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# Effect of antipronation foot orthosis geometry on compression of heel and arch soft tissues

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**Abstract—**This study aimed to understand how systematic changes in arch height and two designs of heel wedging affect soft tissues under the foot. Soft tissue thickness under the heel and navicular was measured using ultrasound. Heel pad thickness was measured while subjects were standing on a flat surface and also while they were standing on an orthosis with 4 and 8 degree extrinsic wedges and 4 and 8 mm intrinsic wedges ($n$ = 27). Arch soft tissue thickness was measured when subjects were standing and when standing on an orthosis with –6 mm, standard, and +6 mm increments in arch height ($n$ = 25). Extrinsic and intrinsic heel wedges significantly increased soft tissue thickness under the heel compared with no orthosis. The 4 and 8 degree extrinsic wedges increased tissue thickness by 28.3% and 27.6%, respectively, while the 4 and 8 mm intrinsic wedges increased thickness by 23.0% and 14.6%, respectively. Orthotic arch height significantly affected arch soft tissue thickness. Compared with the no orthosis condition, the –6 mm, standard, and +6 mm arch heights decreased arch tissue thickness by 9.1%, 10.2%, and 11.8%, respectively. This study demonstrates that change in orthotic geometry creates different plantar soft tissue responses that we expect to affect transmission of force to underlying foot bones.

**Key words:** antipronation, arch profile, extrinsic wedge, foot orthosis, heel pad, intrinsic wedge, plantar foot, pronation, tissue compression, ultrasound.

## INTRODUCTION

Antipronation foot orthoses (APFOs) are commonly used by healthcare professionals to treat a variety of lower-limb pathologies that are thought to be caused by excessive pronation [1–2]. APFOs are purported to function by applying an inversion moment at the rearfoot, reducing calcaneal eversion, and reducing dorsiflexion of the joints forming the medial longitudinal arch of the foot [3]. To achieve this, the orthosis must first alter the load being applied through the sole of the foot.

The principal design features in an APFO are the geometry of the heel and arch sections. In addition to material stiffness [4], these features will alter loads between the plantar aspect of the foot and the orthotic surface [5–7]. Increases in peak pressure in the arch [8–9] and reductions in pressure in the heel [8–11] have been well documented for total contact orthoses used in patients with diabetes. Similarly, both extrinsic [12–13]

**Abbreviations:** APFO = antipronation foot orthosis, CAD/CAM = computer-aided design/computer-aided manufacturing, MRI = magnetic resonance imaging, SD = standard deviation.

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and intrinsic [14] heel wedges have been shown to increase pressure values in the medial heel. However, it is less clear how changes in load at the skin surface affect loads transferred to bone. This will be influenced by mechanical properties of soft tissues residing between the foot-orthosis interface.

The effect of foot orthoses on plantar tissue structures has been quantified previously. Magnetic resonance imaging (MRI) modeling has been used to examine how cushioning materials of different densities and contours affect tissues under the calcaneus [15]. Similarly, lateral radiographs have been used to show that a heel cup that constrains soft tissue displacement increases plantar heel pad thickness compared with use of no heel cup [16]. However, the orthoses used in previous studies did not incorporate a medial wedge. This design feature has been associated with an antipronation effect [12–14], and there are also two different designs (inside and outside the heel cup) with different proposed effects [17]. Thus, it is unclear how an antipronation orthosis will affect plantar soft tissue characteristics under either the calcaneus or medial arch.

Ultrasound is becoming increasingly popular for quantifying soft tissue characteristics [18–19]. As well as being noninvasive, it is portable so, unlike MRI [20], can be used to quantify tissue characteristics in a weight-bearing prone position. Ultrasound has been used to measure foot muscles [21] and skin and plantar aponeurosis [22]. Furthermore, it demonstrates good intra- and interobserver reliability with foot structures [19]. However, to date, only one study has used ultrasound to study the effect of orthotic designs focusing on the heel, and no APFO was included [23].

The aim of this article was to use ultrasound to characterize static bare foot plantar tissue responses to different APFO geometries. Specifically, we examined the effect of incrementally increasing medial heel wedge and arch height on plantar soft tissues. We hypothesized that incrementally increasing wedge and arch height would compress soft tissues in a systematic manner.

## METHODS

### Participants

Twenty-seven participants (14 male/13 female; mean ± standard deviation [SD] age 29.9 ± 6.7 yr, weight 70.7 ± 9.3 kg, and height 1.71 ± 0.08 m) volunteered. Data were collected from the right foot. Participants reported no recent history of lower-limb pathology or surgery and had a neutral foot alignment as defined by the Foot Posture Index [24]. All gave informed written consent to participate.

### Orthoses

The Salfordinsole (Salfordinsole Healthcare Ltd; United Kingdom) was chosen as an example APFO [25], but like most orthotic products it is impenetrable to ultrasound signals. To study its effect on foot tissues, we made an exact copy of the APFO in a rigid plastic sonographic material (Northplex, North Sea Plastics, Ltd; Glasgow, United Kingdom). To create these copies, we created positive plaster of paris molds of the orthotic from milled ethylene vinyl acetate versions of the Salfordinsole based on computer-aided design/computer-aided manufacturing (CAD/CAM) designs. Northplex sheets 3 mm thick were subsequently heat molded and vacuum formed over the Salfordinsole positive models. Northplex allows ultrasound signals to pass through its structure and is almost incompressible in sheet form. It remains very rigid when molded into an APFO shape, being similar to a polypropylene-style foot orthotic. We chose to investigate the effect of varying the size of the medial wedge using two different approaches, both used in practice: an intrinsic wedge (inside the heel cup) measured in millimeters and an extrinsic wedge (under the heel cup) measured in degrees. Our rationale for this choice was that the extrinsic wedge only alters the geometry underneath the orthotic (i.e., the surface in contact with the shoe) but tilts the upper surface and heel cup that is in contact with the heel laterally. In contrast, the intrinsic wedge alters the internal geometry of the heel cup that directly contacts the heel skin [17]. Two Northplex designs were produced for each approach: a 4° and 8° extrinsic wedge and a 4 mm and 8 mm intrinsic wedge.

Three additional Northplex insoles were produced to investigate the effect of varying arch height. The first of these had the standard Salfordinsole arch height. The other two had arch heights that were 6 mm less and 6 mm greater than the standard (–6 mm and +6 mm). All orthotic designs were created and modified using CAD/CAM to strictly control changes in orthotic geometry (iCUSTOM software, Salfordinsole Healthcare Ltd). **Figure 1** shows the design of the different heel and arch geometries.

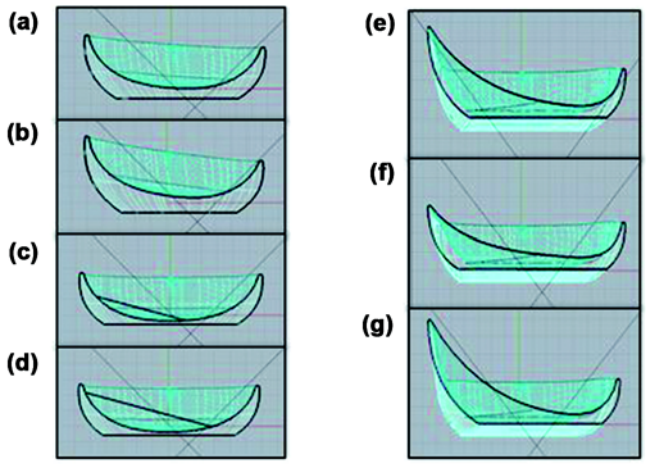


**Figure 1.**
**(a)–(b)** 4° and 8° extrinsic medial wedges. **(c)–(d)** 4 and 8 mm intrinsic medial wedges. **(e)** Standard arch profile. **(f)** –6 mm arch height. **(g)** +6 mm arch height.

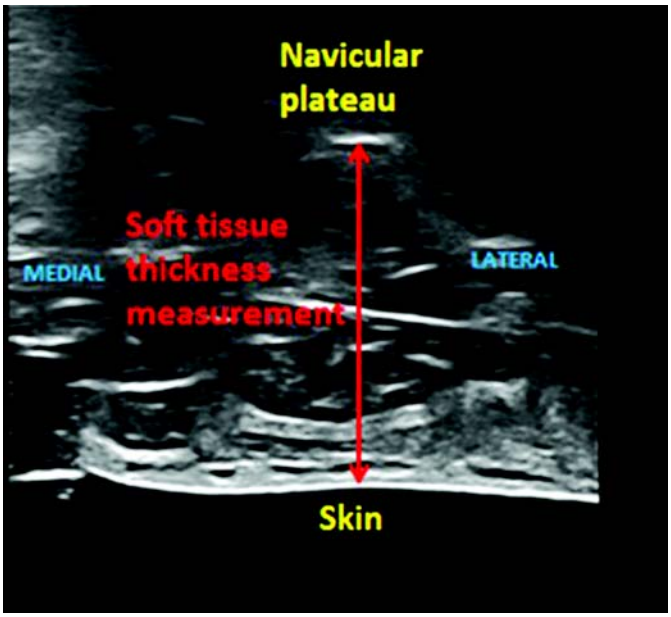


**Figure 2.**
Ultrasound image showing landmarks used to record tissue thickness measurement for baseline arch condition.

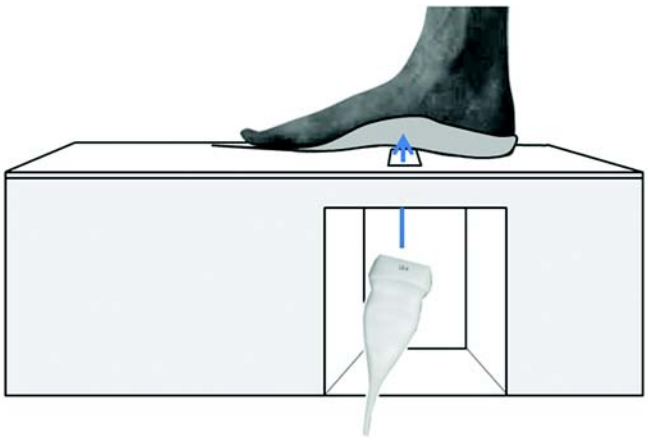

**Figure 3.**
Scanning platform used to enable ultrasound imaging of plantar soft tissues under heel and arch through Northplex insoles.

## Ultrasound and Scanning Platform

A MyLab 70 Xvision ultrasound machine and 13 MHz linear array transducer (Esoate Europe; United Kingdom) were used to image plantar soft tissues above the orthotic. Measures of soft tissue thickness were obtained in the arch (3 arch heights) and heel area (4 heel wedges). For the arch, the navicular was assumed to represent the peak in the medial arch height and correspond to peak orthotic arch height. A plateau on the plantar surface of the navicular was used as an internal bony reference for measures of arch tissue thickness (**Figure 2**). This landmark was imaged in the frontal plane and lateral to the navicular tuberosity by 1/3 of the navicular width. Pilot work ($n$ = 10) showed high intrarater reliability (intraclass correlation coefficient 0.980, 95% confidence interval = 0.922–0.995) of this measure while subjects were standing on the standard APFO. For the heel area, the calcaneal tuberosity was selected as the reference anatomical landmark, viewed in the frontal plane. Due to its superficial location, shape, and tissue properties, it is easily identified and has demonstrated high reliability [19].

A platform incorporating a 50 × 120 mm opening (**Figure 3**) was used to position the ultrasound transducer under the orthotic/foot at the heel and arch sites. Baseline measurements of arch and heel soft tissue thickness (i.e., with no orthotic) were obtained when subjects were standing on the platform. For the heel baseline measurement, tissue was imaged through a flat sheet of Northplex.

Each participant stood with their right foot on the orthotic, which was secured over the platform aperture. Participants stood on one leg and used handrails to prevent sway. To improve the extent to which this static assessment might replicate soft tissue compression in walking, each subject was fitted with a vest weighted by 5 percent of their own body weight. This weight was a

compromise between what was tolerable during testing and what would increase loading to the equivalent of body weight, since forces passing through the foot exceed body weight during stance [26]. The sequence of testing the seven orthotic conditions was randomized with a customized MATLAB program (The MathWorks, Inc; Natick, Massachusetts), and three scans were taken for each condition by a single operator. The probe was removed between each scan for the orthotic conditions (21 times) and the heel and arch baseline conditions (6 times).

### Analysis

Image J software (National Institutes of Health: Bethesda, Maryland) was used to measure the perpendicular distance between the navicular/calcaneus landmarks and skin surface. All images were coded to blind the observer to the orthotic condition; however, as baseline images differed considerably, they were often recognizable. A single operator carried out all measurements. Ultrasound images from 27 subjects were collected for the arch and wedge conditions. For the arch conditions, images collected from 2 subjects could not be used for analysis due to resolution difficulties.

Repeated measures analysis of variance (SPSS version 19 [IBM Corp; Armonk, New York]) was used to examine the effect of (1) arch height, (2) extrinsic wedges, and (3) intrinsic wedges, using absolute measures (millimeters) of tissue thickness ($\alpha = 0.05$). Bonferroni post hoc testing was used to examine significant main effects.

To quantify the effect of varying orthotic arch height and heel wedge, we described differences in tissue thickness between the baseline measurement (no insole) and each orthotic design as percentage change in tissue thickness.

## RESULTS

Arch soft tissue thickness at baseline was mean ± SD 29.9 ± 3.6 mm. Varying the arch height had a significant effect on soft tissue thickness ($F_{(1.6, 39)} = 70.6$, $p < 0.001$) (**Figure 4**). Post hoc testing showed that the three arch

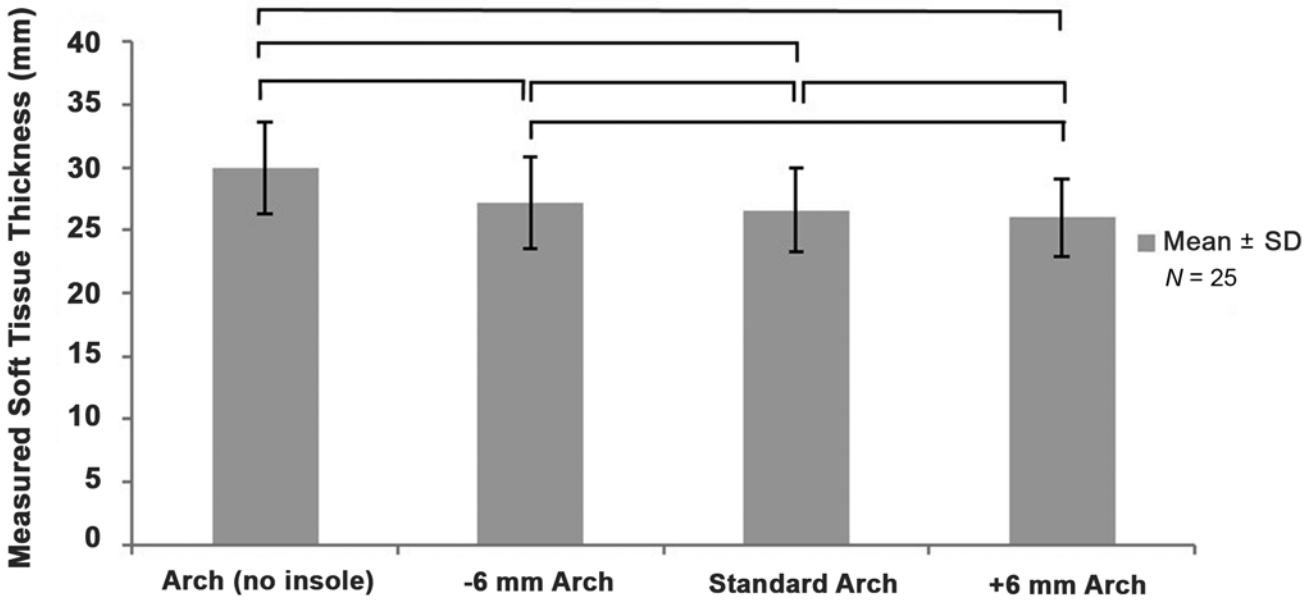


**Figure 4.**
Measured soft tissue thickness for arch baseline, –6 mm, standard, and +6 mm arch heights. Horizontal lines indicate significant differences between insole conditions. Pairwise comparisons are as follows (with Bonferroni correction): arch (no insole) to –6 mm arch ($p < 0.001$, 95% confidence interval [CI] 0.167–0.378), arch to standard arch ($p < 0.001$, 95% CI 0.231–0.434), arch to +6 mm arch ($p < 0.001$, 95% CI 0.278–0.505), –6 mm arch to standard arch ($p < 0.002$, 95% CI 0.019–0.101), –6 mm arch to +6 mm arch ($p < 0.001$, 95% CI 0.053–0.185), and standard to +6 mm arch ($p < 0.004$, 95% CI 0.016–0.104). SD = standard deviation.

heights significantly reduced tissue thickness compared with the baseline condition. The +6 mm arch height resulted in the greatest reduction of tissue thickness (11.8%; $p < 0.001$). This was followed by the standard arch height (10.2%; $p < 0.001$) and the −6 mm arch height (9.1%; $p < 0.001$). There was a 2.37 percent decrease in tissue thickness between the −6 mm and standard arch height ranges (0.4% decrease per millimeter increase in arch height). A 2.26 percent decrease was found between the standard and +6 mm arch height ranges (0.38% decrease per millimeter increase in arch height).

Heel soft tissue thickness at baseline was mean ± SD 8.6 ± 1.7 mm. The extrinsic wedge conditions had a significant effect on soft tissue thickness ($F_{(2, 52)} = 116.6$, $p < 0.001$) (**Figure 5(a)**). Post hoc testing showed that both extrinsic wedges significantly increased tissue thickness compared with baseline. The 4° extrinsic wedge increased tissue thickness by 28.3 percent ($p < 0.001$), while the 8° extrinsic wedge increased tissue thickness by 27.6 percent ($p < 0.001$). Similarly, the intrinsic wedge conditions had a significant effect on tissue thickness ($F_{(2, 52)} = 60.4$, $p < 0.001$) (**Figure 5(b)**). Post hoc testing showed that both intrinsic wedges significantly increased tissue thickness compared with baseline. The 4 mm intrinsic wedge increased tissue thickness by 23 percent ($p < 0.001$), while the 8 mm intrinsic wedge increased tissue thickness by 14.6 percent ($p < 0.001$). The 4 mm wedge caused a significantly greater increase in tissue thickness compared with the 8 mm wedge (8.3% increase; $p < 0.001$). A 4.1 percent reduction in tissue thickness was found between the 4° and 8° extrinsic wedge ranges (1.02% decrease per degree increase in extrinsic wedge). An 8.83 percent decrease was found between the intrinsic wedge ranges (2.21% decrease per millimeter increase in intrinsic wedge).

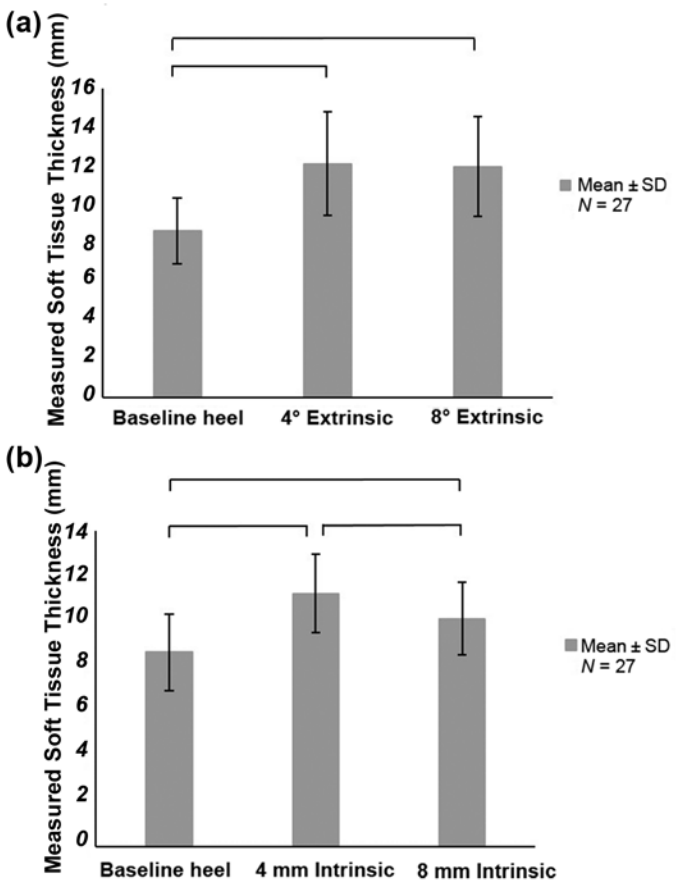


**Figure 5.**
Measured soft tissue thickness for **(a)** extrinsic and **(b)** intrinsic wedges. Horizontal lines indicate significant differences between insole conditions. Pairwise comparisons are as follows (with Bonferroni correction): baseline heel to 4° extrinsic wedge ($p < 0.001$, 95% confidence interval [CI] 0.273–0.425), baseline heel to 8° extrinsic wedge ($p < 0.001$, 95% CI 0.260–0.406), 4° extrinsic wedge to 8° extrinsic wedge ($p < 1.0$, 95% CI 0.029–0.060), baseline heel to 4 mm intrinsic wedge ($p < 0.001$, 95% CI 0.193–0.328), baseline heel to 8 mm intrinsic wedge ($p < 0.001$, 95% CI 0.084–0.215), and 4 mm intrinsic wedge to 8 mm intrinsic wedge ($p < 0.001$, 95% CI 0.064–0.158). SD = standard deviation.

## DISCUSSION

The purpose of this study was to characterize how soft tissue structures in the plantar foot respond to different APFO designs. Specifically, we sought to characterize how increasing both heel wedge (extrinsic and intrinsic) and arch height compresses soft tissue. As hypothesized, incremental increases in arch height and heel wedge (extrinsic and intrinsic) caused soft tissues to compress in a systematic manner (**Figures 4** and **5**).

The effect of the different arch heights on tissue compression was, however, small. The −6 mm and standard arch heights caused a 2.7 and 3.3 mm decrease in tissue thickness, respectively, and the +6 mm arch height resulted in a 3.9 mm decrease in tissue thickness. This 1.2 mm difference in tissue compression between −6 and +6 mm orthotic arch heights suggests that large differences in

orthotic arch heights can have similar effects on arch tissue compression. A number of factors may explain this. First, when the foot is load bearing, the plantar foot structures bear tensile forces and become stiff to resist external loads applied [27]. If soft tissues are already very stiff in the direction of vertical compression, then the orthotic arch profile may only have a small compressive effect, regardless of its geometry. Thus, stiff plantar tissues transfer load directly to bone. Second, the orthotic arch profile may have caused a neuromuscular response to avoid excessive soft tissue compression and pain in the plantar muscles and skin in the arch. This response might be considered an "avoidance tactic" under the threat of excessive muscle tissue compression in the arch due to the orthotic geometry. This neuromuscular response would adjust foot position with each increase in orthotic arch height ensuring that further compression of tissues did not occur, thus reflecting our observations that soft tissue did not significantly compress further with large changes in orthotic geometry.

The heel wedges (both extrinsic and intrinsic) significantly increased soft tissue thickness under the calcaneus compared with baseline. This increase was most likely due to the heel cup of the APFO, which prevented lateral tissue displacement in the orthotic but not baseline condition. This buttressing effect has been observed previously. A 3.57 mm increase in heel pad thickness was reported in a study using lateral radiographs to quantify the effect of a heel cup with subjects in a standing position [16]. Similarly, a 3.3 mm increase in heel pad thickness due to a heel cup was measured using in-shoe ultrasound measures while participants walked on a treadmill [23]. These values are close to those reported in the present study (3.49 and 3.30 mm increase in tissue thickness for the 4° and 8° extrinsic wedges, respectively).

No significant difference in tissue thickness under the heel was observed between the 4° and 8° extrinsic wedges. In contrast, the 8 mm intrinsic wedge resulted in a significantly reduced tissue thickness compared with the 4 mm intrinsic wedge. The extrinsic wedges, due to the confining action of the heel cup, may make the soft tissues stiffer and therefore more difficult to compress even when further wedging is applied. The observation of no further reduction in tissue thickness between the 4° and 8° extrinsic wedges might suggest the heel pad is close to maximum compression and stiffness. Such a scenario may be beneficial for transmission of force from an orthosis designed to influence joint moments. In contrast, the intrinsic wedge elevates the heel within the heel cup, which will reduce the buttressing effect, and this would be greater with the 8 mm than with the 4 mm wedge.

Inevitably, the effect of APFO arch and heel geometry on soft tissue compression was variable between subjects. Increasing the arch height had little effect on some subjects while others displayed larger reductions in tissue thickness between the arch height ranges. For example, one subject had a 0.14 and 1.58 percent decrease in tissue thickness between the –6 mm to standard and standard to +6 mm arch heights, respectively, while another had a 4.6 and 8.0 percent decrease between the –6 mm to standard and standard to +6 mm arch heights, respectively. Likewise, the same was true for both the extrinsic and intrinsic heel wedges. Some subjects experienced the greatest change in thickness (increase in thickness) with the first wedging increment (4° or 4 mm) compared with the baseline measurement, while others had greater change (decrease in thickness) with the second increment in wedging (8° or 8 mm). In the extrinsic wedges, for example, one subject had a 36.7 and 32.9 percent increase in tissue thickness for the 4° and 8° wedges, respectively. In contrast, another subject had a 31.4 percent increase for the 4° wedge and 36.3 percent increase for the 8° wedge.

The manner in which heel and arch soft tissues compress viscoelastically under load will influence how the APFO transfers load from its surface to bones, thus affecting joint moments. If the heel or arch tissues are very stiff, then the loads at the skin surface will be directly transferred to the bones. Alternatively, greater soft tissue compliance could result in loads being dissipated across internal soft tissue structures, such as the columns of collagen and fat in the heel pad and muscle in the arch. Given the difference in tissue type between the heel and the arch, it is likely that the effect of tissue compliance would be different and this may lead to differing responses at these sites. Variability in how APFOs compress soft tissues may in part explain intersubject variability in the effect of APFOs on rear foot kinematics [28–29]

There are some limitations to this study. First, static measurements of tissue thickness may not reflect how tissues behave dynamically. This is especially relevant in the context of suggested neuromuscular responses. While one approach to measuring heel pad compression during walking has been reported [23], no approach is available for arch tissues. Also, the heel pad measures in this static study are very close to those from dynamic studies [16,23]. Second, tissue compression is measured from a

point that is lateral to the navicular tuberosity. The 6 mm changes in arch height occurred at the most medial aspect of the orthosis and tapered to 0 mm at the lateral border under the cuboid. Thus, at the location where arch tissue thickness was measured, there was less than a 6 mm difference between each change in arch profile. Likewise for the heel, incremental increases in wedging (extrinsic and intrinsic) are located at a point on the orthosis that does not correspond to the point at which tissue thickness is measured. However, these measurement limitations would not affect the overall patterns observed in this study. Arguably the feet could have been tested on an orthosis with a heel cup but no heel wedging. While this would explain how heel cups without wedging affect heel tissue, our question was focused on how changes in wedge geometry affect tissues. Finally, participants did not wear footwear. This would have prevented use of our ultrasound probe but it means that constraints applied by a shoe upper on the response of the foot to the different orthotic designs were not included.

## CONCLUSIONS

This is the first study to quantify how systematic changes in arch height and two designs of heel wedging affect soft tissues under the plantar foot. The arch geometry had a significant effect on compression of soft tissues in the arch; however, compression between the ranges in arch height was small. Likewise, for soft tissues under the heel, significant increases in thickness were found with the wedges (extrinsic and intrinsic); however, only the intrinsic wedges resulted in a significant difference between the two ranges (4 mm and 8 mm). The effect of altering APFO arch and heel geometry on tissue compression under the plantar foot is variable between individuals. Tissue properties under the plantar foot affect the transfer of load from the orthosis surface to bone and thus influence how joint moments are altered by APFOs. Further work is required to understand the relationship between how foot orthoses compress soft tissues and alter foot kinetics/kinematics.

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**Author Contributions:**
*Study concept and design:* D. Sweeney, C. Nester, K. Mickle, S. Preece.
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*Data collection:* D. Sweeney, K. Mickle.
*Data processing:* D. Sweeney.
*Data analysis:* D. Sweeney.
*Statistical analysis:* D. Sweeney.
*Data interpretation:* D. Sweeney, S. Preece.
*Writing of manuscript:* D. Sweeney, C. Nester, S. Preece.
*Study supervision:* C. Nester, S. Preece, K. Mickle.
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**Participant Follow-Up:** The authors do not plan to inform participants of the publication of this study because contact information is unavailable. However, all participants were made aware at the time of data collection that data obtained would be used for publication.

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