The immediate effects of foot orthosis geometry on lower limb muscle activity and foot biomechanics

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Abstract

Foot orthoses (FOs) are used to treat clinical conditions by altering the external forces applied to the foot and thereafter the forces of muscles and tendons. However, whether specific geometric design features of FOs affect muscle activation is unknown. The aim of this study was to investigate if medial heel wedging and increased medial arch height have different effects on the electromyography (EMG) amplitude of tibialis posterior, other muscles of the lower limb and the kinematics and kinetics at the rearfoot and ankle.

Healthy participants (n=19) walked in standardised shoes with i) a flat inlay; ii) a standard shape FOs, iii) standard FOs adjusted to incorporate a 6 mm increase in arch height, iv) and standard FOs adjusted to incorporate an 8° medial heel wedging and v) both the 6 mm increase in arch height and 8° increase in medial wedging. EMG was recorded from medial gastrocnemius, peroneus longus, tibialis anterior and in-dwelling tibialis posterior muscles. Motion and ground reaction force data were collected concurrently.

Tibialis posterior EMG amplitude reduced in early stance with all FOs (ηp^2 = 0.23-1.16). Tibialis posterior EMG amplitude and external ankle eversion moment significantly reduced with FOs incorporating medial wedging.

The concurrent reduction in external eversion moment and peak TP EMG amplitude in early stance with medial heel wedging demonstrates the potential for this specific FOs geometric feature to alter TP activation. Medial wedged FOs could facilitate tendon healing in tibialis posterior tendon dysfunction by reducing force going through the TP muscle tendon unit.

Introduction

Foot orthoses (FOs) alter external joint moments (Chicoine et al., 2020; Hart et al., 2020; Nester et al., 2003; Sweeney, 2016; Telfer et al., 2013b), but it is unclear how this affects muscle activation during walking. If FOs reduced the external eversion moment in early stance, less force might be required from tibialis posterior (TP) to resist eversion force based on Newton's third law. This would be reflected in reduced TP EMG signal and presumably less activation would mean less force going through the associated tendon. This could facilitate healing when treating tibialis posterior tendon dysfunction.

Systematic changes in FOs wedging have been shown to result in systematic changes in kinematic and kinetic outcomes and plantar pressure, without changes in EMG (Telfer et al., 2013a; 2013b). However, increases in FO arch height may lead to a ceiling effect, in that increasing arch height more than 3-4 mm above a flat insert may not result in proportional increases in plantar pressure in the medial midfoot (Sweeney, 2016). A systematic review found limited evidence that FOs decrease TP activity in early stance and increase peroneus longus (PL) activity in mid-late stance, but there is otherwise a lack of evidence for the effect of FOs on lower limb muscle activity during walking (Reeves et al., 2019b).

However, the review also found studies under specified the FOs designs investigated. Specific aspects of medial arch FOs geometry may have different mechanisms that exert a therapeutic effect. For example, medial heel wedges or external rearfoot posting, could decrease external eversion moments from early stance (Chicoine et al., 2020; Hart et al., 2020; Nester et al., 2003; Telfer et al., 2013b), which may accompany decreased TP EMG amplitude. In mid-late stance increased height of FOs in the medial arch may reduce TP EMG amplitude due to reduced need for support of medial arch structures.

Reduced external eversion moments in early stance with FOs would reduce the requirement of TP to generate the counter internal inversion moment. However, previous work frequently ignored kinematic and kinetic effects when analysing the effect of FOs on EMG (Reeves et al., 2019b). Any change (or lack of) in EMG data is therefore difficult to explain with respect to kinematics or kinetics. Indwelling EMG is necessary for investigating the activity of deep muscles like TP, but using fine-wire electrodes can be challenging and limits their use (O'Connor et al., 2006; Semple et al., 2009; Stacoff et al., 2007). Consequently, few studies have investigated the effects of FOs on TP EMG to a high standard (Reeves et al., 2019b). The aim of this study was to investigate if, during walking, two specific FOs geometric features would alter EMG of TP, selected other lower limb muscles and the kinematic and kinetic variables of the rearfoot and ankle**.** The FOs geometric features were (1) medial wedging and, (2) medial arch height.

It was hypothesised that compared to a flat inlay 1) medial heel wedging would reduce TP EMG peak amplitude in *early* stance (0-20% of a gait cycle) and increase inversion position at foot contact, reduce peak rearfoot eversion angle and reduce rearfoot ROM; 2) increases in FOs medial arch height would reduce TP EMG peak amplitude in *mid-late* stance (20-60% of a gait cycle) with no effect on kinematics and 3) all FOs would reduce external eversion moment, but have no effect on peak EMG amplitude of medial gastrocnemius (MG), PL or tibialis anterior (TA).

Methods

<u>*Participants*</u>

Healthy participants aged 18-60 years were recruited and screened for a neutral or pronated foot type using the Foot Posture index (FPI) (Redmond, 2005; Redmond et al., 2006). Individuals with a supinated foot type were excluded as such feet would be less likely to receive FOs clinically and might not contact the arch of the FOs. Exclusion criteria were: 1) recent lower limb injury, pain or foot/ankle deformity or pathology; 2) cardiovascular, musculoskeletal or neurological conditions, immune deficiency or haemophilia; 3) using anti-biotics, anti-coagulant/platelet therapy; 4) walking with an aid; 5) high arched/supinated foot posture on one or both feet (FPI ≤ -6). The study was approved by the ethics board of the university and all participants provided written informed consent prior to data collection.

<u>*Design features of foot orthoses*</u>

Participants walked at a self-selected speed in a gait lab in standard shoes (Lonsdale Leyton) with five inserts/FOs in a random order: four FOs and a flat inlay control (**Table 1**. Extreme increases in arch height (6 mm) and medial wedging (8°) from a standard Salfordinsole geometry were used as this was a proof of concept study. The FOs were designed and fabricated with high density Ethylene-vinyl acetate (EVA, 85 Shore A) using a computer-aided design/manufacturing system. The EVA flat inlay with no heel or arch geometry was used as the control condition.

<u>*Indwelling EMG*</u>

Single use fine-wire electrodes (50 mm long, 25 gauge, Chalgren Enterprises Inc., USA) were inserted into TP using the posterior approach (Murley et al., 2009a; Semple et al., 2009). Ultrasound imaging (Linear 60 mm probe, Echo Blaster 128 CEXT, Telemed Medical Systems, Italy) of TP with the leg flexed and everted was performed prior to insertion to ascertain insertion depth and safety window (Won et al., 2011). After insertion the participant inverted their foot several times to encourage the electrode to be embed into the muscle. Electrical stimulation (Dantec Clavis, Natus Neurology Inc., USA) was used to verify electrode placement (ankle inversion without toe flexion). The electrode tips were then attached to a spring contact sensor (bandwidth 10- 2000 Hz, Delsys, Inc., USA).

Surface EMG

Surface EMG was recorded from MG, PL and TA. Placement for PL followed a previous protocol (Reeves et al., 2019a). The guidelines for Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000) were followed for MG and TA. Standard Delsys Trigno™ sensors (99.9% sliver contact material in single differential configuration, inter-electrode distance 10 mm, 4-bar formation), were used for MG and TA and a Deslys Trigno™ Mini sensor for PL (bandwidth of 20-450 Hz , Delsys, Inc., USA).

Protocol

Height, body mass, shoe size and FPI were recorded prior to data collection. Motion data were recorded with a 15-infrared-camera Qualisys system at a sampling rate of 100 Hz (Qualisys OQUS 300, Qualisys AB, Sweden). The ground reaction forces were recorded with four synchronised force plates (BP400600, AMTI, USA) at a sampling rate of 1000 Hz. Both motion and ground reaction force data were synchronised with EMG data (Delsys, Inc., Boston, USA) sampled at 2000 Hz.

Retro-reflective markers (diameter: 10 mm) were placed bilaterally on the medial and lateral femoral epicondyles and the medial and lateral malleoli which were used to define and track the tibia. The two malleoli markers and two markers on the 1st and 5th metatarsal head (MTP1, MTP5) were used to define the foot segment, which was tracked with MTP1 & 5 and a triad cluster on the medial side of the calcaneus. All markers on the foot including the 3-marker triad cluster on the lateral side of the calcaneus for tracking the rearfoot movement were attached on the skin and exposed through apertures with 25 mm diameter in the shoes skin (Bishop et al., 2015; Majumdar et al., 2013). Data was collected on the right limb; however markers were placed on both legs and feet to enable automatic gait event detection. To change FOs, the mounting base of the triad cluster and other skin mounted markers remained on the skin while the marker or cluster was unscrewed to remove the shoe and change the FOs. The triad was locked in the identical position with a lock pin.

Participants were allowed a few minutes habituation in each condition and a self-selected walking speed was established prior to data collection using infrared timing gates (Brower Timing Systems, USA). Participants performed six walking trials per condition over a 6 m walkway in a random order. Indwelling EMG signal amplitude can attenuate after ~30 minutes of walking (Reeves et al., 2020), therefore participants were not asked to repeat any trials due to a missed force plate contact and consequently kinetic data were analysed from a minimum of three walking trials.

Analysis

Kinematic and kinetic data were computed with Visual3D (V.6, C-Motion, Inc., USA). The default segment masses in Visual3D were used based on Dempster's regression equations (Dempster, 1955). The biomechanical model was established based on the anatomical marker positions of the static trial in the flat inlay and used to normalize joint angles. To minimise the influence of walking speed, trials with stride time outside mean±5% per condition were excluded. A low-pass Butterworth filter with a 6 Hz frequency cut-off was used to filter marker trajectories. The external ankle joint moment was calculated using the Newton-Euler method of inverse dynamics, the shank co-ordinate system, positive moment directions defined using the right hand rule of co-ordinate systems and normalized to body mass. Kinematic variables are defined in **Table 2**.

For each muscle a 75 ms window was used to calculate the route mean squared (RMS) EMG, which were normalised to a gait cycle using MATLAB (R2017b). Amplitude was normalised to the peak (maximum of a gait cycle) of the mean RMS signal from the flat inlay. For each condition normalised peak EMG amplitude (subsequently referred to as peak EMG) was then averaged across gait cycles and trials.

Statistics

The measured and computed variables were exported from Visual3D to Excel (Microsoft Office Excel 2013) for the presentation of results. Statistical analysis was performed with SPSS (IBM SPSS Statistics 25). Data were checked for normality by visual inspection of skew in the histograms and are presented as means and medians when non-normal. Outliers were values beyond the first or the third quartile. One-way repeated measures ANOVA (α =0.05) were performed on discrete variables and estimated effect sizes were calculated as partial eta squared (ηp^2). Data were tested for sphericity using Mauchly's test and corrected using a Huynh–Feldt adjustment if necessary. Bonferroni post hoc analysis was applied for significant main effects. Parametric effect size (d) was calculated as the paired mean differences divided by the paired standard deviation (Cohen, 1988).

Results

Participant characteristics

Nineteen participants completed the study (7 females, age = 31 ±7 years, height = 1.71 ±0.08 m, mass= 74 ±12 kg, UK shoe size 8 ±2, FPI (average of both feet) 2 ±2, mean ± SD). FPI ranged from -3 to 6 and 2 participants were classified as having a pronated foot, the remainder had a neutral foot. Marker loss resulted in valid sample sizes reduced by two and three for kinematic and kinetic data respectively. Stride time was consistent across conditions with no statistically significant differences ($p = 0.180$, **Table 3**).

EMG

Five participants were identified as statistical outliers in the TP EMG data, among which two likely due to signal degradation, leaving n=14. There was reduced peak TP EMG in early stance with all FOs compared to the flat inlay ($p = 0.003$, $\eta p^2 = 0.26$) with variable effect sizes (0.23-1.16, **Figure 1, Table 3**). Compared to the flat inlay, peak TP EMG in early stance reduced by 16% ($p= 0.008$) for the standard arch, 5% for the high arch ($p = 1.000$), 19% ($p = 0.031$) for the medial wedge and 20% ($p = 0.040$) for the arch & wedge. There was no significant effect of FOs on mid-late stance TP data ($p = 0.113$, $\eta p^2 = 0.132$).

There was no effect of FOs on peak EMG of TA ($p = 0.157$, $\eta p^2 = 0.100$) or MG ($p = 0.327$, $\eta p^2 = 0.084$). There was a main effect of FOs on peak PL EMG ($p = 0.01$, $\eta p^2 = 0.193$). There were no significant effects after adjusting for multiple comparisons ($p>0.05$), however there were small to moderate increases with FOs ($d= 0.37$-74, **Table 3**).

Kinematics

The rearfoot was less everted with FOs compared with the flat inlay (**Figure 2**). However, there was no significant effect of FOs on discrete variables MaxEv ($p = 0.133$, $\eta p^2 = 0.124$) or MaxES ($p = 0.556$, $\eta p^2 = 0.043$, **Table 4)**. There was a significant main effect of FOs on ROM ($p = 0.011$, $\eta p^2 = 0.233$). The ROM of the medial wedge (7.5° ±2.7°) was significantly reduced ($p = 0.051$) in comparison with the flat inlay (9.6° ±3.4°).

Kinetics

The external ankle inversion/eversion moment increased (inversion direction) with the four FOs (**Figure 2**) versus the flat inlay. There was a significant effect of condition ($p < 0.001$, $\eta p^2 = 0.530$) on MaxMEv (**Table 4**). Decreased MaxMEv was significant for the medial wedge ($p = 0.001$, -30%) and arch & wedge ($p < 0.001$, -38%) versus the flat inlay. There was also a significant effect of condition ($p < 0.001$, $p^2 = 0.540$) on MaxMInv (**Table 4**). Increased MaxMInv was significant for the standard arch ($p = 0.035$, +7%), medial wedge ($p = 0.001$, +15%) and arch & wedge ($p < 0.001$, +19%) and not significant with the high arch ($p = 0.073$, +8%) versus the flat inlay.

Discussion

This study investigated whether medial wedging and increased medial arch height have effects on muscle activity of the lower limb and rearfoot and ankle biomechanics during walking. Peak TP EMG decreased in early stance with the standard FOs and medial wedging, which was partly accompanied by decreased external eversion moment. There was no significant change in the EMG of the other lower limb muscles tested.

EMG

TP EMG peak decreased in early stance with the standard arch and medial heel wedging, but there was no significant effect of increasing arch height. The reduction in peak TP EMG (16-20%) was of a similar magnitude to previously reported with custom and pre-fabricated FOs relative to shoes (12-19%) (Murley et al., 2010). The difference in the two peaks between the previous and present study is likely because Murley et al. (2010) recruited flat footed individuals based on clinical and radiographical measures and the current study included participants with neutral and pronated feet according to the FPI. The heterogeneity of FPI in our sample may partly explain the high variability of TP EMG and the lack of significant effect of increased arch height on TP activity in mid-late stance. Based on the present results, the potential for FOs to reduce TP EMG in early stance does not appear to be specific to pes planus.

Other studies have reported reduced TP activity with FOs versus barefoot, but not footwear alone in both walking and running (Akuzawa et al., 2016; Akuzawa et al., 2021; Maharaj et al., 2018). In one of these studies it was suggested that the effect of the FOs geometry was too subtle or the stiffer FOs material (semi rigid 4-mm polypropylene) compared to the EVA shoe liner may have counteracted any potential effect of the FOs (Maharaj et al., 2018). It is also possible that measurement error due to a change in the recording capacity of the fine-wire electrode was larger than any small effect of the FOs, as indwelling EMG amplitude can reduce over time (Reeves et al., 2020). The order of experimental conditions was either not randomised, or not stated, and reported as barefoot, footwear alone and footwear plus FOs (Akuzawa et al., 2016; Akuzawa et al., 2021; Maharaj et al., 2018). Without knowledge of the within session reliability of the EMG recordings nor the duration of sessions, the results of these studies need to be interpreted with caution, as an order effect cannot be ruled out. It remains unknown whether EMG is sufficiently sensitive to identify possibly subtle effects of FOs on muscle recruitment.

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

As hypothesised, there was no significant change in MG or TA with FOs, which has been reported previously (Barn et al., 2013; Chicoine et al., 2020; Mills et al., 2012; Murley and Bird, 2006; Murley et al., 2010; Telfer et al., 2013a). In one study FOs increased PL activity (Murley et al., 2010) and although not significant, a small to moderate effect was found in the present study for increased PL activity with FOs. If FOs reduced TP EMG this could be accompanied by increased EMG of its antagonist PL. However muscle activity from TP and PL do not necessarily represent equal opposing inversion and eversion moments respectively, due to additional muscle tendon parameters like physiological cross-sectional area and fibre length and different moment arms (Lieber and Friden, 2000; Murley et al., 2009a; Ward et al., 2009).

Kinetics

Our hypotheses on the effect of FO geometry on kinetics can be partially accepted as the external eversion moment reduced across stance (**Figure 2**) for all FOs, however only medial wedging (medial wedge and arch & wedge) decreased the maximum external eversion moment and the wedging and standard arch significantly increased the maximum external inversion moment. There was no effect of the high arch on the discrete moment variables. The reduction in TP EMG amplitude with the wedge and arch & wedge FOs was less (19% and 20% respectively) than the reduction in MaxMEv (reduced maximum eversion of -30% and -38% respectively). A different magnitude of change between EMG and joint moment with medial heel wedging is unsurprising given the non-linear relationship between force and EMG for dynamic contractions. Secondly, the axes of rotation around which the TP acts is not the same as where the external ankle inversion/eversion moment was calculated. Thirdly, as well as TP, the triceps surae, TA and flexor hallucis longus can all contribute to the generation of an inversion moment (Klein et al., 1996). Finally, FOs could have influenced the length of the TP muscle fascicles or tendon, and the energy storage of TP tendon (Maharaj et al., 2016; Maharaj et al., 2017b), which would affect the joint moment, but not necessarily be reflected in the EMG. ~~Energy storage and release in late stance by the TP tendon may have contributed to the greater inversion moment in late stance in the wedge conditions, despite the rearfoot, where the wedge acts, not being in contact with the ground at this stage. Sweeney (2016) also found medially wedged FOs shifted the foot more into inversion with increased maximum internal inversion/eversion ankle moment in mid-late stance. Nevertheless, the present study demonstrated that specific changes in rearfoot posting can change both joint moment and TP muscle activity.~~

Kinematics

Although there was a shift into a more inverted foot position with medial wedging, the changes in discrete kinematic variables were not statistically significant, despite >2° change in peak rearfoot eversion angle. As FOs are designed for a pronated foot and most participants in the study had a neutral foot, this could have limited the effect of the FOs. However, the lack of significant change in kinematics due to medial wedging reflects the large variability in the response to FOs which has been observed previously (Donoghue et al., 2008; Hart et al., 2020; Mills et al., 2009). The present results also support the belief that the therapeutic effect of FOs is related to changes in kinetics rather than kinematics, as it is not known whether a typically small effect of ~2° is clinically meaningful.

Limitations

Possible reduction in indwelling EMG amplitude over time, independent of the FOs effects needs to be considered. The maximum time from which EMG can be accurately recorded with fine-wire electrodes without a drop in amplitude is unknown, but prior work has estimated this to be approximately 20-30 minutes (Reeves et al., 2020). To mitigate this the conditions were randomised, so any signal degradation would likely have been washed out by participants wearing the FOs in different orders and outliers were excluded.

Peak activation of TP has previously been shown to occur in early stance in individuals with a normal arch height and mid-late stance in those with a flat arch (Murley et al., 2009b). Consequently, in our heterogeneous sample the location of peak TP amplitude from the flat inlay could occur at different phases of a gait cycle and so the group mean of the normalised signal from the flat inlay could be <100%. However we chose to normalise to the peak rather than a maximum voluntary contraction (MVC) because an expert consensus suggested that performing a MVC for the purposes of normalisation could alter the position and/or orientation of the recording tips of fine-wire electrodes within the muscle and could damage the wire (Besomi et al., 2019). Additionally, normalising fine-wire EMG signals from shank muscles to the peak has been shown to have greater between-subject and between-session repeatability than normalising to MVCs ~~previous work found peak normalisation superior to normalising to in shank muscles for reducing variability~~ (Onmanee, 2016).

Conclusion

The concurrent reduction in external eversion moment and peak TP EMG amplitude in early stance with medial wedging demonstrates the potential for specific FOs geometry to alter TP biomechanics. If the intention of orthotic treatment was to reduce force through the TP muscle tendon unit then FOs with a medial wedge could be effective.

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References

Akuzawa, H., Imai, A., Iizuka, S., Matsunaga, N., Kaneoka, K., 2016. Calf muscle activity alteration with foot orthoses insertion during walking measured by fine-wire electromyography. Journal of physical therapy science 28, 3458-3462.

Akuzawa, H., Imai, A., Iizuka, S., Matsunaga, N., Kaneoka, K., 2021. Tibialis posterior muscle activity alteration with foot orthosis insertion measured by fine-wire electromyography. Footwear Sci, 1-9.

Barn, R., Turner, D.E., Rafferty, D., Sturrock, R.D., Woodburn, J., 2013. Tibialis Posterior Tenosynovitis and Associated Pes Plano Valgus in Rheumatoid Arthritis: Electromyography, Multisegment Foot Kinematics, and Ultrasound Features. Arthritis Care Res. (Hoboken) 65, 495-502.

Besomi, M., Hodges, P.W., Van Dieën, J., Carson, R.G., Clancy, E.A., Disselhorst-Klug, C., Holobar, A., Hug, F., Kiernan, M.C., Lowery, M., 2019. Consensus for experimental design in electromyography (CEDE) project: Electrode selection matrix. J. Electromyogr. Kinesiol. 48, 128-144.

Bishop, C., Arnold, J.B., Fraysse, F., Thewlis, D., 2015. A method to investigate the effect of shoe-hole size on surface marker movement when describing in-shoe joint kinematics using a multi-segment foot model. Gait Posture 41, 295-299.

Chicoine, D., Bouchard, M., Laurendeau, S., Moisan, G., Belzile, E.L., Corbeil, P., 2020. Biomechanical effects of three types of foot orthoses in individuals with posterior tibial tendon dysfunction. Gait Posture 83, 237-244.

Cohen, J., 1988. Statistical Power Analysis for the Behavioral Sciences, (L. Erlbaum Associates, Hillsdale, NJ). Erlbaum Associates: Hillsdale, NJ, USA.

Dempster, W.T., 1955. Space requirements of the seated operator, geometrical, kinematic, and mechanical aspects of the body with special reference to the limbs. Michigan State Univ East Lansing.

Donoghue, O.A., Harrison, A.J., Laxton, P., Jones, R.K., 2008. Orthotic control of rear foot and lower limb motion during running in participants with chronic Achilles tendon injury. Sports Biomechanics 7, 194-205.

Hart, H.F., Crossley, K.M., Bonacci, J., Ackland, D.C., Pandy, M.G., Collins, N.J., 2020. Immediate effects of foot orthoses on gait biomechanics in individuals with persistent patellofemoral pain. Gait Posture 77, 20-28.

Hermens, H.J., Freriks, B., Disselhorst-Klug, C., Rau, G., 2000. Development of recommendations for SEMG sensors and sensor placement procedures. J. Electromyogr. Kinesiol. 10, 361-374.

Klein, P., Mattys, S., Rooze, M., 1996. Moment arm length variations of selected muscles acting on talocrural and subtalar joints during movement: An in vitro study. J. Biomech. 29, 21-30.

Lieber, R.L., Friden, J., 2000. Functional and clinical significance of skeletal muscle architecture. Muscle Nerve 23, 1647-1666.

Maharaj, J.N., Cresswell, A.G., Lichtwark, G.A., 2016. The mechanical function of the tibialis posterior muscle and its tendon during locomotion. Journal of Biomechanics 49, 3238-3243.

Maharaj, J.N., Cresswell, A.G., Lichtwark, G.A., 2017a. Foot structure is significantly associated to subtalar joint kinetics and mechanical energetics. Gait Posture 58, 159-165.

Maharaj, J.N., Cresswell, A.G., Lichtwark, G.A., 2017b. Subtalar Joint Pronation and Energy Absorption Requirements During Walking are Related to Tibialis Posterior Tendinous Tissue Strain. Sci. Rep. 7.

Maharaj, J.N., Cresswell, A.G., Lichtwark, G.A., 2018. The Immediate Effect of Foot Orthoses on Subtalar Joint Mechanics and Energetics. Med. Sci. Sports Exerc. 50, 1449-1456.

Majumdar, R., Laxton, P., Thuesen, A., Richards, B., Liu, A., Aran-Ais, F., Montiel Parreno, E., Nester, C.J., 2013. Development and evaluation of prefabricated antipronation foot orthosis. J. Rehabil. Res. Dev. 50, 1331-1341.

Mills, K., Blanch, P., Chapman, A.R., McPoil, T.G., Vicenzino, B., 2009. Foot orthoses and gait: a systematic review and meta-analysis of literature pertaining to potential mechanisms. Br. J. Sports Med. 44, 1035-1046.

Mills, K., Blanch, P., Vicenzino, B., 2012. Comfort and midfoot mobility rather than orthosis hardness or contouring influence their immediate effects on lower limb function in patients with anterior knee pain. Clin Biomech 27, 202-208.

Murley, G.S., Bird, A.R., 2006. The effect of three levels of foot orthotic wedging on the surface electromyographic activity of selected lower limb muscles during gait. Clin Biomech 21, 1074-1080.

Murley, G.S., Buldt, A.K., Trump, P.J., Wickham, J.B., 2009a. Tibialis posterior EMG activity during barefoot walking in people with neutral foot posture. J. Electromyogr. Kinesiol. 19, E69-E77.

Murley, G.S., Landorf, K.B., Menz, H.B., 2010. Do foot orthoses change lower limb muscle activity in flat-arched feet towards a pattern observed in normal-arched feet? Clin Biomech 25, 728-736.

Murley, G.S., Menz, H.B., Landorf, K.B., 2009b. Foot posture influences the electromyographic activity of selected lower limb muscles during gait. J Foot Ankle Res 2, 35-35.

Nester, C.J., van der Linden, M.L., Bowker, P., 2003. Effect of foot orthoses on the kinematics and kinetics of normal walking gait. Gait Posture 17, 180-187.

O'Connor, K.M., Price, T.B., Hamill, J., 2006. Examination of extrinsic foot muscles during running using mfMRI and EMG. J. Electromyogr. Kinesiol. 16, 522-530.

Onmanee, P., 2016. DEVELOPMENT OF EMG MEASUREMENT USING FINE-WIRE AND SURFACE SENSORS IN LOWER LIMB MUSCLES FOR GAIT ANALYSIS. PhD Thesis, University of Salford.

Redmond, A., 2005. The Foot Posture Index: User guide and manual, p. 19.

Redmond, A.C., Crosbie, J., Ouvrier, R.A., 2006. Development and validation of a novel rating system for scoring standing foot posture: the Foot Posture Index. Clin Biomech 21, 89-98.

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

Reeves, J., Jones, R., Liu, A., Bent, L., Plater, E., Nester, C., 2019b. A systematic review of the effect of footwear, foot orthoses and taping on lower limb muscle activity during walking and running. Prosthet. Orthot. Int. 0, 0309364619870666.

Reeves, J., Starbuck, C., Nester, C., 2020. EMG gait data from indwelling electrodes is attenuated over time and changes independent of any experimental effect. J. Electromyogr. Kinesiol. 54, 102461.

Semple, R., Murley, G.S., Woodburn, J., Turner, D.E., 2009. Tibialis posterior in health and disease: a review of structure and function with specific reference to electromyographic studies. J Foot Ankle Res 2.

Stacoff, A., Quervain, I.K.-d., Dettwyler, M., Wolf, P., List, R., Ukelo, T., Stüssi, E., 2007. Biomechanical effects of foot orthoses during walking. The Foot 17, 143-153.

Sweeney, D., 2016. Investigation into the variable biomechanical responses to antipronation foot orthoses. PhD Thesis, University of Salford.

Telfer, S., Abbott, M., Steultjens, M., Rafferty, D., Woodburn, J., 2013a. Dose–response effects of customised foot orthoses on lower limb muscle activity and plantar pressures in pronated foot type. Gait Posture 38, 443-449.

Telfer, S., Abbott, M., Steultjens, M.P.M., Woodburn, J., 2013b. Dose response effects of customised foot orthoses on lower limb kinematics and kinetics in pronated foot type. J. Biomech. 46, 1489-1495.

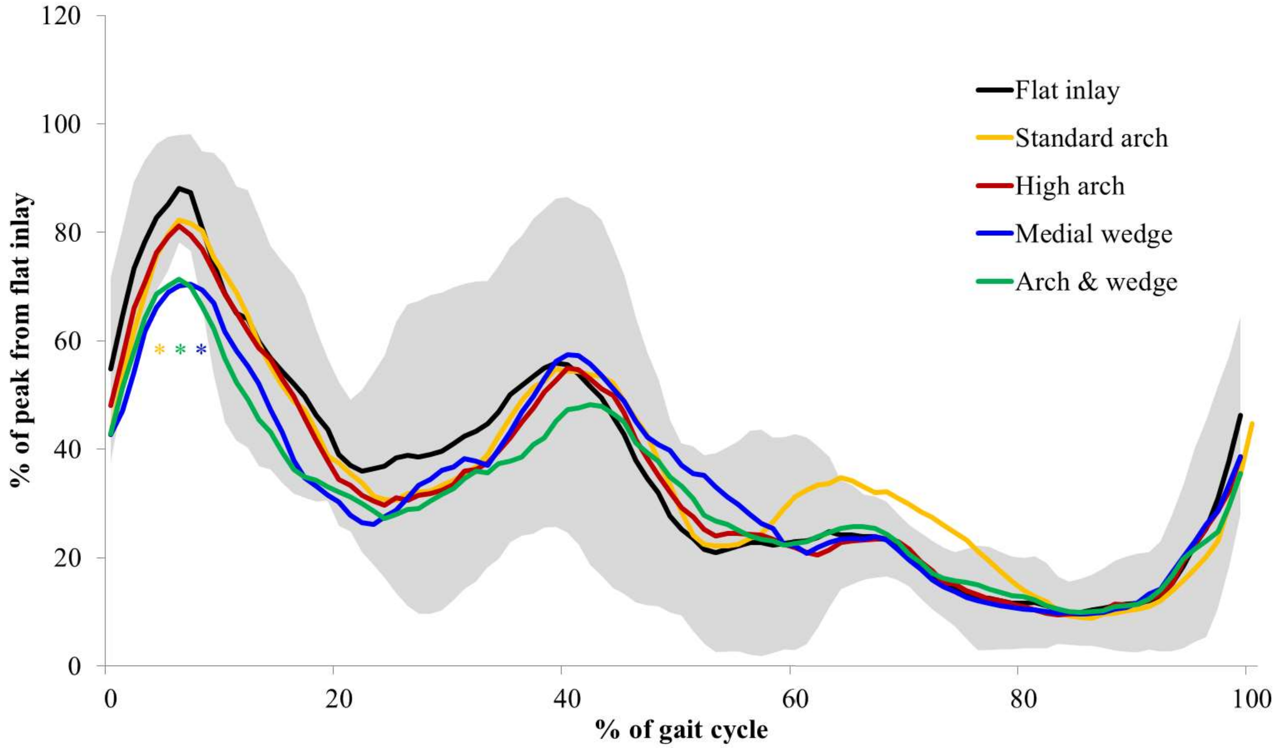
Ward, S.R., Eng, C.M., Smallwood, L.H., Lieber, R.L., 2009. Are current measurements of lower extremity muscle architecture accurate? Clin. Orthop. Relat. Res. 467, 1074-1082.

Won, S.J., Kim, J.Y., Yoon, J.S., Kim, S.J., 2011. Ultrasonographic Evaluation of Needle Electromyography Insertion Into the Tibialis Posterior Using a Posterior Approach. Arch. Phys. Med. Rehabil. 92, 1921-1923.

Figure 1. Tibialis posterior EMG (n=14) over the gait cycle. Black lines: flat inlay; yellow lines: standard arch; red lines: high arch; blue lines: medial wedge and green lines: arch & wedge. Yellow, blue and green * indicate the condition achieved statistically significant effect ($p<0.05$). The grey shaded area represents the standard deviation of the flat inlay.

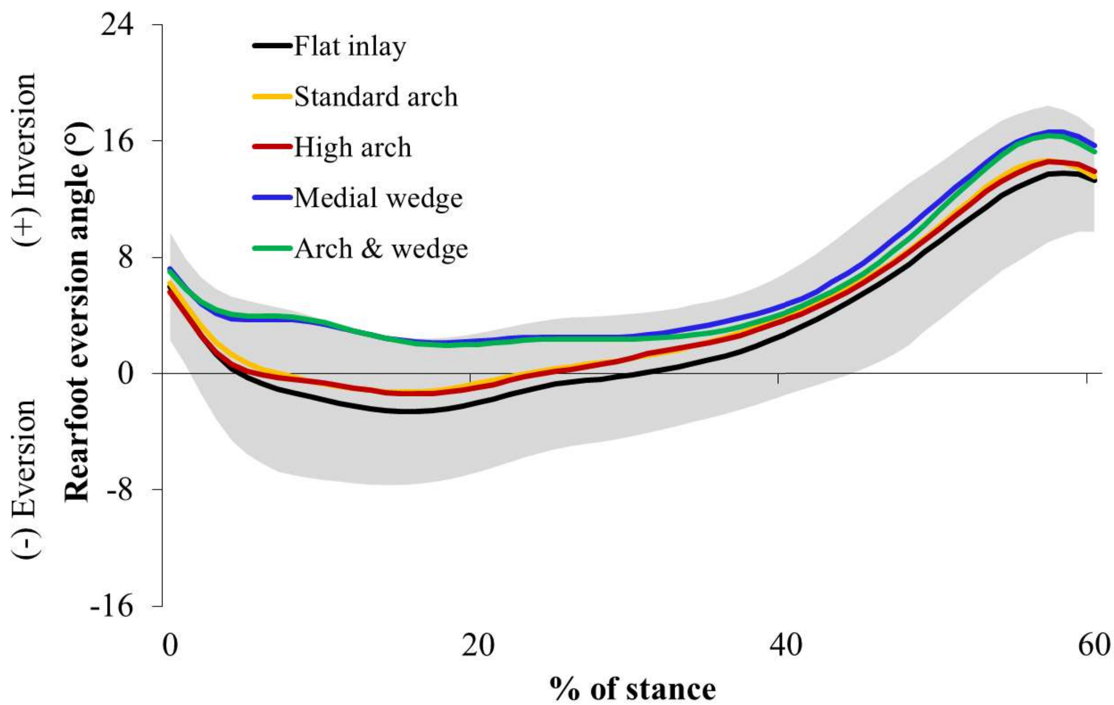
Figure 2. Rearfoot eversion angle (n=16) across stance. Black lines: flat inlay; yellow lines: Salfordinsole; red lines: high arch; blue lines: wedge and green lines: arch & wedge. The grey shaded area represents the standard deviation of the flat inlay.

Figure 3. External ankle inversion/eversion moment (n=17). Black lines: flat inlay; yellow lines: Salfordinsole; red lines: high arch; blue lines: medial wedge and green lines: arch & wedge. Yellow, blue and green * indicate the condition achieved statistically significant effect ($p<0.05$). The grey shaded area represents the standard deviation of the flat inlay.



120
100
80
60
40
20
0
% of peak from flat inlay
0
20
40
60
80
100
% of gait cycle
Flat inlay
Standard arch
High arch
Medial wedge
Arch & wedge
* * *

Figure


24
16
8
0
-8
-16
Flat inlay
Standard arch
High arch
Medial wedge
Arch & wedge
(+) Inversion
(-) Eversion
Rearfoot eversion angle (°)
0
20
40
60
% of stance

Figure

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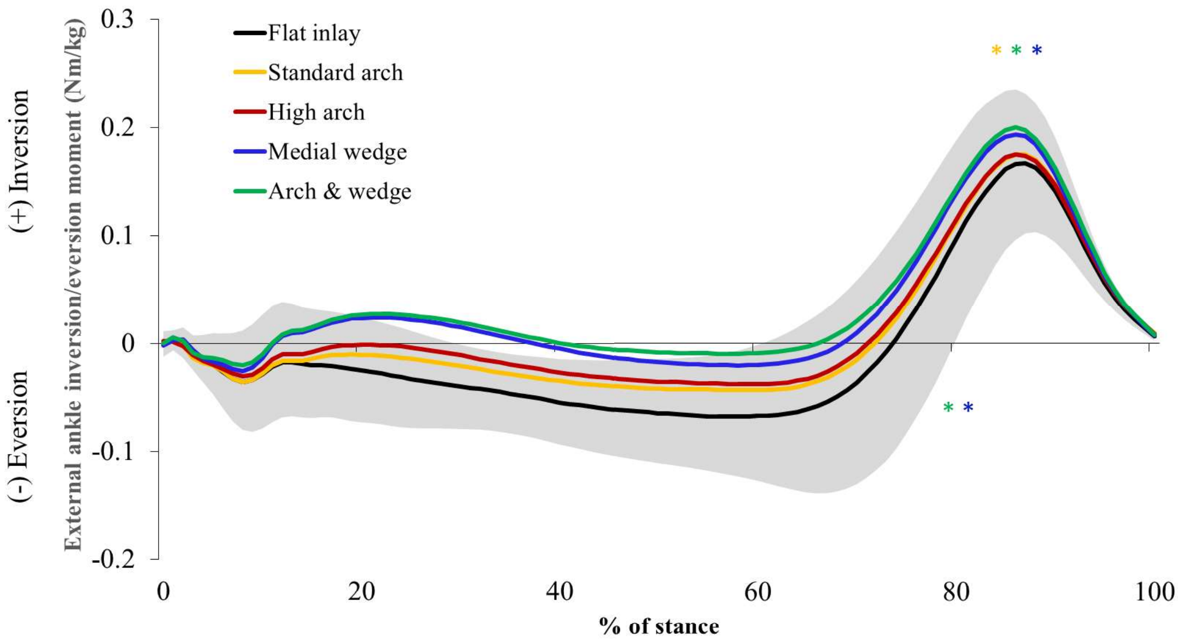
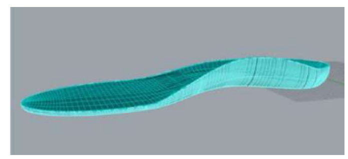
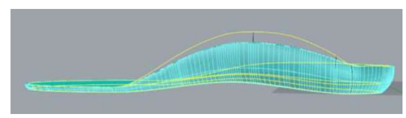
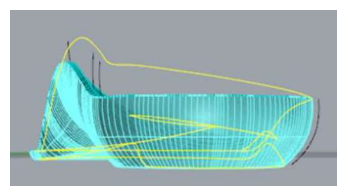
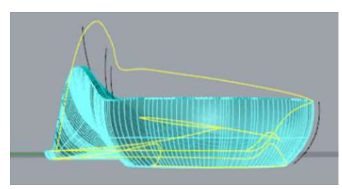


Table 1. Experimental conditions. The yellow lines represent the border of the foot orthosis above the standard Salfordinsole

Condition	Description	Image
Flat inlay	3 mm insole made from EVA, which was the same material as the FOs conditions	
Salfordinsole (standard arch)	The standard Salfordinsole (20 mm arch height)	
High arch	Salfordinsole with a 6 mm increase in arch height (26 mm arch height in total)	
Medial wedge	Salfordinsole with an additional 8° medial heel wedging (standard 20 mm arch height)	
Arch & wedge	Salfordinsole with both a 6 mm increase in arch height (26 mm arch height in total) and 8° medial heel wedging)	

EVA= Ethylene-vinyl acetate, FOs= foot orthoses

Table 2. Definition of the discrete kinematic and kinetic variables

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

Table 3. Mean ± SD right stride time and peak EMG amplitude expressed as a percentage of the peak of the flat inlay

	Flat inlay	Standard arch	High arch	Medial wedge	Arch & wedge
Right stride time (s)	1.08 ± 0.06	1.08 ± 0.06	1.08 ± 0.06	1.07 ± 0.07	1.08 ± 0.07
EMG					
Peak TP early stance (%)	95 ±8	79 ±17*	90 ±25	77 ±18*	75 ±25*
Effect size (d) vs. flat inlay		1.16	0.23	0.97	0.93
Peak TP mid-late stance (%)	81 ±23	76 ±26	76 ±32	85 ±31	72 ±27
Effect size (d) vs. flat inlay		0.30	0.22	0.20	0.43
Peak MG EMG (%)	100 ± 0	98 ± 9	101 ± 7	100 ± 5	96 ± 10
Effect size (d) vs. flat inlay		0.20	0.10	0.04	0.36
Peak PL EMG (%)	100 ± 0	104 ± 11	104 ± 10	110 ± 14	109 ± 12
Effect size (d) vs. flat inlay		0.37	0.37	0.70	0.74
Peak TA EMG (%)	100 ± 0	100 ± 10	100 ± 9	97 ± 8	94 ± 12
Effect size (d) vs. flat inlay		0.04	0.05	0.36	0.48

* $p<0.05$ with respect to flat inlay

Table 4. Mean ± SD discrete kinematic (n = 16) and kinetic variables (n=17)

	Flat inlay	Standard arch	High arch	Medial wedge	Arch & wedge
Kinematics					
MaxEv (°)	-3.08 ± 5.12	-1.40 ± 7.37	-2.61 ± 4.74	0.36 ± 7.22	-0.61 ± 4.25
Effect size (d) vs. flat inlay		0.26	0.14	0.59	0.49
ROM (°)	9.56 ± 3.38	8.96 ± 2.96	8.64 ± 3.42	7.51 ± 2.74*	7.81 ± 2.85
Effect size (d) vs. flat inlay		0.22	0.55	0.79	0.55
MaxES (°)	5.64 ± 3.35	6.21 ± 5.17	5.52 ± 3.75	6.78 ± 5.07	6.78 ± 2.31
Effect size (d) vs. flat inlay		0.14	0.04	0.28	0.34
Kinetics					
MaxMEv (Nm/kg)	-0.11 ± 0.04	-0.09 ± 0.04	-0.09 ± 0.05	-0.08* ± 0.04	-0.07* ± 0.04
Effect size (d) vs. flat inlay		0.51	0.65	1.29	1.52
MaxMInv (Nm/kg)	0.18 ± 0.07	0.19* ± 0.07	0.19 ± 0.08	0.21* ± 0.08	0.22* ± 0.08
Effect size (d) vs. flat inlay		0.83	0.74	1.24	1.75

* $p<0.05$ with respect to flat inlay

Declaration of conflicting of interests

C.N. owns equity in Salfordinsole Healthcare Ltd. (Nuneaton, UK) that manufactures foot orthoses. Other Authors have no conflicts of interest to declare.

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