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

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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.

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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

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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: FH0606/2005) 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.

<i>General anthropometric</i>	
Gender ratio (female/male)	15/15
Age mean (SD) years	21.8 (4.3)
Height mean (SD) cm	171.0 (10.0)
Weight mean \pm (SD) kg	73.3 (15.5)
Left or right foot count	13 right 17 left
<i>Clinical measurements</i>	
Arch index [^] (SD) [mean]	0.30 (0.07)
Normalised navicular height to truncated foot length ^{**} (SD) [mean]	0.18 (0.04)
<i>Radiographic measurements</i>	
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)
<i>Walking velocity</i>	
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.

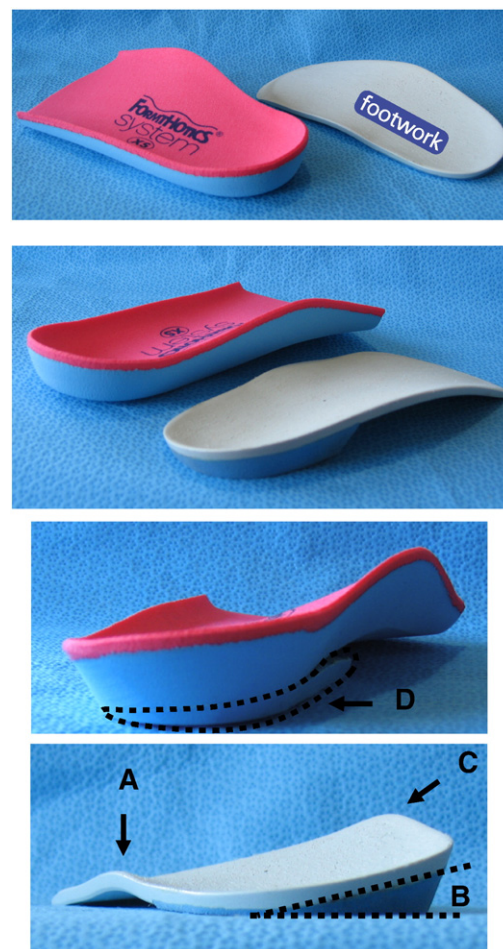


Fig. 1. Modified prefabricated (left) and customised (right) foot orthoses (left foot). Features of the customised orthosis: A – cuboid notch; B – 20° medial wedge; C – medial arch flare; and modified prefabricated orthoses: D – medial heel wedge.

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 perspective. 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,

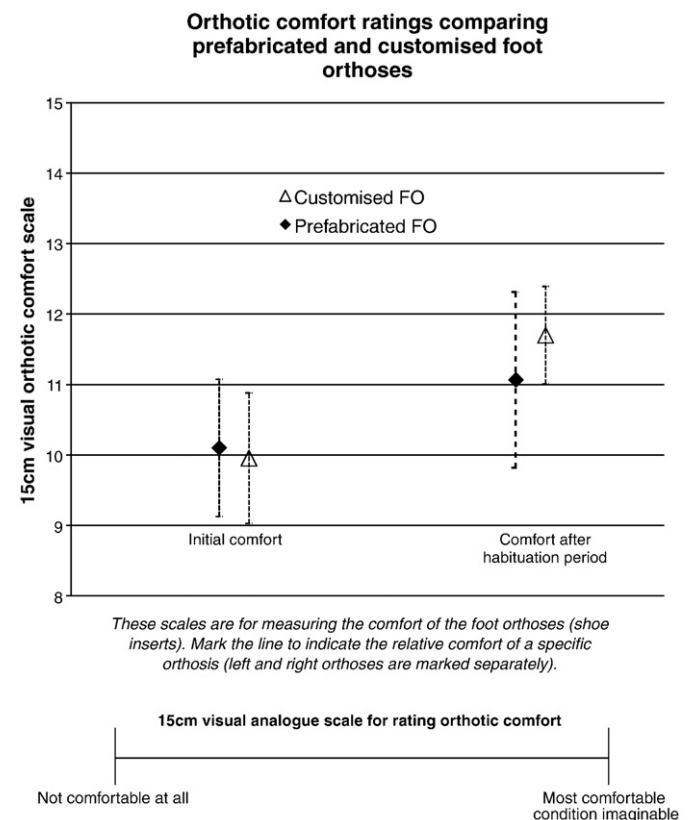


Fig. 2. Foot orthoses comfort scores for the left and right foot recorded during the initial dispensing and following the wear-in period.

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 $\pm 5\%$ 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

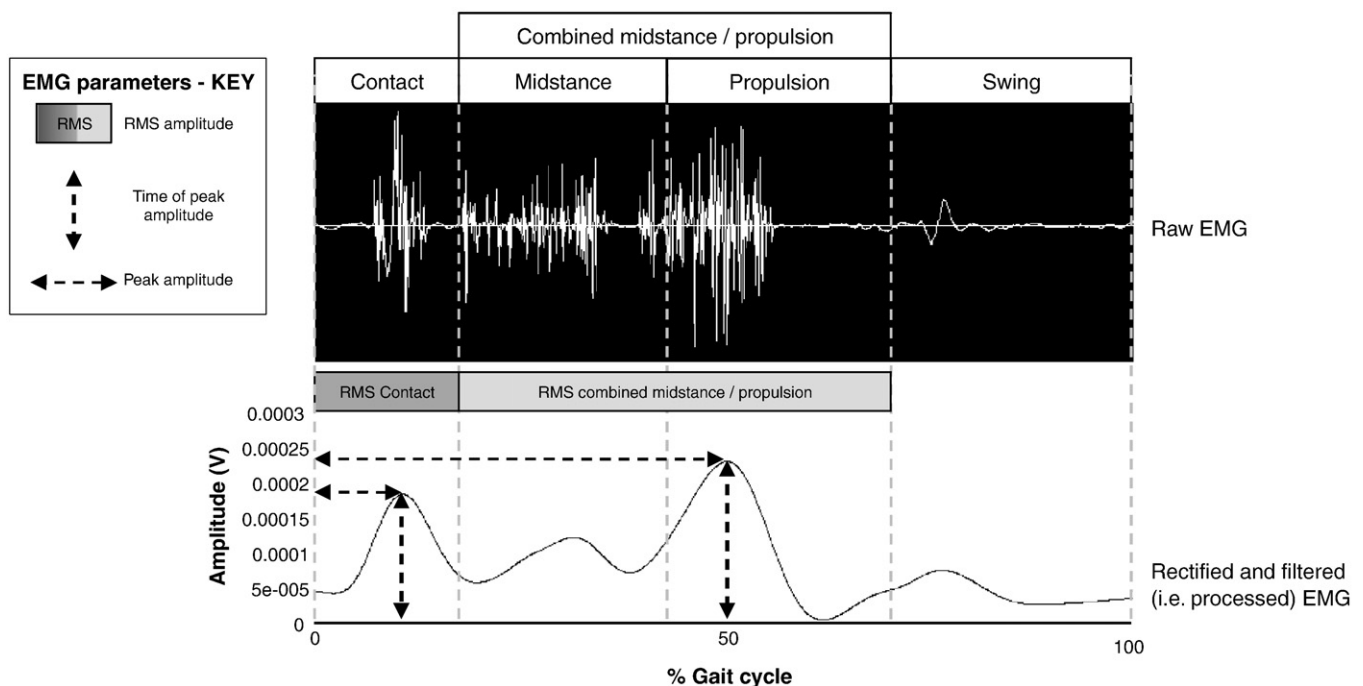


Fig. 3. A single gait cycle showing raw and processed EMG for tibialis posterior from a single participant. Time of peak amplitude, peak amplitude and RMS amplitude (root mean square) were derived from the linear envelope (processed curve).

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).

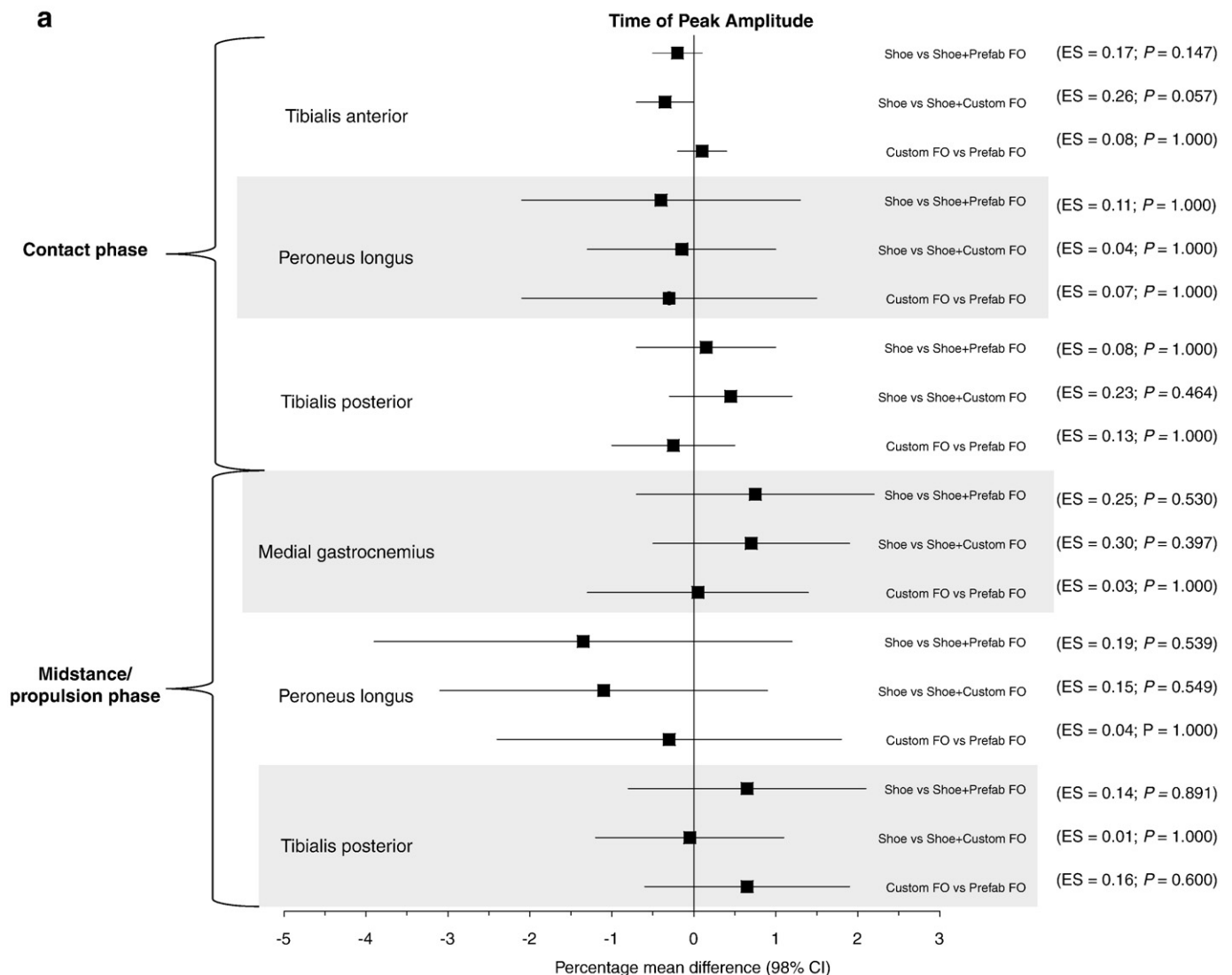


Fig. 4. a–c. Forest plots for *post hoc* comparison between shoe-only condition, customised foot orthoses and modified prefabricated foot orthoses. Bonferroni-adjusted (98%) confidence intervals for multiple comparisons. The change in direction (i.e. positive or negative) of each plot is relative to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

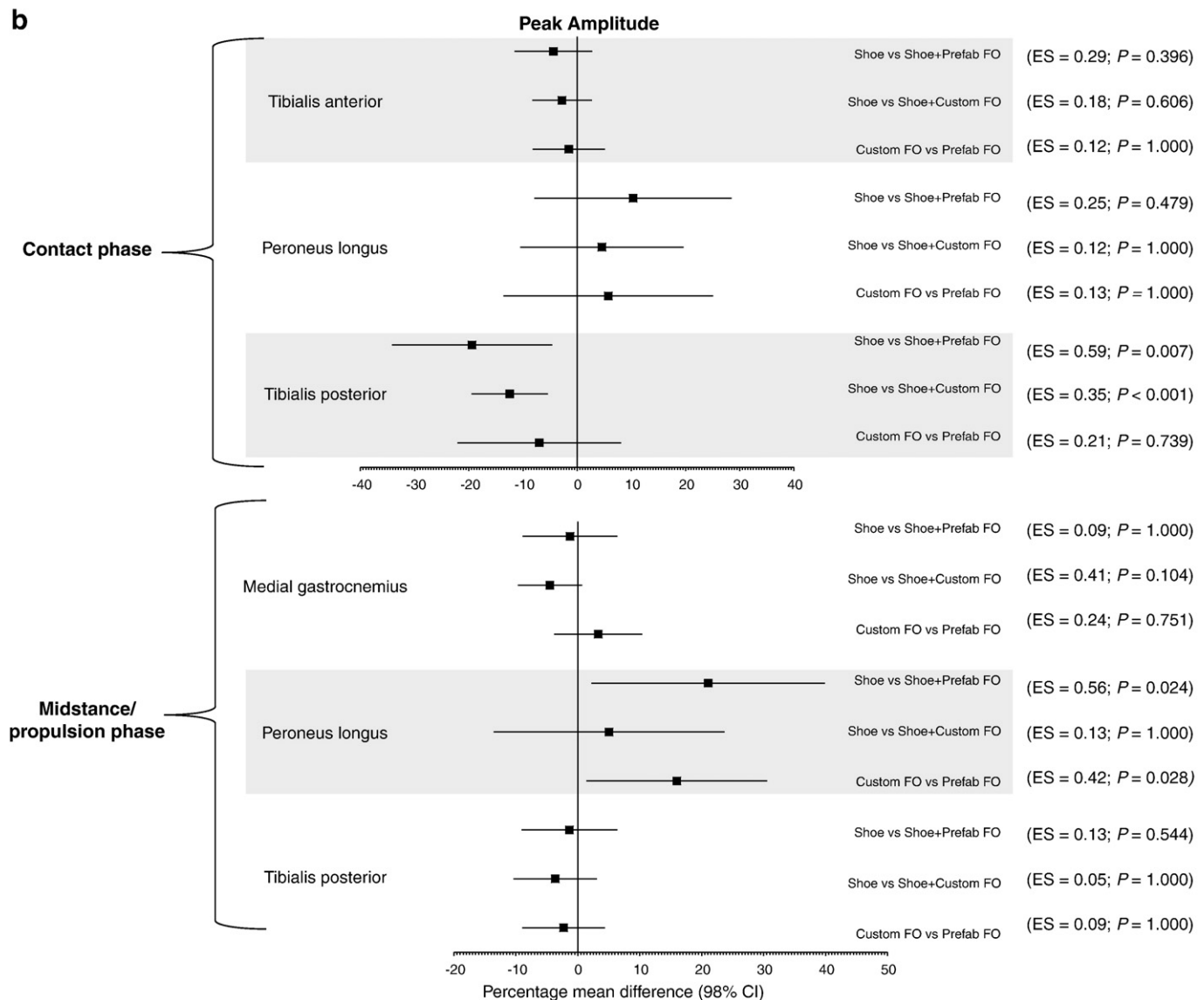


Fig. 4 (continued).

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 differ-

ences 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,

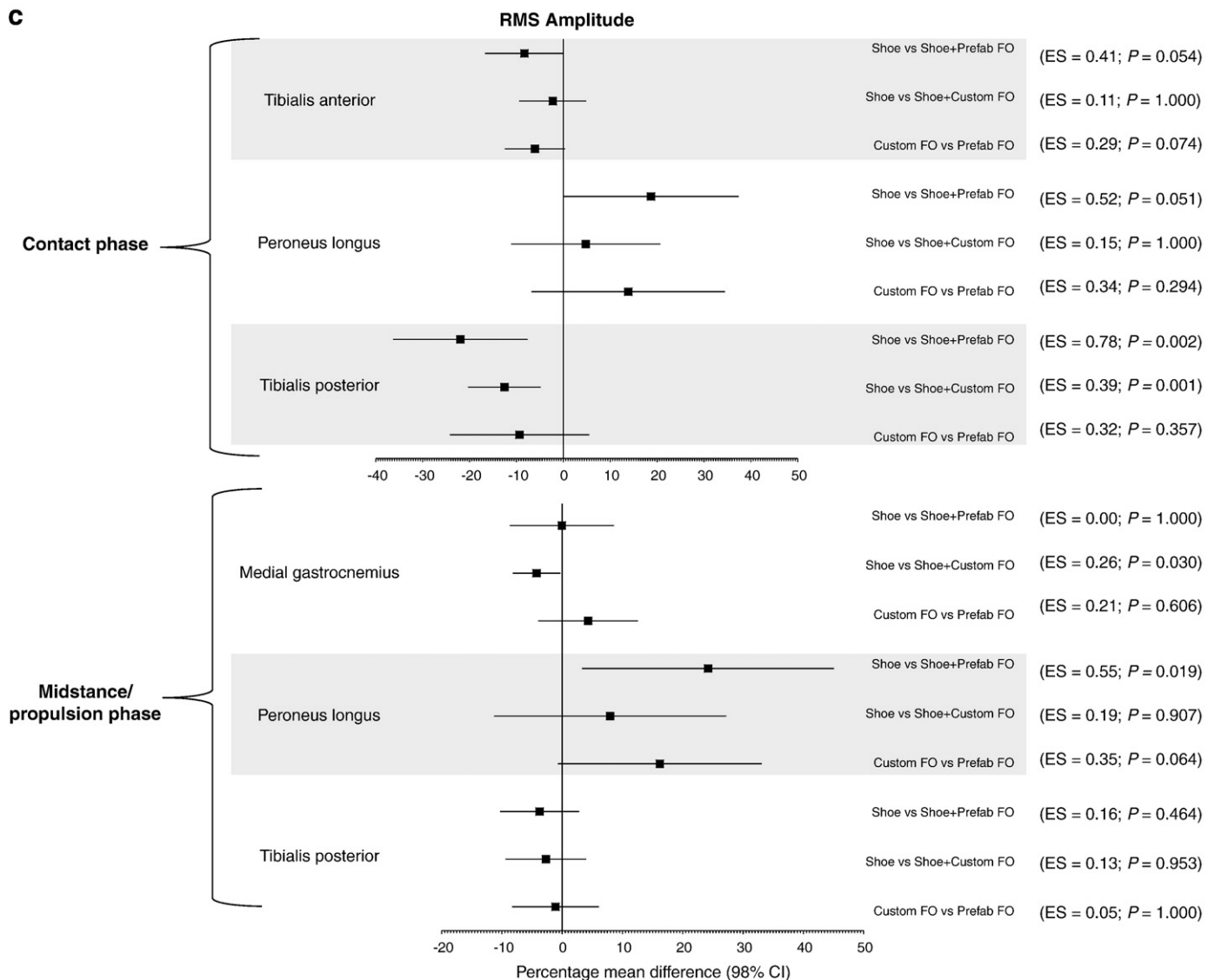


Fig. 4 (continued).

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

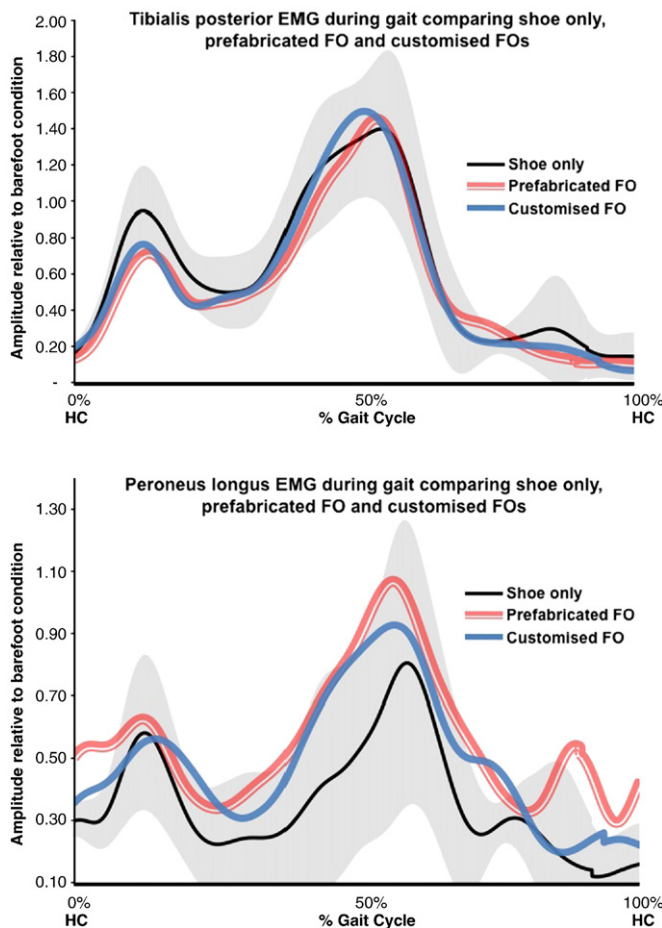


Fig. 5. EMG ensemble average for tibialis posterior and peroneus longus derived from a single gait cycle for all participants. Grey shading represents 95% confidence interval for the shoe-only condition. For clarity, error is not shown for the modified prefabricated and customised FO conditions. The 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 customised 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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