. As such, the radiographic measurements were regarded as the gold-standard for assessing skeletal alignment of the foot in a static weightbearing position.

Both clinical measures, the arch index and normalised navicular height ratios were selected because they provided valid and reliable measures of foot posture. When comparing these two clinical tests, normalised navicular height displayed the strongest association with radiographic angles, especially the calcaneal inclination angle.

Ninety-one participants underwent foot posture screening; thirty-two participants exhibited normal-arched foot posture, 31 participants were classified as having flat-arched feet, and a further 28 could not be classified has having either normal or flat-arched feet based on various radiographic measurements. The participants with normal- and flat-arched feet were subsequently recruited to a series of laboratory-based EMG studies.

The participants with normal-arched feet were involved in the *third study* (Chapter 5), which investigated the reliability of the EMG protocol used for this thesis. Within each test session, the timing and amplitude EMG parameters for all muscles displayed a low to moderate coefficient of variation, indicating that these measures could justifiably be used to compare differences between groups in a single session. To minimise between-participant variability, sub-maximal normalisation values were utilised. However, re-application of electrodes (i.e. between-session variability) resulted in large random error between sessions, particularly for tibialis posterior and peroneus longus, suggesting that this approach could not be used to evaluate changes over time.

The EMG data obtained from the participants with normal-arched feet in the reliability study were then compared to EMG data from participants with flat-arched feet in the *fourth study* (Chapter 6). The findings of this study indicated that during contact phase, people with flat-arched feet exhibited increased activity of tibialis anterior and decreased activity of peroneus longus. During midstance/propulsion, they also exhibited increased activity of tibialis posterior and decreased activity of peroneus longus, compared to those with normal-arched feet.

With these findings in mind, the *fifth study* (Chapter 7) was conducted to determine whether prefabricated and customised foot orthoses change muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet. The results indicated that tibialis posterior was significantly altered with both prefabricated and customised foot orthoses, while peroneus longus was only altered with prefabricated foot orthoses. The finding of altered peroneus longus activity with prefabricated foot orthoses was the only parameter that changed towards a more normal pattern of activity with foot orthoses. Otherwise, there were few differences between prefabricated and customised foot orthoses.

8.2 Synthesis

The following discussion is a synthesis of the main findings from each of the four studies relating to; (i) foot posture classification (Chapter 3); (ii) implications of EMG reliability findings (Chapter 5); (iii) effect of foot posture on selected lower limb muscles during gait (Chapter 6); and (iv) effect of prefabricated and customised foot orthoses on the electromyographic activity of selected lower limb muscles during gait (Chapter 7). The implications of these studies are discussed in detail below.

8.2.1 Development of the foot posture classification protocol

The clinical measures of foot posture (arch index and normalised navicular height) utilised in Chapter 3 were the most valid tests for predicting radiographic alignment of the foot. Further, as normative values were available for the clinical and radiographic measurements, the data presented in the foot posture screening protocol was based on a representation of real population characteristics rather than a theoretical ‘normal’ or ‘ideal’ foot described by Root and colleagues [127].

The participants with flat-arched feet displayed radiographic joint angles greater than one standard deviation from the mean of the normal-arched group. From a clinical perspective, the participants with flat-arched feet in this thesis resembled the kind of severely flat-arched feet that may be susceptible to musculoskeletal injury, and which are often prescribed foot orthoses. As such, these participants provided a useful cohort for investigating whether foot posture and foot orthoses influence lower limb muscle activity.

It is not clear, however, whether variations in foot posture in the sagittal or transverse planes provide the best descriptor of the flat-arched foot. Subsequently, we chose to include participants that displayed flat-arched feet with deformity in *either* or *both* planes (i.e. sagittal plane using lateral view x-ray angles or transverse plane using A-P view x-ray angles). The minimum requirement for participants to display a flat-arched foot in only a single plane (i.e. not in both planes) may have influenced the findings of the normal- and flat-arched foot study presented in Chapter 6 and the effect of the foot orthoses presented in Chapter 7. While it is uncertain what influence planal dominance may have on the four muscles investigated, it could be hypothesised that a flat-arched foot with more transverse plane malalignment (i.e. where the forefoot is abducted relative to midfoot) would require greater orthotic support to resist transverse plane motion than sagittal plane motion. Accordingly, it could also be proposed that such a foot type would be better suited to a foot orthosis with a feature such as a medial arch flare[4], which is considered to better resist the transverse plane motion of the foot than a foot orthoses without this feature. However, this is speculative and in the absence of evidence-based guidelines for foot orthosis prescription, we believe that the two

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[4] A medial arch flare is a crescent-shaped extension or widening of the medial arch of the shell of the foot orthosis.

pairs of foot orthoses issued to participants with flat-arched feet had features considered to reduce foot pronation in both the sagittal and transverse planes.

8.2.2 Implications of EMG reliability findings

The largest component of work in this thesis was the reliability study presented in Chapter 5. The rationale for conducting this study was to build on the work of Kadaba et al [162] and Bogey et al [160] by investigating other muscles such as tibialis posterior, and to apply more rigorous statistical analysis to the EMG data using absolute measures of reliability.

The results of this study demonstrated that the re-application of electrodes results in large random error between sessions, especially for tibialis posterior and peroneus longus. These findings had implications for the design of the final study that investigated foot orthoses presented in Chapter 7. That is, it was not feasible to conduct a study that required a repeated-test design (i.e. to evaluate the effect of foot orthoses over time). Accordingly, a single-session design was utilised to investigate the short to intermediate effects of foot orthoses on muscle activity.

A positive finding, however, was that within-session variability between gait trials among the temporal and amplitude characteristics was low to moderate, with averaged coefficient of variation values between 5 and 25%. This indicates that within-session variability for these variables was low enough to justify proceeding with further studies involving a single-session design, like those presented in Chapters 6 and 7.

With respect to normalisation of EMG amplitude parameters, there is a lack of clear evidence in the literature regarding the best method to adopt in gait research. It could be suggested that any normalisation method that reduces inter-subject variability is advantageous, as small and potentially important group or treatment effects can go undetected because of imprecision in the estimates (i.e. wide confidence intervals). Therefore, a key finding from the reliability study presented in Chapter 5 was that although both normalisation techniques were associated with poor between-session reliability, sub-maximal normalisation values produced consistently less within-session variability between participants for all EMG amplitude variables, compared to MVIC and un-normalised values, respectively.

While this finding indicated that subsequent investigations comparing interventions or individuals should incorporate a sub-maximal reference condition to decrease variability in the data, we were unable to apply this protocol for the foot posture/EMG study (Chapter 6). Instead, a MVIC normalisation approach was used because the participants with flat-arched feet were not instructed to undertake any additional walking trials, such as a ‘very fast’ walking speed, that could be used to sub-maximally normalise the EMG amplitude data. Despite this limitation, the differences in muscle activity comparing participants with normal- and flat-arched feet were large enough to detect several significant findings.

While the foot posture study (Chapter 6) was limited to the use of MVIC for normalisation, a sub-maximal normalisation protocol was utilised in the orthoses study (Chapter 7) to reduce between-participant variability. This was done by normalising the EMG amplitude data for the shoe only and foot orthoses conditions to the values obtained for the barefoot walking condition.

In summary, the reliability of some EMG variables related to timing were acceptable for between- and within-session analysis. However, most amplitude characteristics displayed unacceptable levels of error for between-session analysis for several muscles. Overall, surface and intramuscular electrodes were associated with stable electromyography for within-session reliability, however unacceptable error for between-session reliability. The results of the study indicated that it was not possible to conduct subsequent studies with a repeated measures design reliably (i.e. where the electrodes are removed and reattached).

8.2.3 Effect of foot posture on selected lower limb muscles during gait

In Chapter 6, the differences in EMG activity comparing people with normal- and flat-arched feet were large enough to detect statistically significant differences. This was despite the large magnitude of random error reported in Chapter 5, and despite adopting the MVIC normalisation approach (which increases between-participant variability compared to submaximal normalisation). The finding that people with flat-arched feet displayed greater EMG amplitude for tibialis posterior and tibialis anterior, and lower EMG amplitude for peroneus longus was similar to the findings of some studies included in the systematic review presented in Chapter 2. However, the unique aspect of the findings presented in this thesis is that they are based on a considerably larger sample size than has been previously studied (n=60 compared to 18 to 43 in previous studies). Furthermore, other than the early descriptive work of

Gray and Basmajian in 1968 [161], this study provides the only evidence that foot posture influences tibialis posterior muscle activation in adults without pain or dysfunction, and that the differences in muscle activity are specific to different phases of the gait cycle.

While these findings suggest that a relationship exists between foot posture and EMG muscle activity, they do not explain the cause and effect of this relationship (i.e. it cannot necessarily be inferred that flat-arched feet lead to changes in muscle activity). Foot posture is also strongly influenced by some systemic conditions that cause muscle dysfunction, such as neurological [7] and rheumatological disease [163]. It is possible, although less likely, that altered muscle activity in healthy people, without disease or pain, may lead to changes in foot posture.

In Chapter 6, it was suggested that the differences in muscle activity in people with flat-arched feet (i.e. greater EMG amplitude for tibialis posterior and tibialis anterior, and lower EMG amplitude for peroneus longus) may reflect some level of neuromuscular compensation to reduce overload of soft tissue structures that support the medial longitudinal arch. As the flat-arched foot generally exhibits less joint congruency at articulations such as the talo-navicular joint (discussed in Chapters 1 and 3), the neuromuscular activity observed in this study may reflect a strategy to enhance joint congruency, increase joint stability and reduce stress in tissues that support this joint. While it is uncertain whether the differences in muscle function observed with flat-arched feet serve to protect against musculoskeletal injury, another mechanism by which the participants with flat-arched feet may have altered their EMG muscle activity was by simply walking faster. It could be speculated that a faster walking velocity may provide a strategy for participants to increase tibialis posterior and tibialis anterior EMG amplitude as a means of increasing stability of the medial longitudinal arch – since EMG amplitude of some lower limb muscles increases linearly with walking speed [164-166]. The finding that participants with flat-arched feet walked slightly faster in this study, therefore, is novel and is worthy of further examination.

8.2.4 Effect of prefabricated and customised foot orthoses on the electromyographic activity of selected lower limb muscles during gait

The final study presented in this thesis (Chapter 7) investigated whether prefabricated and customised foot orthoses (FOs) alter muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet. While participants showed small to moderate changes for tibialis posterior and peroneus longus, only the prefabricated FOs altered peroneus longus EMG amplitude in the midstance/propulsion phase to a pattern closer to that observed in people with normal-arched feet. Figure 8.1 below illustrates an overview of the significant differences in EMG activity found between normal- and flat-arched feet and the direction of change observed with the use of foot orthoses.

Figure 8.1. Effect of foot posture and foot orthoses on muscle activity in people with flat-arched feet.

Muscle	Phase of gait cycle	Difference in EMG amplitude observed for flat-arched foot relative to normal-arched foot	Difference in EMG amplitude observed with foot orthoses (i.e. relative to no foot orthoses)	Do foot orthoses change muscle activity in people with flat-arched feet towards a pattern observed in people with normal-arched feet?
Tibialis posterior	Contact phase	—	↓	No
	Midstance/ propulsion phase	↑	—	No
Peroneus longus	Contact phase	↓	—	No
	Midstance/ propulsion phase	↓	↑	Yes
Tibialis anterior	Contact phase	↑	—	No

KEY

— No significant differences in EMG amplitude

↑ Significant increase in EMG amplitude

↓ Significant decrease in EMG amplitude

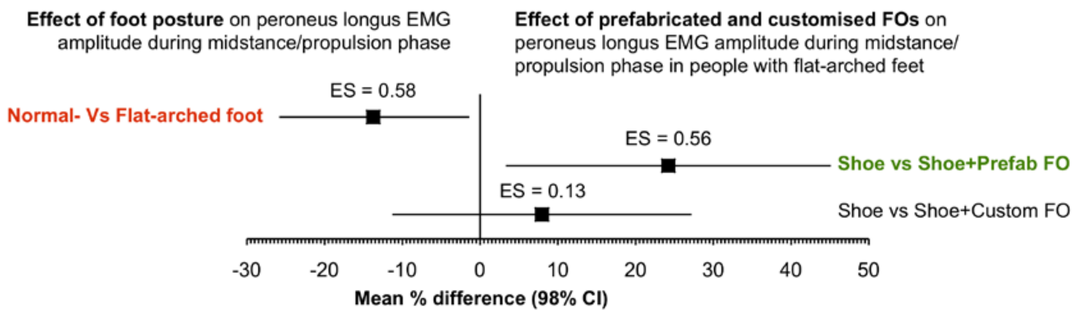
Yes Muscle activity changed toward a pattern observed in people with normal-arched feet

No Muscle activity did not changed toward a pattern observed in people with normal-arched feet

The prefabricated and custom FOs were issued to participants two weeks prior to EMG testing, so the differences observed can be considered to be short- to intermediate-term responses. It is unclear whether these or further changes may exist when FOs are worn for a longer period of time. However, previous kinematic studies have demonstrated that the immediate effect of FOs on ankle kinematics during running are similar to those observed at three [167] and six weeks [168] in asymptomatic and symptomatic runners, respectively. Therefore, it is unlikely that different observations would be made if the study in Chapter 7 were conducted after a longer habituation period.

As discussed in Section 8.2.2 above, different normalisation approaches were used in Chapter 6 and Chapter 7. In Chapter 6, EMG amplitude data from participants with normal- and flat-arched feet were normalised with MVICs. In contrast, EMG amplitude data comparing the effect of foot orthoses in Chapter 7 were normalised to a sub-maximal normalisation protocol (i.e. relative to the barefoot walking condition). Therefore, the differences in EMG amplitude caused by foot posture and foot orthoses are not directly comparable. Nonetheless, Figure 8.2 below shows that the effect sizes for the differences are opposite and almost equal.

Figure 8.2. Forest plot presenting the effect of flat-arched foot posture and foot orthoses on peroneus longus RMS EMG amplitude during the midstance/propulsion period of stance phase. ES – Effect size.



One factor that may have influenced some of the outcomes of this experiment was the type of FOs and the customisation process that was undertaken for both styles of FOs that were issued to participants. The obvious differentiating feature of the prefabricated and customised FOs in this study was the harder thermo-plastic (polypropylene) material used for the customised FOs compared to the more compliant foam (dual-density polyethylene foam) material in the prefabricated FOs. It could be argued that any other styles or features of FOs, such as a medial heel skive, could have been justifiably matched to the participants' flat-arched feet in this study and could possibly have led to further changes in muscle activity under similar experimental conditions. Foot orthosis prescription for research trials is difficult because there are no universally accepted guidelines for their design or prescription and there are no means of quantifying and reporting, in a standardised way, the mechanical behaviour of the orthoses tested [70,169,170]. Accordingly, reaching a standardised prescription for the customised FO (see Appendix 6) in this study was inherently difficult. From a clinical perspective, the level of rearfoot wedging built-in to customised FOs is generally not standardised, rather it is supposed to be ‘customised’ to provide sufficient support or correction for a particular individual [170,171]. However, contrary to this belief, there is evidence to suggest that different levels of rearfoot orthotic wedging tend to have a similar effect on muscle activity [172]. Furthermore, the participants with flat-arched feet were relatively homogenous in their foot posture characteristics and there was no compelling reason to provide them with different levels of rearfoot correction. With these issues in mind, there is a need for the development of consensus guidelines for the prescription of customised orthoses [169].

8.3 Limitations

The findings of the studies presented in this thesis need to be interpreted in light of several limitations relating to; (i) the generalisability of the sample, (ii) the selection of the foot posture screening protocol, (iii) the gait evaluation protocol, and (iv) inherent limitations in electromyography. Each limitation is discussed in detail below.

8.3.1 Generalisability of the sample

The young and asymptomatic adult participants recruited to these studies provided a convenient cohort for investigating the effect of foot posture and foot orthoses on muscle activity during walking. However, the trade-off for recruiting a relatively homogenous sample without musculoskeletal injury is that the results may not be generalisable to symptomatic or clinical populations. Further research is therefore required to investigate the relationship between foot posture and muscle activity in clearly defined musculoskeletal conditions, such as tibialis posterior tendon dysfunction and medial tibial stress syndrome. However, it may not be feasible to use in-dwelling electrodes to record muscle activity in highly symptomatic individuals due to the potential discomfort associated with the procedure.

8.3.2 Foot posture screening protocol

The availability of valid and reliable methods of classifying foot posture for this research was limited to a few static weightbearing tests. In the clinical setting, the assessment of foot posture often includes a series of additional assessments, such as joint range of motion and visual gait analysis. It is possible that classifying foot posture based on a model that includes dynamic gait characteristics and joint range of motion may have led to greater understanding of the effect of foot posture and foot orthoses on muscle activity during walking. For example, it could be hypothesised that during stance phase, tibialis posterior muscle activity would be influenced by motion of the navicular relative to the talus; that is, greater motion of the navicular may require greater activity from tibialis posterior.

In addition, the foot posture screening protocol outlined in Chapter 3 was not designed to identify those who would benefit or respond to FOs, particularly as the participants were asymptomatic. The normative data used to formulate the inclusion values in the foot screening protocol were also not derived from a prospectively constructed random sample of people. Therefore, the reference values may vary to some degree from actual population characteristics.

8.3.3 Gait evaluation protocol

Another limitation to the studies presented in this thesis was that the gait trials only involved participants walking, un-fatigued and at their self-selected velocities. Although various significant findings were presented in Chapters 6 and 7, it would be worthwhile repeating these studies following real-life fatiguing activities, such as extended periods of walking. For example, fatiguing the intrinsic foot muscles with isometric exercises has been shown to increase foot pronation (assessed by navicular drop) [173]. It could be suggested that this may place greater demand on muscles such as tibialis posterior and tibialis anterior to resist pronation of the medial longitudinal arch. Furthermore, a protocol that investigates the effect of foot orthoses on lower limb muscle activity with fatiguing exercises, such as extended periods of walking and running, may provide further insights about the effect of this intervention.

8.3.4 Electromyography limitations

In addition to the reliability issues discussed in section 8.2.2 above, EMG has other inherent limitations than require some discussion. Firstly, the use of intramuscular electrodes to assess tibialis posterior and peroneus longus often lead to participant discomfort. Participants usually described low to mild discomfort during the insertion procedure with approximately 1 in 20 describing severe pain, although this was not quantified. When participants experienced severe pain, the wires were removed and a second attempt at relocating new wire electrodes was undertaken. During walking, participants usually described ‘mild’ pain for the first couple of minutes (i.e. during the warm-up period), which frequently subsided to ‘no’ or ‘low’ pain after this period. Accordingly, the discomfort caused by the indwelling electrode wires may have perturbed the participants’ normal walking pattern and therefore influenced muscle activity.

Secondly, it remains unclear how changes in EMG amplitude relate to muscle physiological performance and changes in muscle action (i.e. eccentric, concentric and isometric). Other modalities are now available for assessing lower limb muscle activity, such as magnetic resonance imaging (MRI) [174] and dynamic ultrasound [175]. Currently though, these modalities have very limited capability to assess gait. Nevertheless, as these technologies develop further and become more readily available for biomechanical research, further insights about the effect of foot posture and foot orthoses on muscle function may become apparent. Ultimately, such modalities may preclude the need to use invasive indwelling electrodes to assess deep muscle activity during gait.

8.4 Indications for future research

The findings that foot posture and foot orthoses influence lower limb EMG muscle activity during gait provide a basis for further research. Firstly, there is a case for repeating the gait experiments presented in Chapter 6 (effect of foot posture) and Chapter 7 (effect of foot orthoses) under running gait conditions. An investigation focused on running gait would provide some insight into the fatigue characteristics of lower limb muscles in people with flat-arched feet, and whether such characteristics could be modified with foot orthoses.

Secondly, further research should be undertaken to investigate whether having flat-arched feet is associated with physical differences in muscle and tendon morphology, compared to having normal-arched feet. For example, it could be hypothesised that because people with flat-arched feet displayed greater EMG amplitude for tibialis posterior during midstance/propulsion phase of gait compared to those normal-arched feet, the tibialis posterior muscle and tendon may also undergo hypertrophic adaption to meet the demands of supporting the medial longitudinal arch. As mentioned in Section 8.3.4 above, some developments have already occurred in this area with the use of MRI [174] and ultrasonography [175] to assess muscle function.

Thirdly, future research will need to pay greater attention to the issue of between-participant variability in motor control. There is growing evidence that dynamic movement of the foot [80] and lower limb muscle activation [21] are highly variable between individuals. There is, therefore, a need for the development of an evidence-based and patient-specific model of foot biomechanics [80]. Such a model may enable a more evidence-based approach to prescription of foot orthoses [80].

Finally, the greatest challenge for future research in this field is to determine whether differences in muscle activity comparing foot posture and foot orthoses are clinically important. To address this, prospective studies and clinical trials will need to be performed to ascertain whether these EMG patterns are related to risk of injury and reduction in symptoms, respectively.

8.5 Conclusion

The findings of these projects have addressed the aim to determine the relationship between foot posture, foot orthoses and lower limb electromyographic activity during walking. This thesis has contributed to the body of knowledge in this discipline by investigating the reliability of EMG in gait research and by demonstrating that foot posture and foot orthoses significantly influence lower limb muscle activity during walking.

There were two main findings of clinical importance from the studies undertaken in this thesis. Firstly, people with flat-arched feet exhibit greater EMG amplitude for tibialis posterior and tibialis anterior, and decreased EMG amplitude for peroneus longus during specific phases of the walking gait cycle, compared to people with normal-arched feet. Secondly, foot orthoses have a significant effect on tibialis posterior and peroneus longus in people with flat-arched feet. However, the only EMG parameter that changed towards a pattern observed in normal-arched feet was peroneus longus EMG amplitude during the midstance/propulsion phase.

Further research, however, is required to determine whether differences in muscle function between normal-arched and flat-arched feet are associated with injury, and whether changes in muscle function with foot orthoses are associated with clinical outcomes.

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APPENDICES

Appendix 1 – University ethics confirmation

La Trobe University
Faculty of Health Sciences
MEMORANDUM

TO: Dr Karl Landorf
Dr Hylton Menz
School of Human Biosciences

SUBJECT: *Reference:* **FHEC06/205**

Student or Other Investigator: George Murley, James Wickham, Adam Bird, Mark Whiteside

Title: **The effect of foot posture on EMG activity of the tibialis posterior muscle**

DATE: 23 March 2007

The Faculty Human Ethics Committee (FHEC) has considered and approved the above project. You may now proceed.

Please note that Informed Consent forms need to be retained for a minimum of 5 years. Please ensure that each participant retains a copy of the Informed Consent form. Researchers are also required to retain a copy of all Informed Consent forms separately from the data. The data must be retained for a period of 15 years.

Please note that any modification to the project must be submitted in writing to FHEC for approval. You are required to provide an annual report (where applicable) and/or a final report on completion of the project. A copy of the progress/final report can be downloaded from the following website:
www.latrobe.edu.au/rgso/forms-resources/forms/ethic-prog-final.rtf.

Please return the completed form to The Secretary, FHEC, Faculty of Health Sciences Office, La Trobe University, Victoria 3086.

A copy of this memorandum is enclosed for you to forward to the student(s) concerned.

Natalie Humphries
Secretary
Faculty Human Ethics Committee
Faculty of Health Sciences

Appendix 2 – Radiation Safety Committee of the Victorian Department of Human Services approval and registration of project

Department of Human Services

Incorporating: Health, Community Services, Aged Care and Housing

50 Lonsdale Street
GPO Box 4057
Melbourne Victoria 3001
DX210081
www.dhs.vic.gov.au
Telephone: 1300 650 172
Facsimile: 1300 785 859

Our Ref:
Your Ref:

27 March 2007

George Murley
Department of Podiatry
La Trobe University
Bundoora VIC 3086

Dear Mr Murley,

Re: Notification of the research project:

The effect of foot posture on the EMG activity of the tibialis posterior muscle

I refer to your correspondence regarding the above research project and the research proposal obtaining Human Research Ethics Committee approval.

Based on this approval, in accordance with the Code of Practice for the Exposure of Humans to Ionizing Radiation for Research Purposes (2005), I approve of the research project being commenced, subject to any other required approvals.

The licence for research with human volunteers has been amended to include the above project. A copy of the licence has been forwarded to Radiation Safety Officer of Latrobe University . This person is listed as the one responsible for the administration of this licence. Please advise the Radiation Safety Section when this project is complete.

Should you have any further queries regarding this matter please do not hesitate to contact by phone on Mr Julian Marwick on (03) 9096 5676 or by email at julian.marwick@dhs.vic.gov.au.

Yours sincerely,

DR BRAD CASSELS
MANAGER
RADIATION SAFETY PROGRAM

State Government Victoria

Appendix 3 – Participant consent and withdrawal of consent forms

DEPARTMENT OF PODIATRY
Electromyography Laboratory

Consent for project title:

PROJECT 1: THE EFFECT OF FOOT POSTURE ON EMG ACTIVITY OF THE TIBIALIS POSTERIOR MUSCLE

Primary Investigator:	MR GEORGE MURLEY, ASSOCIATE LECTURER, DEPARTMENT OF PODIATRY, SCHOOL OF HUMAN BIOSCIENCES, LA TROBE UNIVERSITY
Other Researchers:	DR KARL LANDORF, SENIOR LECTURER, DEPARTMENT OF PODIATRY, SCHOOL OF HUMAN BIOSCIENCES, LA TROBE UNIVERSITY
	DR HYLTON MENZ, SENIOR RESEARCH FELLOW, SCHOOL OF PHYSIOTHERAPY, LA TROBE UNIVERSITY
	DR JAMES WICKHAM, LECTURER, DEPARTMENT OF ANATOMY AND PHYSIOLOGY, SCHOOL OF HUMAN BIOSCIENCES, LA TROBE UNIVERSITY
	DR ADAM BIRD, LECTURER, DEPARTMENT OF PODIATRY, SCHOOL OF HUMAN BIOSCIENCES, LA TROBE UNIVERSITY
	MR MARK WHITESIDE, UNDERGRADUATE PODIATRY HONOURS STUDENT, DEPARTMENT OF PODIATRY, SCHOOL OF HUMAN BIOSCIENCES, LA TROBE UNIVERSITY

The researchers conducting this experiment support the principles governing both the ethical conduct of research, and the protection at all times of the interests, comfort and safety of participants.

This form and the accompanying Subject Information Package are given to you for your own protection. They contain a detailed outline of the experimental procedures, and possible risks. Your signature below indicates you understand and consent to the following items.

(1) You have received the Subject Information Package.
(2) You have been given the opportunity to discuss the experiment (with one of the researchers prior to commencing).
(3) You clearly understand the responsibilities and risk.
(4) You voluntarily agree to take part in the project.
(5) Your participation may be terminated at any point in time without jeopardising your involvement at La Trobe University.
(6) You are aware of the relative risk of exposure to ionising radiation during the X-ray screening procedure.
(7) You are aware of any adverse reactions that may occur as a result of a needle puncture.
(8) You are not currently taking blood thinning medication (i.e. Warfarin, Asprin)

Last Name:___________________________

Given Name:__________________________

Age: ______________ Phone No (H): _______________________ (M): __________________

Address: ____

Name and phone number of a contact person in case of an emergency:

Name:___________________________________ Phone: ___________________________________

Family Doctor:______________________________ Phone: ___________________________________

Subject Signature:_________________________ Date: __________

Witness: _________________________

Witness Signature: _________________________

DEPARTMENT OF PODIATRY
Electromyography Laboratory

Consent for project title:

PROJECT 2: THE EFFECT OF FOOT POSTURE ON EMG ACTIVITY OF THE TIBIALIS POSTERIOR MUSCLE

Primary Investigator: **MR GEORGE MURLEY**, ASSOCIATE LECTURER, DEPARTMENT OF PODIATRY, DIVISION OF ALLIED HEALTH, LA TROBE UNIVERSITY

Other Researchers: **DR KARL LANDORF**, SENIOR LECTURER, DEPARTMENT OF PODIATRY, DIVISION OF ALLIED HEALTH, LA TROBE UNIVERSITY
A/PROF HYLTON MENZ, DIRECTOR, MUSCULOSKELETAL RESEARCH CENTER, LA TROBE UNIVERSITY
DR JAMES WICKHAM, LECTURER, DEPARTMENT OF ANATOMY AND PHYSIOLOGY, SCHOOL OF HUMAN BIOSCIENCES, LA TROBE UNIVERSITY
MR ADAM BIRD, LECTURER, DEPARTMENT OF PODIATRY, DIVISION OF ALLIED HEALTH, LA TROBE UNIVERSITY
MS LISA SCOTT, UNDERGRADUATE PODIATRY HONOURS STUDENT, DEPARTMENT OF PODIATRY, DIVISION OF ALLIED HEALTH, LA TROBE UNIVERSITY
MS BIANCA DAVID, UNDERGRADUATE PODIATRY HONOURS STUDENT, DEPARTMENT OF PODIATRY, DIVISION OF ALLIED HEALTH, LA TROBE UNIVERSITY

The researchers conducting this experiment support the principles governing both the ethical conduct of research, and the protection at all times of the interests, comfort and safety of participants.

This form and the accompanying Subject Information Package are given to you for your own protection. They contain a detailed outline of the experimental procedures, and possible risks. Your signature below indicates you understand and consent to the following items.

(1) You have received the Subject Information Package.
(2) You have been given the opportunity to discuss the experiment (with one of the researchers prior to commencing).
(3) You clearly understand the responsibilities and risk.
(4) You voluntarily agree to take part in the project.
(5) Your participation may be terminated at any point in time without jeopardising your involvement at La Trobe University.
(6) You are aware of the relative risk of exposure to ionising radiation during the X-ray screening procedure.
(7) You are aware of any adverse reactions that may occur as a result of a needle puncture.
(8) You are not currently taking blood thinning medication (i.e. Warfarin, Asprin).

Last Name:___________________________

Given Name:__________________________

Age: _____________ Phone No (H): _______________________ (M): __________________

Address: ____

Name and phone number of a contact person in case of an emergency:

Name:__________________________________ Phone: __________________________________

Family Doctor:___________________________ Phone: __________________________________

Subject Signature:________________________ Date: _________

Witness:_____________________________ Witness Signature:________________________

La Trobe University
University Human Ethics Committee

Withdrawal of Consent for Use of Data Form

This form is to be used by participants who wish to withdraw consent for the use of unprocessed research data.

Project Title:
THE EFFECT OF FOOT POSTURE ON EMG ACTIVITY OF THE TIBIALIS POSTERIOR MUSCLE

I, (the participant), wish to WITHDRAW my consent to the use of data arising from my participation. Data arising from my participation must NOT be used in this research project as described in the Information and Consent Form. I understand that data arising from my participation will be destroyed provided this request is received within four weeks of the completion of my participation in this project. I understand that this notification will be retained together with my consent form as evidence of the withdrawal of my consent to use the data I have provided specifically for this research project.

Participant's name (printed):

……………………………………………………………

Signature:

……………………………………………………………

Date:

Appendix 4 – Participant information forms

DEPARTMENT OF PODIATRY
Electromyography Laboratory

PARTICIPANT INFORMATION PACKAGE

ITEM 1: PROJECT TITLE

Project 1: The effect of foot posture on EMG activity of the tibialis posterior muscle; neutral feet

ITEM 2: PROJECT AIM

The aim of this project is to quantify the EMG activity of the tibialis posterior muscle in participants with neutral foot posture.

ITEM 3: RATIONALE

The tibialis posterior muscle is located deep inside the lower leg and the tendon of this muscle inserts into the inner arch of the foot. In some people the tendon undergoes degeneration and this can result in a chronic, painful and debilitating foot condition. To research the effect of tendon degeneration on the tibialis posterior muscle, further study is first needed on healthy people with different foot postures. Therefore, participants with neutral foot posture are required for this study.

ITEM 4: TEST PROCEDURE.

- As a participant in this study you must **not** have:

(i) a history of angina or stroke
(ii) any condition which may limit your ability to walk
(iii) any current or recurring lower limb injury

- You must not be taking blood thinning medication (i.e. Warfarin, Asprin)
- You must also be 18 years old or over.

Screening Procedure – Clinical tests
Only people with neutral foot posture are required for this study. In order to accurately assess your foot posture, three tests will be carried out **before** you can be recruited to participate in the EMG experiment. The first tests involve (i) taking a footprint and measuring the characteristics of the footprint (ii) measuring the height of your arch and the length of your foot. If the measurements collected from these two tests both fall within a range of specific values, then you will be asked to have X-rays taken of your feet by a qualified radiologist at La Trobe University Radiology. This is a final test to ensure your foot posture is suitable for this study.

Screening Procedure – Plain film X-Rays
This research study involves exposure to a very small amount of radiation. As part of everyday living, everyone is exposed to naturally occurring background radiation and receives a dose pf about 2 millisieverts (mSv) each year. All people on earth are exposed to background radiation. Background radiation comes from the sun, the earth, the air and all around us. The ill effects at very high doses of radiation have been well documented, for example increased life threatening cancer rates and sometimes death has been reported in populations exposed to nuclear explosions or in patients undergoing radiotherapy treatment.

However, at tiny or trivial doses of radiation, similar to those being received from being a participant in this research, the risks are not completely known and have to be estimated using theoretical models based on the very high radiation dose data. The acknowledged theoretical model suggests that the risk is less than about 1 in 100,000. This model is based on a conservative approach and the actual risk maybe a lot smaller. Compared to other risks in everyday life this risk is considered negligible. For example: This theoretical risk is approximately the same to smoking less than one quarter of a cigarette, travelling 8 km by car, or travelling 80 km by commercial aircraft.

The effective dose from these studies is about 0.003 mSv. At this dose level, no harmful effects of radiation have been demonstrated, as any effect is too small to measure. The risk is believed to be minimal.

The X-rays will be evaluated by the primary investigator after which you will be contacted by phone and informed whether your foot posture is suitable for the EMG experiment.

EMG Experiment
Participants in this study will be required to attend two EMG testing sessions. The testing sessions will be approximately two weeks apart. The EMG experiment will take place at La Trobe University, School of Human Biosciences Biomechanics Laboratory in the Health Sciences building 2 (level 1). Participants will be required to wear shorts for the application of the electrodes and the duration of the experiment. To ensure that you can walk in your normal manner during the EMG experiment, you must not undertake strenuous exercise (that could potentially cause muscle soreness) within four days of your scheduled testing session.

Testing will involve the application of 4 miniature surface electrodes, 3 intramuscular wire electrodes placed in lower leg muscles and two paper-thin electrodes on the sole of the foot. The small area of skin used for surface electrode attachment will be swabbed with an alcohol solution, shaved and lightly abraded with sandpaper (low grade sandpaper). The intramuscular wire electrodes will be inserted beneath (35 – 40 mm) the skin using a small unused 25 gauge sterilised hypodermic needle. This procedure is undertaken with the aid of an ultrasound unit to guide the insertion of the needle. Once this needle/wire is located within the muscle, the needle is withdrawn and discarded whilst the wires (1/4 mm in diameter) stay within the muscle for the duration of the experiment.

To ensure the correct position of the electrode wires within the tibialis posterior muscle, a muscle stimulator is connected to the wire electrodes to lightly stimulate the muscle. The muscle stimulator creates an involuntary muscle contraction and this is not a painful experience. The electrodes are then connected to a light EMG device which is clipped to a tool belt around your waist. The EMG equipment is connected to the computer via a long narrow cable.

During the experiment, you are required to walk along an 8 metre flat walkway whilst your muscle EMG is recorded on the computer. You will be instructed to perform the following walking tasks (5 trials of each); a very slow walking pace, natural pace (self selected), a brisk pace, very fast pace. You will also complete trials in standard footwear and running shoes, in standard footwear with additional medial heel wedging, with lateral heel wedging, with lateral forefoot wedging, with foot orthoses, with an ankle brace. Finally you will walk at a self selected speed on a treadmill.

These walking speeds and testing conditions are not uncomfortable or strenuous. Toward the end of the testing procedure, you will be required to conduct maximal voluntary contractions of the muscles being tested to provide a baseline against which the testing signal can be compared. The total duration of the EMG experiment will be up to 3 hours. Only you and the research investigators will be present during the testing session.

ITEM 5: OTHER RISKS AND DISCOMFORTS.

The pain of the needle insertion is the normal pain resulting from a needle puncture and usually only lasts a few seconds (where the pain is minimal and transitory). There may be slight bleeding from the needle puncture and subsequently, minor bruising and redness around the electrode insertion site can occur and last for a couple of days. As with any needle puncture into human skin, there is a very low risk of developing infection (at the site of needle insertion). As the investigators will use fresh sterilised needle/electrodes and insert the needle under sterile conditions, the risk of infection is minimised. This procedure will not cause injury that would prevent you from participating in normal activities. Please note that a first aid officer is on call if any adverse events occur during testing.

ITEM 6: ENQUIRIES

Questions concerning the procedure and / or the rationale used in the investigation are welcome at any time. Please ask for clarification of any information which you feel is not explained to your satisfaction. Your initial contact is the person conducting the experiment (George Murley, Phone 9479 5776).

ITEM 7: FREEDOM OF CONSENT

Participation in this project is entirely voluntary. You are free to deny consent before or during the experiment. In the latter case, such withdrawal of consent should be performed at any time you specify, and not at the end of a particular trial. Your participation and / or withdrawal of consent will not influence your present or future involvement at La Trobe University. In the case of student involvement, it will not influence grades awarded by the university. You have the right to withdraw from the experiment, and the right shall be preserved over and above the goals of the experiment. The informed consent forms (one of which you are reading now) will be stored in a locked filing cabinet in the primary investigator's office. No one apart from the primary researcher and the senior investigator will have access to the computer records and written records. The raw data and consent forms will be destroyed in five years time after collection. Participants will be compensated $30.00 for participating in the EMG experiment.

ITEM 8: PUBLICATION OF RESULTS.

It is possible that results from this experiment will be displayed in a thesis format, presented at a conference, or published in a peer reviewed journal. It should be noted however that subject information would be expressed anonymously (e.g. subject 1, subject 2, etc), with no mention of the subjects names or personal details. The EMG data collected in this project will be compared to EMG data collected in future EMG studies. Furthermore results of the experiment will be made available to each subject upon request. This may entail a mailing of results to your home residence, or if you prefer, a discussion with George Murley in person.

ITEM 9: FUNDING BODY.

Funding for this project has been obtained from the Health Sciences Postgraduate Support Grant, and has been sought from the Australian Podiatry Education and Research Fund (APERF) and the Faculty Health Sciences Starter Grant.

ITEM 10: COMPLAINTS

Further enquires and/or complaints should be addressed to The Secretary, Health Sciences Human Ethics Committee, La Trobe University, Victoria, 3086, Telephone 9479 3573

LA TROBE

DEPARTMENT OF PODIATRY
Electromyography Laboratory

PARTICIPANT INFORMATION PACKAGE

ITEM 1: PROJECT TITLE

Project 2: The effect of foot posture on EMG activity of the tibialis posterior muscle; flat arched feet

ITEM 2: PROJECT AIM

The aim of this project is to quantify the EMG activity of the tibialis posterior muscle in participants with flat arch foot posture.

ITEM 3: RATIONALE

The tibialis posterior muscle is located deep inside the lower leg and the tendon of this muscle inserts into the inner arch of the foot. In some people the tendon undergoes degeneration and this can result in a chronic, painful and debilitating foot condition. To research the effect of tendon degeneration on the tibialis posterior muscle, further study is first needed on healthy people with different foot postures. Therefore, participants with high arch foot posture are required for this study.

ITEM 4: TEST PROCEDURE.

- As a participant in this study you must **not** have:

(i) a history of angina or stroke
(ii) any condition which may limit your ability to walk
(iii) any current or recurring lower limb injury

- You must not be taking blood thinning medication (i.e. Warfarin, Asprin)
- You must also be 18 years old or over

Screening Procedure – Clinical tests
Only people with flat arch foot posture are required for this study. In order to accurately assess your foot posture, three tests will be carried out **before** you can be recruited to participate in the EMG experiment. The first tests involve (i) taking a footprint and measuring the characteristics of the footprint (ii) measuring the height of your arch and the length of your foot. If the measurements collected from these two tests both fall within a range of specific values, then you will be asked to have some X-ray taken of your feet by a qualified radiologist at La Trobe University Radiology. This is a final test to ensure your foot posture is suitable for this study.

Screening Procedure – Plain film X-Rays
This research study involves exposure to a very small amount of radiation. As part of everyday living, everyone is exposed to naturally occurring background radiation and receives a dose pf about 2 millisieverts (mSv) each year. All people on earth are exposed to background radiation. Background radiation comes from the sun, the earth, the air and all around us. The ill effects at very high doses of radiation have been well documented, for example increased life threatening cancer rates and sometimes death has been reported in populations exposed to nuclear explosions or in patients undergoing radiotherapy treatment. However, at tiny or trivial doses of radiation, similar to those being received from being a participant in this research, the risks are not completely known and have to be estimated using theoretical models based on the very high radiation dose data. The acknowledged theoretical model suggests that the risk is less than about 1 in 100,000. This model is based on a conservative approach and the actual risk maybe a lot smaller. Compared to other risks in everyday life this risk is considered negligible. For example: This theoretical risk is approximately the same to smoking less than one quarter of a

cigarette, travelling 8 km by car, or travelling 80 km by commercial aircraft.

The effective dose from these studies is about 0.003 mSv. At this dose level, no harmful effects of radiation have been demonstrated, as any effect is too small to measure. The risk is believed to be minimal.

The X-rays will be evaluated by the primary investigator after which you will be contacted by phone and informed whether your foot posture is suitable for the EMG experiment.

EMG Experiment
The EMG experiment will take place at La Trobe University, School of Human Biosciences Biomechanics Laboratory in the Health Sciences building 2 (level 1). Participants will be required to wear shorts for the application of the electrodes and the duration of the experiment. To ensure that you can walk in your normal manner during the EMG experiment, you must not undertake strenuous exercise (that could potentially cause muscle soreness) within four days of your scheduled testing session.

Testing will involve the application of 4 miniature surface electrodes, 3 intramuscular wire electrodes placed in lower leg muscles and two paper-thin electrodes on the sole of the foot. The small area of skin used for surface electrode attachment will be swabbed with an alcohol solution, shaved and lightly abraded with sandpaper (low grade sandpaper). The intramuscular wire electrodes will be inserted beneath (35 – 40 mm) the skin using a small unused 25 gauge sterilised hypodermic needle. This procedure is undertaken with the aid of an ultrasound unit to guide the insertion of the needle. Once this needle/wire is located within the muscle, the needle is withdrawn and discarded whilst the wires (1/4 mm in diameter) stay within the muscle for the duration of the experiment.

To ensure the correct position of the electrode wires within the tibialis posterior muscle, a muscle stimulator is connected to the wire electrodes to lightly stimulate the muscle. The muscle stimulator creates an involuntary muscle contraction and this is not a painful experience. The electrodes are then connected to a light EMG device which is clipped to a tool belt around your waist. The EMG equipment is connected to the computer via a long narrow cable.

During the experiment, you are required to walk along an 8 metre flat walkway whilst your muscle EMG is recorded on the computer. You will be instructed to perform the following walking tasks (5 trials of each) at a self selected natural pace. 1) Wearing standard footwear 2) running shoes 3) standard footwear with additional medial heel wedging 4) with lateral heel wedging 5) with lateral forefoot wedging 6) with foot orthoses 7) with an ankle brace. Finally you will walk at a self selected speed on a treadmill.

These walking speeds and testing conditions are not uncomfortable or strenuous. Toward the end of the testing procedure, you will be required to conduct maximal voluntary contractions of the muscles being tested to provide a baseline against which the testing signal can be compared. The total duration of the EMG experiment will be up to 3 hours. Only you and the research investigators will be present during the testing session.

ITEM 5: OTHER RISKS AND DISCOMFORTS.

The pain of the needle insertion is the normal pain resulting from a needle puncture and usually only lasts a few seconds (where the pain is minimal and transitory). There may be slight bleeding from the needle puncture and subsequently, minor bruising and redness around the electrode insertion site can occur and last for a couple of days. As with any needle puncture into human skin, there is a very low risk of developing infection (at the site of needle insertion). As the investigators will use fresh sterilised needle/electrodes and insert the needle under sterile conditions, the risk of infection is minimised. This procedure will not cause injury that would prevent you from participating in normal activities. Please note that a first aid officer is on call if any adverse events occur during testing.

ITEM 6: ENQUIRIES

Questions concerning the procedure and / or the rationale used in the investigation are welcome at any time. Please ask for clarification of any information which you feel is not explained to your satisfaction. Your initial contact is the person conducting the experiment (George Murley, Phone 9479 5776).

ITEM 7: FREEDOM OF CONSENT

Participation in this project is entirely voluntary. You are free to deny consent before or during the experiment. In the latter case, such withdrawal of consent should be performed at any time you specify, and not at the end of a particular trial. Your participation and / or withdrawal of consent will not influence your present or future involvement at La Trobe University. In the case of student involvement, it will not influence grades awarded by the university. You have the right to withdraw from the experiment, and the right shall be preserved over and above the goals of the experiment. The informed consent forms (one of which you are reading now) will be stored in a locked filing cabinet in the primary investigator's office. No one apart from the primary researcher and the senior investigator will have access to the computer records and written records. The raw data and consent forms will be destroyed in five years time after collection. Participants will be compensated $30.00 for participating in the EMG experiment.

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ITEM 9: FUNDING BODY.

Funding for this project has been obtained from the Health Sciences Postgraduate Support Grant, and has been sought from the Australian Podiatry Education and Research Fund (APERF).

ITEM 10: COMPLAINTS

Further enquires and/or complaints should be addressed to The Secretary, Health Sciences Human Ethics Committee, La Trobe University, Victoria, 3086, Telephone 9479 3573

Appendix 5 – X-ray request form

Southern Cross Medical Imaging
Cnr Plenty Rd & Kingsbury Drive
Bundoora
VIC 3083

Dear Radiographer,

_______________________________ is attending for plain film radiographs of both feet as part of a screening procedure in a research study at La Trobe University.

For this research study, we require **bilateral (weight-bearing) anterior-posterior and lateral projections of the participants feet**.

Thank you for your assistance.

Kind regards,

George Murley
Chief Investigator
Department of Podiatry
La Trobe University

*****PLEASE ENSURE PARTICIPANTS ARE FULLY WEIGHTBEARING*******

- **Dorso-plantar projections - x-ray tube angled 15 degrees cephalad and centred at the base of the third metatarsal**

- **Lateral projections - tube angled 90 degrees and centred to the base of third metatarsal.**

Appendix 6 – Customised orthosis prescription form

footwork
PODIATRIC LABORATORY

CUSTOM DESIGN ORTHOTICS PRESCRIPTION FORM

Footwork Podiatric Laboratory 2 / 27-28 Carl Court Hallam 3803
Postal Address: P.O. Box 5218 Hallam 3803
Tel: 1300 66 77 44 Fax: 1300 65 01 83
Email: info@footwork.com.au

PURCHASE ORDER #

PATIENT

Name: ______________________
Age: __________ Sex: __________
Footwear - Type: Dunlop Volley Size: (US) ______

PODIATRIST

EMG RESEARCH STUDY - George Murley
Health Sciences Podiatry Clinic
Corner Kingsbury Drive and Plenty Road
LA TROBE UNIVERSITY BUNDOORA VIC 3086
MOB (0408 341370)

CORRECTION

	L	R
REAR FOOT (TOTAL)	20 °	20 °
FORE FOOT + / - (Varus / Valgus)	°	°

TYPE

TRADITIONAL [ ] CONCAVE WEDGE [X] WEDGE [ ]

ARCH FILL TECHNIQUE

	L	R
INVERTED	[X]	[X]
MID	[ ]	[ ]
MODIFIED ROOT	[ ]	[ ]

OPTIONAL:

	L	R
MINIMAL ARCH FILL	[ ]	[ ]
MAXIMUM ARCH FILL	[ ]	[ ]

CAST MODIFICATIONS:

	L	R
MEDIAL HEEL SKIVE 0° 15°	[ ] mm	[ ] mm
TRI-PLANAR HEEL SHAVE	[ ]	[ ]
NO PLASTER FILL B/W 1 / 5	[ ]	[ ]
CUBOID NOTCH (S / M / L)	L	L
PLANTAR 5TH RAY GRIND	[ ]	[ ]
PLANTAR FASCIAL ACCOMODATION	[ ]	[ ]
PLANTAR FLEXED 1ST RAY ACCOM.	[ ]	[ ]
EXTRA HEEL EXPANSION	[ ] mm	[ ] mm
MEDIAL FLARE	[X]	[X]
MEDIAL WRAP	[ ]	[ ]
OTHER..............................	[ ]	[ ]

SHELL

MATERIAL

Polyprop.	6.0 mm	[ ]	Carbon fibre (Superform)	
	4.5 mm	[ ]	Semi rigid	[ ]
	4.0 mm	[X]	Semi flex	[ ]
	3.0 mm	[ ]		
	Other	[ ]		

EVA TYPE:.......................................

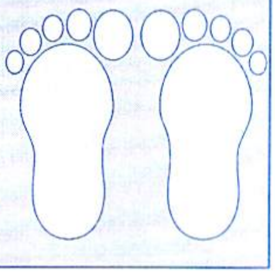
SHAPE

Standard [ ] Low Profile [X] Court [ ]

SHELL MODIFICATIONS

	L	R
HEEL APERTURE (S / M / L)	[ ]	[ ]
PORON BUTTON	[ ]	[ ]
CAMBRELLE BUTTON	[ ]	[ ]
1ST RAY CUT OUT	[ ]	[ ]
LATERAL PLANTAR GRIND	[ ]	[ ]
GAIT PLATE	[ ]	[ ]
INCREASE HEEL PITCH	[ ] mm	[ ] mm
OTHER..............................	[ ]	[ ]
HEEL STABILISER (POST)	[X]	[X]

COMMENTS:



COVER NO COVER

SHAPE

Standard [ ]
Standard PLUS [ ]
Web [ ]
Full [ ]

MATERIAL

Vinyl	[ ]	Poron 1.5	[ ]
PSVlies	[ ]	3.0	[ ]
Agotex	[ ]	3.0 red	[ ]
Lunasoft	[ ]	Spenco 1.5	[ ]
		3.0	[ ]
		Cambrelle base	[ ]

Return Date: *(one day prior to issue)*
/ /

Office Use Only
IN.................. *OUT*.................. *Turn around*..................