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The influence of footwear on the electromyographic activity of selected lower limb muscles during walking

Lisa A. Scott [a,b,c], George S. Murley [a,b,*], James B. Wickham [d]

[a] Department of Podiatry, La Trobe University, Bundoora, Vic., Australia
[b] Lower Extremity and Gait Studies Program, La Trobe University, Bundoora, Vic., Australia
[c] Department of Podiatry, Northern Health, Bundoora, Vic., Australia
[d] School of Biomedical Sciences, Charles Sturt University, Orange, NSW, Australia

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ABSTRACT

The purpose of this study was to compare the effects of a standard flexible shoe and a stability running shoe on lower limb muscle activity during walking. Twenty-eight young asymptomatic adults with flat-arched feet were recruited. While walking, electromyographic (EMG) activity was recorded from tibialis posterior and peroneus longus via intramuscular electrodes; and from tibialis anterior and medial gastrocnemius via surface electrodes. Three experimental conditions were assessed: (i) barefoot, (ii) a standard flexible shoe, (iii) a stability running shoe. Results showed significant differences for the peak amplitude and the time of peak amplitude for tibialis anterior, peroneus longus and medial gastrocnemius when comparing the three experimental conditions ($p < 0.05$). Significant differences were detected primarily between the barefoot and shoe conditions and with relatively small effect sizes for peroneus longus, tibialis anterior and medial gastrocnemius. Few significant differences were found between the two shoe styles. We discuss how these changes are most likely associated with the shoe upper bracing the foot, the shape of the shoe outer-sole and weight of the shoes. Further research is needed to investigate differences between these shoe styles when participants walk for longer distances (i.e. over 1000 m) and following fatigue.

1. Introduction

Over the last two decades, the construction of athletic footwear (footwear used for walking and running) has evolved dramatically to now include a range of materials with properties that are thought to support and cushion the lower limb to assist with locomotion. These features include plastic reinforced heel counters, wedge shaped midsoles of varying densities, and air or gel insoles.

Through the use of different combinations of these features, manufacturers of athletic shoes offer a hierarchy of shoe properties ranging from less to more support that are designed to match individual foot types. Ryan et al. (2011) suggest that such a hierarchy presupposes that high-arched or supinating feet are matched with cushioned shoes; over-pronating flat-arch foot types with stability or motion control (MC) shoes; and those with normal foot types use either neutral or stability shoes. Anecdotally, it is presumed that correctly matching specific foot types with specific levels of support protects individuals from exercise-related injuries, however evidence of this relationship remains unclear and is subject to ongoing research (Ryan et al., 2011).

With this in mind, it is of interest to consider the biomechanical effects (i.e. kinematic, kinetic and electromyographic) associated with using athletic footwear. The findings of kinematic and kinetic studies investigating MC and or stability features show a systematic effect. For example, in running gait these shoes are associated with reductions in mean loading rate, peak rearfoot eversion, eversion excursion and internal tibial rotation when compared to a cushion running shoe (Butler et al., 2006, 2007; Cheung et al., 2011; Clarke et al., 1983; Perry and LaFortune, 1995). Generally, the findings above indicate that stability and MC shoes are associated with reduction in movements that are associated with foot pronation.

Less is known about the effects of MC or stability running footwear on lower limb muscle activity (Murley et al., 2009b). While two recent studies have evaluated the effects of stability footwear on lower leg muscle fatigue characteristics during running (Cheung and Ng, 2009, 2010) there are no published studies that describe the EMG activity during walking gait of important lower limb muscles, such as tibialis posterior, with the use of stability or MC footwear. Therefore, it is not possible to generalise findings from current research to individuals who use stability running shoes

* Corresponding author at: Department of Podiatry, Faculty of Health Sciences, La Trobe University, Bundoora 3086, Vic., 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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for activities of daily living. Accordingly, it is unknown whether stability running footwear systematically alters lower limb muscle activity during walking, compared to standard, less supportive footwear.

The aim of this study, therefore, was to compare lower limb muscle activity during walking in people with excessively pronated feet whilst wearing a commercially available stability running shoe and a standard flexible shoe compared to a barefoot condition.

Our first hypothesis was that the stability running shoe would decrease peak EMG amplitude activation of all muscles tested, compared to the standard flexible shoe and barefoot conditions. Our second hypothesis was that footwear would have no significant effect on timing of peak amplitude.

2. Methods

2.1. Participants

Twenty-eight young adults with flat-arched feet (14 male and 14 female) were recruited to this study (Table 1). The participants were without symptoms of macrovascular disease (e.g. angina, stroke, peripheral vascular disease) and/or neuromuscular disease or any condition that affected their ability to walk. Ethics approval was obtained from the La Trobe University Faculty of Health Sciences Ethics Committee Victoria, Australia (Ethics I.D: FHEC06/205) and it was registered with the Radiation Safety Committee of the Victorian Department of Human Services.

Table 1
Participant anthropometric and foot posture characteristics.

General anthropometric	
Gender ratio (female/male)	14/14
Age mean (SD) years	21.2 (3.8)
Height mean (SD) cm	171.0 (10.0)
Weight mean (SD) kg	73.3 (16.0)
Left or right foot count	15 right 13 left
Clinical measurements[a]	
Arch index[b] ratio (SD) [mean]	0.30 (0.06)
Normalised navicular height to truncated foot length ratio[c] (SD) [mean]	0.17 (0.04)
Radiographic measurements[a]	
CIA mean (SD) degrees	15.7 (4.4)
C1MA mean (SD) degrees	142.5 (6.0)
TNCA mean (SD) degrees	28.3 (8.6)
T2MA mean (SD) degrees	27.7 (9.2)
Walking velocity	
Metres per second (SD) m s^{-1}	1.21 (0.12)

CIA, calcaneal inclination angle; the angle formed by the inferior surface of the calcaneus and the supporting surface; C1MA, calcaneal first metatarsal angle; 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; TNCA, talo-navicular coverage angle; formed by the bisection of the anterior-medial and anterior-lateral extremes of the talar head and the bisection of the proximal articular surface of the navicular; T2MA, talus-second metatarsal angle; 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.

[a] Refer to Murley et al. (2009c) for reference values for normal-arched feet.

[b] 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.

[c] 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.

2.2. Screening protocol

Only participants with flat-arched foot posture were recruited for this study as these individuals are usually directed to purchase a commercial stability type shoe. A screening protocol developed by Murley et al. (2009c) was used to categorise foot posture and included clinical and radiographic measures. 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).

One foot from each participant was included in data analysis in order to satisfy the assumption of independence for data (Menz, 2004). In instances where *both* the participant's left and right feet met the inclusion criteria, the following rule was applied; if both feet were flat-arched, then the 'flatter' foot was used as a reference. Readers are directed to the work of Murley et al. (2009c) for detailed information about the foot posture screening protocol.

2.3. Footwear

Two styles of footwear were used in this investigation and were chosen on pragmatic terms (i.e. we selected shoe styles that we believe are worn by people in the community). Furthermore, we aimed to determine whether more supportive footwear influences muscle activity compared to less supportive and perhaps more fashionable shoes (i.e. ballet flats, canvas style shoes). At present, no studies have compared these frequently worn styles of shoes.

The footwear included in this study was the Dunlop Volley™ (Dunlop Footwear, Victoria, Australia) and the Nike Air Structure Triax +10™ (Nike Incorporated, Oregon, USA), which represented the standard flexible shoe and stability running shoe, respectively. The standard flexible shoe (SFS) used in this experiment had a canvas upper and a thin, flat rubber sole. The test shoe was a stability running shoe (SRS) comprising a range of features aimed at controlling moderate pronation, including a dual-density midsole made of polyurethane (PU) and phylon. The standard flexible and stability running shoes are shown in Figs. 1a and 1b and several shoe properties are described in Table 2.

2.4. Experimental protocol

Participants attended a single testing session in the EMG gait laboratory. 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

Fig. 1a. Dunlop Volley (standard flexible shoe).

Fig. 1b. Nike Air Structure Triax +10 (stability running shoe).

Table 2
Specifications of shoes tested in current study.

Specification	Standard flexible shoe	Stability running shoe
Outsole material	Rubber	Rubber
Midsole material	Nil	Ethylene vinyl acetate (EVA) and polyurethane (PU)
Shock absorption features	Nil	Air sole unit in heel and PU lateral crash pad
Motion-control features	Nil	Dual density medial post and footbridge
Approx. relative heel height	0.5 cm	2 cm
Weight (men's shoe)	360.8 g	367.3 g
	(mean difference: 6.5 g; 95% CI: −20.7–33.7)	
Weight (women's shoe)	323.1 g	334.1 g
	(mean difference: 11 g; 95% CI: −14.9–36.9)	

Differences in shoe weight comparing standard flexible and stability running shoes were not statistically significant, as the 95% confidence intervals for the mean difference included a zero value.

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 sterilised hypodermic needle with the exposed electrode tips bent 3 mm and 5 mm to prevent the contact areas 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 the fibula to the lateral malleolus, starting from the head of the fibula (Leis and Trapani, 2000). This process of fine-wire electrode construction and positioning of wires in vivo was undertaken in accordance with previous work (Murley et al., 2009a,d).

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% silver electrode and an inter-electrode distance (centre-to-centre) of 10 mm. The application of surface electrodes followed the recommendations of SENIAM (Hermens et al., 1999). For tibialis anterior, the surface electrode was placed at approximately 20% of the distance from the tibial tuberosity to the inter-malleoli line, starting from the tuberosity of the 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).

The temporal characteristics of the walking cycle were measured using circular force sensitive resisters (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. Footswitches were adhered to the skin using fixomull® stretch (BSN medical Pty, Ltd.) adhesive fixation and remained in situ for all conditions tested.

During testing, participants first walked barefoot and then under two randomly allocated conditions: (i) stability running shoe or (ii) standard flexible shoe. Participants walked continuously for three minutes with the stability running shoe and the standard flexible shoe just prior to recording, to ensure the participant was comfortable and to reduce any cross-over effect from the respective condition. They were asked 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, with the trial being repeated.

2.5. EMG data processing

The raw EMG signal was passed through a differential amplifier at a gain of 1000 at 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 averaged gait cycles derived from 8 ipsilateral steps). The first two of six trials were not included in the final analysis to further reduce cross-over effects from the respective condition. Two EMG parameters were analysed for each muscle, including; (i) time of peak amplitude; and (ii) peak amplitude. These parameters have been utilised in previous single-session investigations (Murley and Bird, 2006; Murley et al., 2009a,d). The method of EMG processing is presented in Fig. 2. The following phases of the gait cycle were assessed and were based on when these muscles were most active in normal-arch feet: contact and combined midstance/propulsion phase – tibialis posterior and peroneus longus; contact phase – tibialis anterior; and combined midstance/propulsion phase – medial gastrocnemius (Murley et al., 2009a).

2.6. Statistical analysis

To test for differences between conditions, a series of One-Way Repeated Measure ANOVA tests were conducted. 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. Statistically significant univariate F-statistics were evaluated with *post hoc* analysis ($p = 0.05$). The percentage mean difference, 95% confidence intervals and effect sizes were calculated for significant *post hoc* findings.

To provide an estimate of the effect size for each condition, standardised mean differences (SMD = mean difference/pooled standard deviation) were calculated. SMD greater than 1.2 were

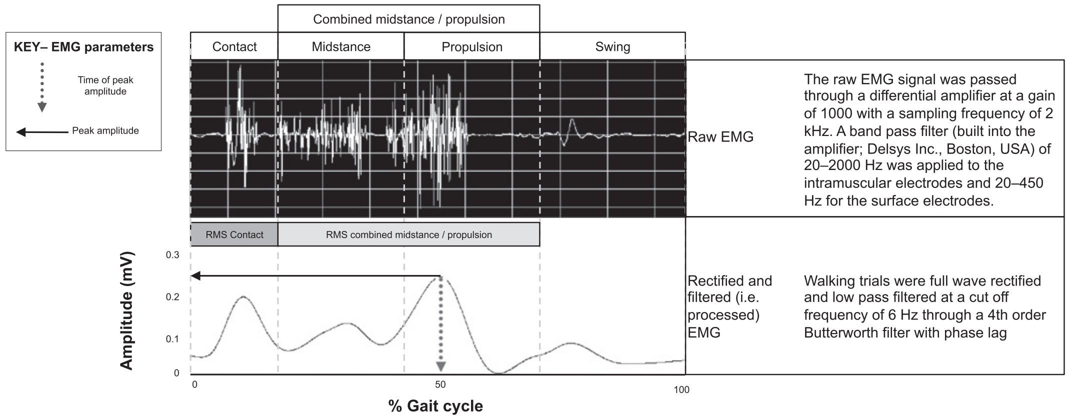

Fig. 2. Raw and processed EMG for tibialis posterior from a single participant outlining schematically how peak amplitude and the time of peak amplitude were calculated.

considered large, 0.6–1.2 moderate and less than 0.6 were considered small (Hopkins et al., 2009b).

3. Results

3.1. Effect of footwear on lower limb muscle EMG activity

During contact phase, significant within participant effects were detected for tibialis anterior time of peak amplitude ($F_{2, 54} = 4.66$, $p = 0.014$) and peak amplitude ($F_{1.57, 42.26} = 26.65$, $p < 0.001$)[1].

During midstance/propulsion phase, significant within participant effects were detected for peroneus longus time of peak amplitude ($F_{2, 54} = 4.00$, $p = 0.024$) and peak amplitude ($F_{1.52, 41.08} = 4.90$, $p = 0.019$).[1] In addition, significant within participant effects were detected for medial gastrocnemius time of peak amplitude ($F_{2, 54} = 13.42$, $p < 0.001$) and peak amplitude ($F_{2, 54} = 3.52$, $p = 0.036$).

Multiple pair-wise comparisons between conditions revealed significant findings for tibialis anterior, peroneus longus and medial gastrocnemius time of peak and peak EMG amplitude, this data is presented in Tables 3 and 4 below. Fig. 3a–d present EMG ensemble averaged tracings (from 28 participants) for all muscles and conditions with statistically significant findings indicated.

Tibialis anterior peak amplitude increased significantly with both shod conditions compared to walking barefoot; and occurred temporally earlier with the flexible shoe condition compared to walking barefoot.

Medial gastrocnemius peak amplitude increased significantly with the running shoe condition compared to walking barefoot; and was temporally delayed with the running shoe condition compared to walking barefoot and the flexible shoe conditions.

Peroneus longus peak amplitude was reduced with both shod conditions compared to walking barefoot; and occurred temporally earlier with the running shoe condition compared to walking barefoot.

4. Discussion

4.1. Effect of footwear on lower limb muscle EMG activity

The aim of this study was to compare lower limb muscle activity during walking in people with excessively pronated feet whilst wearing a commercially available stability running shoe and a standard flexible shoe. The results indicated that both styles of footwear influenced lower limb muscle activity by increasing tibialis anterior and decreasing peroneus longus (EMG) peak amplitude compared to walking barefoot. In addition, the stability running shoe increased medial gastrocnemius EMG amplitude compared to walking barefoot. There were no significant findings detected for tibialis posterior.

4.2. Tibialis anterior

In the contact phase of gait, EMG amplitude of tibialis anterior increased significantly with the standard flexible shoe (21.0% increase) and stability running shoe (24.0% increase) compared to walking barefoot. Additionally, time of peak amplitude occurred significantly earlier with the standard flexible shoe (5.5% of the gait cycle) compared to barefoot (6.0% of the gait cycle). We believe these results are most likely to be related to the weight and the rear projection of the shoe outer-sole of the standard flexible and stability running shoes. The rear projection of the outer-sole may increase the plantar flexion moment at the ankle joint and subsequently increase demand for tibialis anterior, compared to the heel-to-ground dynamics of walking barefoot. In terms of the additional weight of walking shod compared to barefoot, this is expected to increase demand for tibialis anterior to resist the additional weight born from the shoes during contact phase of gait (i.e. when tibialis anterior resists plantar flexion of the ankle joint).

4.3. Peroneus longus

During the midstance/propulsion phase of gait, the stability running shoe and standard flexible shoe significantly decreased peroneus longus peak EMG amplitude by 16.0% and 20.0%, respectively, compared to walking barefoot. In addition, there was a temporal shift toward earlier time of peak amplitude (approximately 2.5% earlier in gait) with the stability running shoe compared to barefoot (Fig. 3).

With respect to the decrease in peak EMG amplitude, we believe this may be related to the stability features of the shoes (medial and lateral support caused by the shoe upper and lacing), compared to walking barefoot. These features may brace the foot and therefore reduce the demand from peroneus longus to assist with medio-lateral stability. The explanation of a bracing mechanism of footwear is supported by a similar study that investigated 3-dimensional kinematics of almost identical footwear (Chuter and Smith, 2011). They reported a significant reduction in peak inversion during walking and running in a dual density midsole running shoe compared to barefoot. The authors suggested there might be less demand on anti-pronatory muscles associated with the use of footwear.

[1] Derived from Greenhouse–Geisser adjusted F-statistic and degrees of freedom.

Table 3
Statistical findings for *time of peak amplitude* for tibialis posterior, tibialis anterior, peroneus longus and medial gastrocnemius.

Muscles	EMG parameter/phase of gait cycle	Condition	Mean value (SD) % gait cycle	Mean difference (% gait cycle)	Comparison (95% confidence interval for difference)	Effect size (standardised mean difference)	*P*-value
Tibialis posterior	Peak amplitude/ midstance-propulsion phase	Barefoot (BF)	45.58 (4.36)	BF vs SFS = −0.34	BF vs SFS (−1.7–1.0)	BF vs SFS = 0.08	0.604
		Standard flexible shoe (SFS)	45.92 (4.39)	BF vs SRS = −0.73	BF vs SRS (−1.9–0.5)	BF vs SRS = 0.17	0.216
		Stability running shoe (SRS)	46.31 (4.12)	SFS vs SRS = −0.39	SFS vs SRS (−1.2–0.4)	SFS vs SRS = 0.09	0.352
Tibialis anterior	Peak amplitude/ contact phase	Barefoot (BF)	6.02 (1.85)	BF vs SFS = 1.7	BF vs SFS (0.1–0.8)	BF vs SFS = 0.31	0.008*
		Standard flexible shoe (SFS)	5.53 (1.58)	BF vs SRS = 2.6	BF vs SRS (0.0–0.7)	BF vs SRS = 0.20	0.069
		Stability running shoe (SRS)	5.67 (1.53)	SFS vs SRS = 0.9	SFS vs SRS (0.0–0.7)	SFS vs SRS = 0.09	0.323
Peroneus longus	Peak amplitude/ midstance-propulsion phase	Barefoot (BF)	50.11 (6.90)	BF vs SFS = −1.0	BF vs SFS (−0.1–3.5)	BF vs SFS = 0.26	0.070
		Standard flexible shoe (SFS)	48.44 (7.37)	BF vs SRS = −2.2	BF vs SRS (0.9–4.2)	BF vs SRS = 0.38	0.004*
		Stability running shoe (SRS)	47.55 (6.02)	SFS vs SRS = −1.2	SFS vs SRS (−1.2–3.0)	SFS vs SRS = 0.14	0.397
Medial gastrocnemius	Peak amplitude/ midstance-propulsion phase	Barefoot (BF)	41.58 (3.59)	BF vs SFS = −1.0	BF vs SFS (−2.0–0.1)	BF vs SFS = 0.28	0.64
		Standard flexible shoe (SFS)	42.56 (3.45)	BF vs SRS = −2.2	BF vs SRS (−3.1 to −1.3)	BF vs SRS = 0.60	<0.001*
		Stability running shoe (SRS)	43.80 (3.39)	SFS vs SRS = −1.2	SFS vs SRS (−1.0 to −0.6)	SFS vs SRS = 0.35	0.001*

* Significant findings.

Table 4
Statistical findings for *peak amplitude* for tibialis posterior, tibialis anterior, peroneus longus and medial gastrocnemius.

Muscles	EMG parameter	Condition	Mean value (SD) (mV)	Mean difference (mV)	95% confidence interval for difference (mV)	Effect size (standardised mean difference)	*P*-value
Tibialis posterior	Peak amplitude	Barefoot (BF)	0.26 (0.19)	BF vs SFS = −0.02	BF vs SFS (−0.05–0.02)	BF vs SFS = 0.08	0.367
		Standard Flexible shoe (SFS)	0.28 (0.18)	BF vs SRS < 0.01	BF vs SRS (−0.03–0.03)	BF vs SRS = 0.00	0.996
		Stability running shoe (SRS)	0.26 (0.19)	SFS vs SRS = 0.02	SFS vs SRS (<−0.01–0.04)	SFS vs SRS = 0.08	0.131
Tibialis anterior	Peak amplitude	Barefoot (BF)	0.09 (0.04)	BF vs SFS < −0.02	BF vs SFS (−0.03–−0.01)	BF vs SFS = 0.57	<0.001*
		Standard Flexible shoe (SFS)	0.12 (0.04)	BF vs SRS = −0.03	BF vs SRS (−0.04 to −0.02)	BF vs SRS = 0.69	<0.001*
		Stability running shoe (SRS)	0.12 (0.04)	SFS vs SRS < −0.01	SFS vs SRS (−0.01–<0.01)	SFS vs SRS = 0.10	0.184
Peroneus longus	Peak amplitude	Barefoot (BF)	0.17 (0.13)	BF vs SFS = 0.03	BF vs SFS (<0.01–0.06)	BF vs SFS = 0.30	0.020*
		Standard Flexible shoe (SFS)	0.13 (0.12)	BF vs SRS = 0.03	BF vs SRS (<0.01–0.05)	BF vs SRS = 0.23	0.033*
		Stability running shoe (SRS)	0.14 (0.10)	SFS vs SRS < 0.01	SFS vs SRS (−0.02–<0.01)	SFS vs SRS = 0.05	0.451
Medial gastrocnemius	Peak amplitude	Barefoot (BF)	0.06 (0.03)	BF vs SFS < −0.01	BF vs SFS (<−0.01–<0.01)	BF vs SFS = 0.11	0.076
		Standard Flexible shoe (SFS)	0.06 (0.03)	BF vs SRS < −0.01	BF vs SRS (<−0.01–<−0.01)	BF vs SRS = 0.14	0.034*
		Stability running shoe (SRS)	0.06 (0.03)	SFS vs SRS < −0.01	SFS vs SRS (<−0.01–<0.01)	SFS vs SRS = 0.03	0.503

* Significant findings.

The response of peroneus longus to medio-lateral stability is demonstrated by a recent study investigating taping and bracing. Franettovich et al. (2011) found peroneus longus peak amplitude decreased whilst participants with a pronated foot posture walked unshod in an ankle brace (33.0% decrease) and augmented low-Dye tape (29.4% decrease). In our study, the finding that peroneus longus peak amplitude is affected by footwear (and not tibialis posterior) may indicate peroneus longus has a larger role in maintaining foot stability during midstance-propulsion. Further research is needed to investigate whether this pattern is altered in those with recurrent ankle instability, given that delayed peroneus longus latency in response to standing perturbation has previously been reported in these individuals (Hopkins et al., 2009a).

4.4. Medial gastrocnemius

During midstance/propulsion period, medial gastrocnemius amplitude increased significantly with the stability running shoe (7.0% increase) compared to barefoot walking. This finding was opposite to what we expected would occur using a shoe with an elevated heel height (relative to forefoot height). That is, we expected medial gastrocnemius EMG amplitude to decrease due to the elevated heel position and subsequent unloading/reduced demand for triceps surae. However, when considering EMG research related to 'high-heeled' shoes, it is well established that extremely elevated heel positions systematically reduce medial gastrocnemius EMG amplitude (Gefen et al., 2002; Lee et al., 1990). With this in mind, it could be suggested that a critical height or threshold is

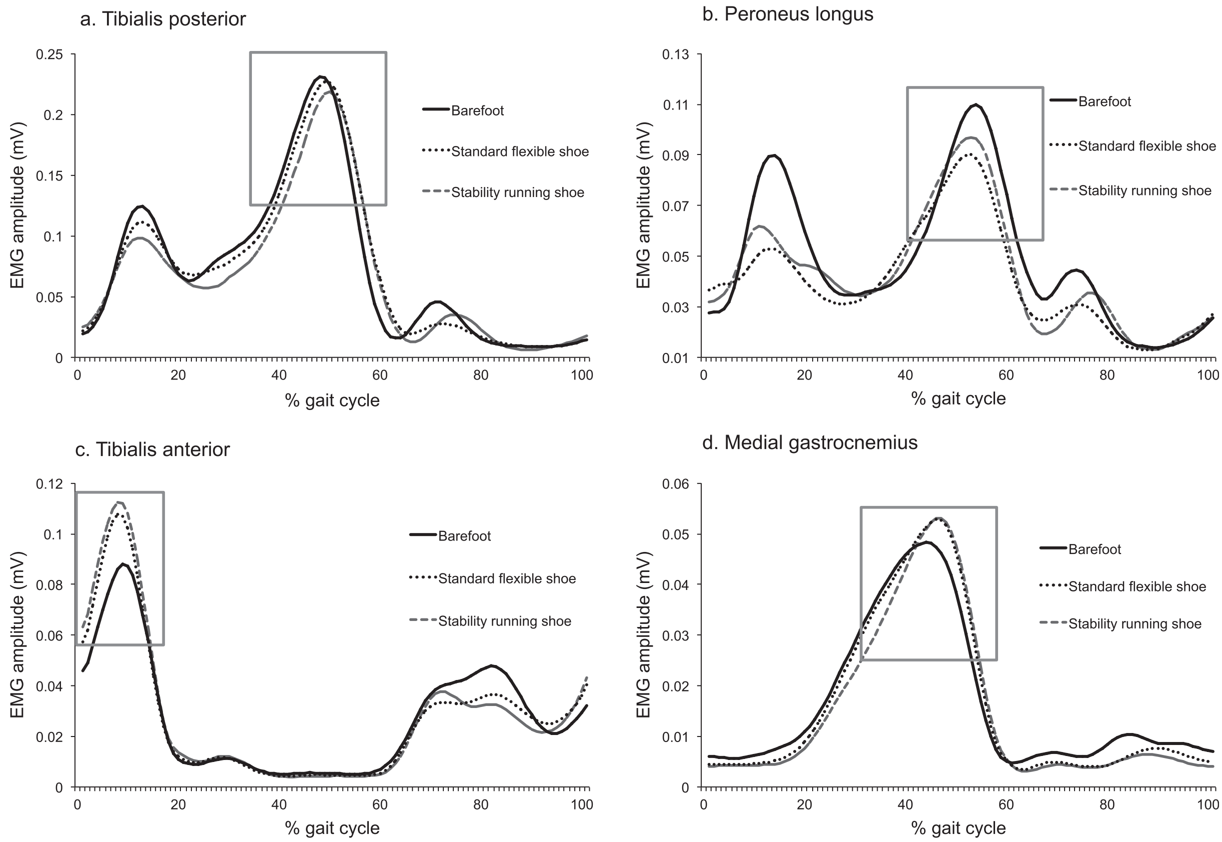

Fig. 3. (a–d) EMG ensembles averages comparing barefoot, standard flexible shoe and stability running shoe for all muscles. Statistical analyses were carried out for the phases shown within the superimposed rectangles. Peak values are less than those presented in the results (Table 4) due to the process of averaging the data to construct the ensemble curves.

needed before medial gastrocnemius is sufficiently unloaded to display reductions in peak amplitude.

With regards to the time of peak amplitude, the stability running shoe significantly delayed time of peak amplitude for medial gastrocnemius (43.8% of the gait cycle) compared to walking in the standard flexible shoe (42.6% of the gait cycle) and walking barefoot (41.6% of the gait cycle). When considering that peak amplitude also increased during this phase of the gait cycle with the stability running shoe, we are uncertain what overall effect this has on the workload undertaken by medial gastrocnemius during propulsion.

4.5. Tibialis posterior

There were no significant findings detected for tibialis posterior comparing the footwear and barefoot conditions. Tibialis posterior is regarded as a dynamic invertor and stabiliser of the medial longitudinal arch (Semple et al., 2009) and in those with a pronated foot posture there is evidence to suggest this muscle has greater EMG activation during walking (Murley et al., 2009b). With this in mind, we hypothesised that the stability running shoe would decrease EMG amplitude of tibialis posterior as a result of the support/bracing features incorporated into the shoe (i.e. rigid heel counter and dual density midsole), however this was not detected in the results. To our knowledge, no study has yet detected significant gait-related changes in tibialis posterior muscle activity using stability or MC footwear.

4.6. Limitations

Whilst the footwear used in this study was chosen on pragmatic terms (as described in the methods section), we recognise that the stability running shoe was primarily designed to suit the mechanics of a running gait, rather than walking. It could be suggested that further differences between the two styles of shoes would be observed if the experiment were repeated under running conditions and or with the addition of other biomechanical measurements (i.e. using kinematic and kinetic instrumentation).

This study did not include an extended habituation (i.e. wear in) period for the participants to adjust to the two styles of footwear in the time leading up to their laboratory testing session, nor did we determine the type of shoes worn regularly by participants. We acknowledge that it would have been useful to collect information about the participants shoe wearing habits. However, it is currently very difficult to accurately characterise the properties and style of footwear in the absence of rigorous standardised procedures. Anecdotally, we believe that these two styles of shoes are worn frequently, and we speculate that participants may have already worn one or both footwear designs prior to the analysis. With these issues in mind, it is possible that findings of this study are affected by the participant's lack of habituation to the shoes investigated.

Furthermore, we did not collect data about our participant's level of physical activity or chosen sport, as we do not believe these variables would necessarily affect how individuals would respond to different styles of footwear. While these results can be generalised to those with flat-arched foot posture, further research is required to investigate whether differences exist in those with normal- and high-arched feet.

5. Conclusion

Supportive shoes such as stability or MC footwear have long been recommended to those with over-pronating or flat-arch foot types. In this study, both styles of footwear significantly altered the EMG activation of leg muscles compared to barefoot walking. However, the only EMG parameter that differed between the shoe styles was medial gastrocnemius time of peak amplitude. We conclude therefore that individuals with flat-arched feet display a similar pattern of lower leg muscle activity when they are walking short distances in either stability or casual/flexible shoes. Clinical trials are required to explore the effects of stability and MC footwear on symptomatic participants with musculoskeletal pathology.

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Lisa Scott is a podiatrist who graduated from La Trobe University with first class Honours in 2008. She is currently completing her Masters degree (Master of Applied Science, Podiatry) at La Trobe University investigating the efficacy of customised foot orthoses in the treatment of Achilles tendinopathy. Lisa combines her work as a podiatrist in private practice and public health.

Dr George S Murley is a Podiatrist who graduated from La Trobe University with Honours in 2002. He later completed his PhD related to the effect of foot posture and foot orthoses on lower limb muscle activity in 2010 and was awarded the Stephen Duckett Higher Degree Research Prize for best PhD thesis in the Faculty of Health Sciences. George has continued to teach at La Trobe University where his research is focused on investigating mechanisms linking foot posture and lower limb injury.

Dr James Wickham is a lecturer in Anatomy and Physiology in the School of Biomedical Sciences at Charles Sturt University, Australia having joined CSU in 2008. Prior to this James received a Bachelor of Sports Science (Exercise Science) from The University of New South Wales in 1992 and a Dip. Ed. (Phy Ed.) in 1993 from the same institution. In 1995 he received a 1st Class Honours Degree from the University of Wollongong and a PhD in 2002. From 1999–2008 James was employed as an Anatomy lecturer at La Trobe University in Melbourne. His research interest is in electromyography and energy expenditure.