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The immediate effect of lateral wedging on first metatarsophalangeal joint kinematics and centre of pressure

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ABSTRACT

Background: Lateral wedges are often prescribed to increase the first metatarsophalangeal joint (MPJ) range of motion or alter the centre of pressure (COP) in the foot. This study explored the effect of lateral wedge design and placement on first MPJ extension and COP during walking and running gait.
Methods: A randomised crossover design was used and 24 healthy participants ran and walked in 10 insole conditions comprising differing combinations of inclination, placement and contour. First MPJ extension and foot COP were examined. Time-series data were analysed across the stance phase using statistical parametric mapping.
Results: Lateral wedges significantly reduced first MPJ extension during both walking ($p < 0.001$; 100% of stance) and running ($p = 0.004$; 14–72%, and $p = 0.017$; 76–99% of stance). Similarly, lateral wedge placement reduced first MPJ joint extension during walking ($p < 0.001$; 100% of stance) and running ($p = 0.003$; 13–69%, and $p = 0.012$; 78–100%). Full-length or 6° lateral wedges shifted the COP medially relative to the midline of the foot ($p = 0.01$). Compared to sham, lateral wedges placed on contoured insoles exhibited a smaller reduction in first MPJ during walking ($p = 0.008$) and shifted the COP medially during both walking ($p < 0.001$) and running ($p = 0.020$).
Conclusion: Where the intention of using lateral wedging is to shift the COP medially, these data indicate that a wedge which is of higher inclination (6°) or spanning the full-length of the insole, should be used. Conversely, if the goal is to reduce extension at the first MPJ, these findings suggest that both 3° and 6° inclination wedges are suitable, with either forefoot or full-length applications proving effective.

1. Introduction

A lateral wedge is a common foot orthosis modification, included in approximately one-third of designs [1,2]. This modification tapers from thick on the lateral side to thin on the medial side and is frequently used to alter the centre of pressure (COP) trajectory or affect the function of the first metatarsophalangeal joint (MPJ) [1]. The medial-lateral position of COP is associated with frontal plane kinematics of the foot and lower limb [3]. Where the COP lies relative to joint axes is thought to influence joint moments, and subsequently, the function of the foot [4].

Theoretically, first MPJ extension is essential for efficient propulsion during gait and is often linked clinically to the Windlass mechanism [5–8]. First proposed by Hicks [1954], based on observations in cadaveric specimens, the Windlass mechanism proposes an association whereby extension of the first MPJ results in elevation of the medial longitudinal arch and necessitates downward movement of the first metatarsal head. For several decades, the Windlass mechanism has persisted as a fundamental concept often used to explain or clinically

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rationalise the function of foot orthoses [6,8–11]. Gomez-Carrion et al. [2024] demonstrated a 39% reduction in the force required to dorsiflex the hallux with the use of a kinetic wedge (a modification commonly placed at the forefoot of an orthosis); these authors suggested that this would enhance engagement of the Windlass mechanism. Limited extension of the first MPJ has been associated with the development of foot pathologies such as plantar heel pain, overload of the lesser metatarsals, and compensatory changes to gait [8,12–17].

Current evidence exploring the influence of lateral wedges lacks sufficient consideration of wedge design and has not compared their effects between walking and running [18]. Considering the differences between gait types, such as the increased vertical ground reaction force and reduced contact time during running, it is reasonable to assume that a lateral wedge, which aims to influence load and temporal parameters of gait, would act differently when used for walking compared to running [19–22]. A recent survey of podiatrists revealed that 70% believe lateral wedging functions differently during walking compared to running gait [1], despite the lack of supporting evidence [18]. It has also been shown that first MPJ kinematics and COP vary between walking and running gait [23,24]. These outcomes have not been investigated in a manner that reflects clinical practice, despite being the two most common reasons for lateral wedge use. Notably, the majority of studies employ full-length lateral wedges with relatively steep inclinations (typically around 6°), whereas in clinical settings, wedges are more commonly applied to the forefoot and at a lower angle (2–3°) [1, 18]. This disconnect limits the clinical applicability of current findings and highlights a critical need for research that better mirrors the conditions under which lateral wedges are prescribed clinically. The aim of this study was to explore the effect of lateral wedges on first MPJ extension and COP and compare these outcomes between walking and running gait, whilst considering the influence of insole and wedge design (contour, inclination, and placement).

2. Methods

2.1. Study participants

Twenty-five individuals who regularly ran (minimum 10 km/week), were uninjured for the previous 6 months, were over 18 years of age, and were comfortable running on a treadmill, were recruited to participate. A minimum weekly running volume of 10 km was chosen to ensure participants had consistent exposure to the mechanical demands of running. This threshold reflects habitual activity and aligns with definitions of recreational runners in previous literature [25,26]. Prospective participants were excluded if they did not meet the above criteria or reported a systemic medical condition affecting their joints. Power calculations (conducted with RStudio version 3.6.3) were based on data from Turner et al. [27] who reported a standard deviation (SD) of 6.9° (mean 29.2°) when measuring peak first MPJ dorsiflexion in walking gait from a cohort of 25 healthy adults. Therefore, 25 participants would provide a detectable effect size of 1.4° assuming 90% power and an alpha level of 0.05. The institutional ethics committee approved the study (AUTEC# 22/121), and participants provided informed consent.

2.2. Study protocol

Participants wore standardised footwear (ASICS Gel Cumulus 23) for all trials. After an initial five-minute warm-up, participants walked (1.4 m/s) and then ran (2.8 m/s) for two minutes in each insole condition. A standardised speed was used to minimise speed-related variability [28]. Similar speeds have been used in previous research [29,30]. Data were collected during the final 30 s of each trial. Ten insole conditions were examined using a randomised crossover design (supplementary material). These included five contoured insoles (Formthotics™ Original Single Medium, Foot Science International, Christchurch, NZ) and five sham insoles (identical material but with no arch profile or other contour), each with matched combinations of forefoot or full-length and 3° or 6° inclination lateral wedges. Both the insoles and lateral wedges were made from polyethylene foam with a density of 140 kg/m^3 and hardness of approximately 61 Asker C. Full-length wedges ran from the most proximal aspect of the insole to the most distal aspect. In contrast, forefoot wedges extended from the calcaneo-cuboid joint to the sulcus.

2.3. Three-dimensional gait experimental setup

Kinematic data were gathered using an 8-camera motion capture system (Vicon, Oxford, UK) recording at 200 Hz, whilst kinetic data were simultaneously collected from a synchronised instrumented treadmill (Bertec, Columbus, OH, USA) at 1000 Hz. Rigid clusters of four retro-reflective markers were positioned on the right anterior shin and lateral thigh. Individual reflective markers were placed on the anterior and posterior superior iliac spines (bilaterally) and to the right greater trochanter, medial and lateral femoral epicondyles, and medial and lateral malleoli. Additionally, markers were placed on the posterior of the right shoe at the superior aspect of the calcaneus as well as the medial and lateral aspect just above the midsole, the medial apex of the sustentaculum tali, the lateral apex of the peroneal tubercle, the base and head of each the first, second and fifth metatarsals and finally, the most dorsal and distal aspect of the first proximal phalanx [31]. A static calibration was initially collected, and the knee and ankle markers were removed for the subsequent trials.

2.4. Data processing

Initial marker labelling and gap filling were conducted in Vicon Nexus (v2.12.1, Vicon, Oxford, UK). Data were cropped to 20 steps and exported to Visual3D (C-Motion Inc., MD, United States). Filter cutoffs were set to 15 Hz for the kinematic and kinetic output. An adapted version of the Leardini model [32] previously used by Weir et al. [2018] was used to model the foot. First MPJ kinematics were calculated as a planar angle using vectors created by a virtual marker (150% offset axially from the first metatarsal shaft) and two physical markers on the first metatarsal head and hallux, then projected onto the YZ (sagittal) plane with the forefoot segment as a reference. Foot COP was calculated for the right foot segment with the resolution coordinate system set to a virtual foot segment. Centre of pressure data represented the position from the anteroposterior axis, normalised to foot width (calculated as the distal radius of the foot segment), where a negative number represented a location medial to the axis, and a positive number indicated a lateral position. For example, a COP of −1 indicates 100% of the radius of the foot (half the distance between markers placed on the first and fifth metatarsal heads), medial to the longitudinal axis. All time-series data were interpolated to 101 points for subsequent statistical analysis.

2.5. Statistical analysis

Continuous demographic data were analysed using mean and SD, demographic data were presented as raw values and percentages. Biomechanical data, including first MPJ extension and COP position, were tested across the entire time series of the stance phase using statistical parametric mapping (SPM) and open-source code SPM1d [33]. All statistical analyses were conducted in Matlab (Mathworks Inc., MA, USA). A significant portion of the data violated the assumption of data normality, so non-parametric tests were applied. The main effect (for wedge inclination, placement, and insole contour) was examined using a Friedman test, and additional pairwise comparisons were conducted using the Wilcox signed rank test to determine the within-group effect of inclination and placement. For the main effect, the alpha level was set at 0.05.

Table 1
Participant demographic and clinical data.

Characteristic	Mean (SD)
Sex	9 women (38%), 15 men
Age (years)	36.6 (8.1)
Weekly running mileage (km)	41.9 (22.6)
Typical weekly running frequency (days per week)	4.3 (1.7)
Body mass (kg)	72.8 (8.9)
Height (m)	1.75 (0.08)
Body mass index (kg/m^2)	23.8 (2.5)
Foot Posture Index	2.5 (3.1)
Maximum first MPJ extension (NWB) (°)	76.0 (12.1)
First MPJ range of motion (WB) (°)	46.3 (7.6)

Kg, kilograms; Km, kilometres; m, meters; NWB, non-weight bearing; WB, weight-bearing; °, degrees

3. Results

3.1. Participants

Twenty-five participants were recruited for the study (Table 1); however, data from one participant were excluded due to data corruption. Therefore, data from 24 participants were analysed. Table 2 provides an overview of COP and 1st MPJ kinematic outcomes.

3.2. First metatarsophalangeal joint kinematics

3.2.1. Wedge inclination – walking

Three-degree wedges significantly reduced first MPJ extension between 0% and 17% ($p = 0.002$), 21–24% ($p = 0.021$), 35–41% ($p = 0.015$), 52–56% ($p = 0.021$) and 58–87% ($p = 0.003$) of the stance phase. Compared to a control condition, 6° wedges significantly reduced first MPJ extension ($p < 0.001$). A reduction in first MPJ extension (with wedges) occurred between 0% and 95% of stance in walking. Pairwise comparisons between the two wedge inclinations (3° vs. 6°) were only significant in walking (p=0.001). During walking, the 6° wedge significantly reduced first MPJ extension, compared to the 3° condition, during two periods of stance (3–35% and 38–85%) (Fig. 1).

3.2.2. Wedge inclination – running

Pairwise comparison between 3° and 6° wedges revealed no significant difference between the two inclinations for running. Compared to the control condition, 3° wedges significantly reduced first MPJ extension between 14% and 100% of stance ($p < 0.001$). A reduction in first MPJ extension was observed between 0% and 100% when comparing the 6° wedge to the control ($p < 0.001$) (Fig. 2).

3.2.3. Wedge placement – walking

Significant differences were found between full-length and forefoot wedges compared to the control, and when comparing full-length to forefoot wedges. Differences between the forefoot wedges and the control were found between 2% and 15% ($p = 0.003$), 38–39% ($p = 0.024$),

Table 2
Main effect and pairwise comparison of design factors on COP and first MPJ kinematics.

Outcome measure	Factor	Gait type (bold font indicates significant effect)	Region of significance (% of stance [P])	Pairwise comparison (bold font indicates significant effect)	Region of significance (% of stance [P])	Mean difference
COP	Inclination	**Walking**	14 – 20 [0.009]	Control vs. 3°	-	-
				Control vs. 6°	12 – 22 [0.005]	0.03*
				3° vs. 6°	-	-
		Running	29 – 50 [0.013]	Control vs. 3°	-	-
				Control vs. 6°	28 – 52 [0.003]	0.04*
				3° vs. 6°	-	-
	Placement	**Walking**	14 – 23 [0.008]	**Control vs. full-length**	11 – 24 [0.001]	0.04*
				Control vs. forefoot	-	-
				Full-length vs. forefoot	23 – 27 [0.022]	0.03*
		Running	25 – 70 [0.005]	**Control vs. full-length**	26 – 52 [0.002]	0.03*
				Control vs. forefoot	-	-
				Full-length vs. forefoot	-	-
	Contour	**Walking**	10 – 78 [< 0.001]	**Sham vs. Contoured**	10 – 77 [< 0.001]	0.03*
		Running	13 – 18 [0.020]	**Sham vs. Contoured**	13 – 19 [0.009]	0.03*
First MPJ	Inclination	**Walking**	0 – 98 [< 0.001]	**Control vs. 3°**	0 – 17 [0.002]	0.86°
					21 – 24 [0.021]	0.74°
					35 – 41 [0.015]	0.80°
					52 – 56 [0.021]	0.91°
					58 – 87 [0.003]	1.29°
				Control vs. 6°	0 – 95 [< 0.001]	1.53°
				3° vs. 6°	3 – 35 {0.001]	0.43°
					38 – 85 [< 0.001]	0.56°
		Running	14 – 72 [0.004] 76 – 99 [0.017]	**Control vs. 3°**	14 – 100 [< 0.001]	1.89°
				Control vs. 6°	0 – 100 [< 0.001]	1.80°
				3° vs. 6°	-	-
	Placement	**Walking**	0 – 99 [< 0.001]	**Control vs. full-length**	0 – 100 [< 0.001]	1.60°
				Control vs. forefoot	2 – 15 [0.003]	0.79°
					38 – 39 [0.024]	0.79°
					50 – 53 [0.022]	0.92°
					55 – 84 [< 0.001]	1.27°
				Full-length vs. forefoot	1 – 26 [0.003]	0.59°
		Running	13 – 69 [0.003] 78 – 100 [0.012]	**Control vs. full-length**	0 – 100 [< 0.001]	2.48°
				Control vs. forefoot	77 – 94 [0.007]	1.70°
				Full-length vs. forefoot	-	-
	Contour	**Walking**	31 – 75 [0.008]	**Sham vs. Contoured**	49 – 73 [< 0.001]	0.54°
		Running	-	-	-	-

COP, centre of pressure, MPJ, metatarsophalangeal joint; °, degrees, *radii from the midline (normalised to foot width)

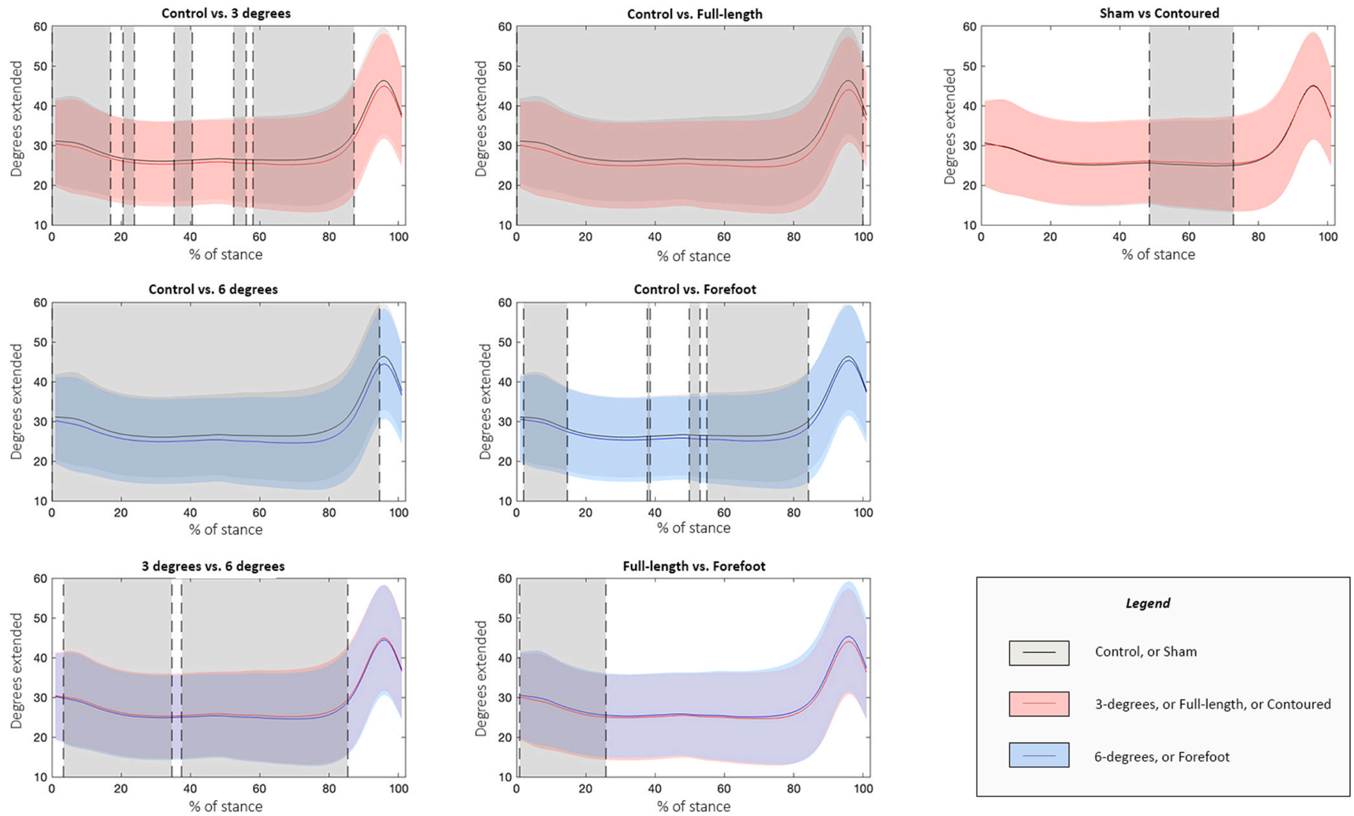

Fig. 1. First MPJ joint kinematics and SPM analysis for walking gait. The x axis represents 0–100% stance phase. The y-axis is the first MPJ position (degrees extended). Vertical shading represents periods of (SPM) significance. Coloured shading represents standard deviation. Note, the purple shadowing is a result of red being overlaid on blue.

50–53% ($p = 0.022$) and 55–84% ($p < 0.001$) of stance. Full-length wedges exerted a significant difference from 0% to 100% ($p < 0.001$) of stance compared to the control. Compared to forefoot wedges, those that ran the full length of the insole significantly reduced first MPJ extension between 1% and 26% of stance in walking ($p = 0.003$) (Fig. 1).

3.2.4. *Wedge placement – running*

Pairwise comparisons revealed a difference when contrasting wedge placement (full-length or forefoot) to the control in running; however, there was no significant difference when directly comparing placements. Compared to the control, forefoot wedges reduced first MPJ extension between 77% and 94% of stance ($p = 0.007$). Full-length wedges reduced the joint's extension for the entire stance (0–100%, $p < 0.001$) (Fig. 2).

3.2.5. *Insole contour – walking*

The contour of the insole onto which the wedge was placed demonstrated a significant effect during walking (49–73% of stance, $p = 0.008$). Participants exhibited a greater reduction in first MPJ extension when the wedge was placed on a sham insole (Fig. 1).

3.2.6. *Insole contour – running*

There were no significant differences in first MPJ extension during running between wedges placed on a contoured and sham insole (Fig. 2).

3.3. *Centre of pressure*

3.3.1. *Wedge inclination*

Six-degree wedges created a medial shift in the foot COP compared to the control condition from 12% to 22% of stance ($p = 0.005$) during walking (Fig. 3) and from 28% to 52% of stance ($p = 0.003$) during running (Fig. 4). No significant differences were observed between 3° wedges and the control, nor between 3° and 6° wedges.

3.3.2. *Wedge placement*

Compared to the control, full-length wedges elicited a medial shift in COP from 11% to 24% of stance ($p = 0.001$) during walking (Fig. 3) and 26–52% of stance ($p = 0.002$) during running (Fig. 4). Contrasting full-length to forefoot wedges, a difference was observed between 23% and 27% of stance ($p = 0.022$) during walking, whereby full-length wedges shifted the COP medially compared to forefoot wedges. No significant differences were observed between the two wedge placements for running gait or when contrasting forefoot wedges to the control for either gait type.

3.3.3. *Insole contour*

Wedges applied to the sham insole shifted the COP more medially than those on contoured insoles for both walking (10–77% of stance, $p < 0.001$) (Fig. 3) and running (13–19% of stance, $p = 0.009$) (Fig. 4).

4. Discussion

This study explored the effect of lateral wedge inclination and placement on first MPJ extension and foot COP during walking and running. A reduction in first MPJ extension was observed with lateral wedge use regardless of wedge placement or inclination, for both walking and running. Data also demonstrated a medial shift in COP with lateral wedging.

The reduction in the first MPJ extension contrasts Du et al. [2024], who reported no significant influence of lateral wedging on first MPJ function during walking. Differing wedge design may partly explain these contrasting results. Du et al. [2024] used 7- and 10-mm material, the inclination of which depended on the width. Du et al. [2024]

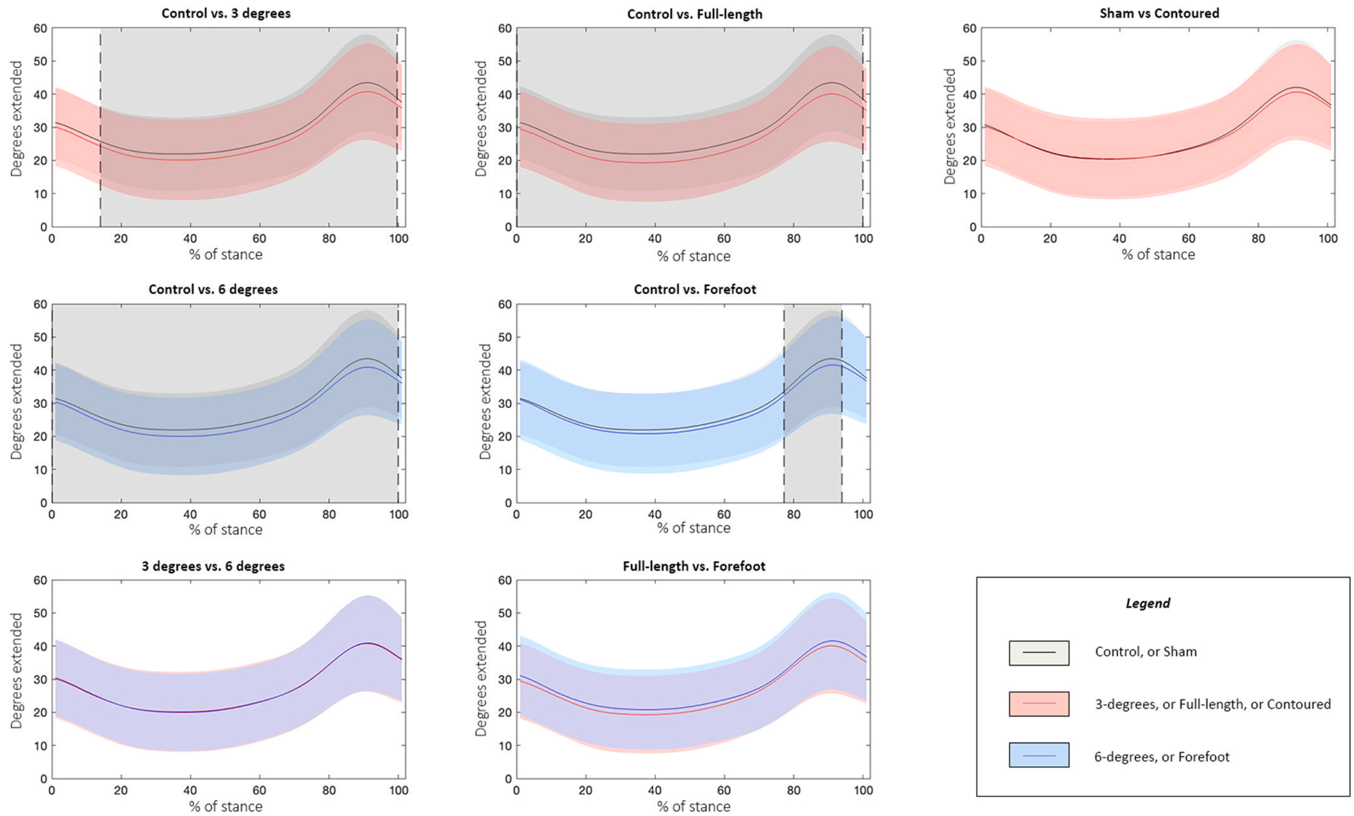

Fig. 2. First MPJ joint kinematics and SPM analysis for running gait. The x axis represents 0–100% stance phase. The y-axis is the first MPJ position (degrees extended). Vertical shading represents periods of (SPM) significance. Coloured shading represents standard deviation. Note, the purple shadowing is a result of red being overlaid on blue.

reported using material with a density of 940 kg/m^3, which exceeds the typical maximum density of materials used in foot orthoses and may have been a reporting error. By contrast, the current study used wedges made from polyethylene foam with a density of 140 kg/m^3.

The reduction in the first MPJ extension may be associated with the increased plantar loading of the first metatarsal head, indicated by the medial shift in COP. Although these variables were assessed independently in this study and any relationship can only be inferred theoretically, the patterns observed are consistent with plausible biomechanical interactions between plantar pressure distribution and first MPJ kinematics. Theoretically, the Windlass mechanism requires downward displacement of the first metatarsal head for the first MPJ to extend; thus if the metatarsal head is unable to plantarflex adequately, first MPJ extension may be mechanically constrained. Several previous studies have supported this association between the metatarsal head position and first MPJ kinematics [35–39]. For example, Roukis et al. [1996] reported a 19% decrease in first MPJ extension when the first metatarsal head was elevated using an orthosis, reinforcing the concept that altered loading or positioning of the first ray can influence first MPJ motion.

Kinematic coupling within the foot may also contribute to reducing first MPJ extension [40]. Williams et al. [41] demonstrated that a 19% restriction in first MPJ range of motion reduced midtarsal motion by approximately 10%. Similarly, Manfredi-Marquez et al. [42] observed that midfoot supination occurred concurrently with first MPJ extension, suggesting bidirectional interactions between these segments. Lateral wedging has been shown to induce navicular abduction and plantarflexion [43], and although this study did not measure these motions, theoretical extrapolation suggests that such midfoot movements could influence first MPJ motion through kinematic coupling mechanisms. Together, these theoretical considerations highlight how changes in plantar loading patterns, such as the observed medial COP shift, may interact with first MPJ motion, even though direct causal relationships cannot be established from the present data.

The second main finding of this study was that, compared to the control condition, 6° or full-length wedges shifted the COP medially. This contradicts previous research, which suggests that lateral wedges shift the COP laterally [30,34,44–50]. Existing research provides limited insight into how lateral wedges influence lateral shifts in COP, particularly given that this effect differs from those observed with medial or midsole wedge placements. Variability in methodological approaches across studies further complicates the interpretation, making it difficult to isolate the mechanisms underlying the conflicting results identified in the present study. Medial forefoot wedges, for example, have also been shown to shift the COP laterally [44]. Lateral wedges built into the midsole of the shoe shift COP medially [51]. Zhang et al. [2022] proposed that the medial toe box being generally deeper than the lateral toe box could account for both medial and lateral wedges shifting COP laterally. These authors hypothesised that a lack of space in the lateral toe box increased pressure between the foot and the insole when a lateral wedge was in place. If this assumption is true, the deep lateral toe box of shoes used in the current study may have allowed the lateral wedges to function without the internal pressure produced by the shoe upper, and therefore function more naturally, shifting the COP medially.

The current study found no difference between 3° and 6° wedges, except for their effect on first MPJ extension during walking; here, 6° wedges further reduced first MPJ extension compared to 3° wedges. This contrasts with previous evidence, which suggested that higher inclination wedges elicited larger biomechanical effects [30,44,51–53]. Within the context of healthy foot mechanics, the present study suggests that increasing wedge inclination beyond 3° does not appear to yield additional changes in first MPJ kinematics or COP during running, and only minimal change during walking. Previous studies have also noted that wedges steeper than 5° may reduce comfort and tolerance [49,51,54–56], and given the similar biomechanical effects observed here, a

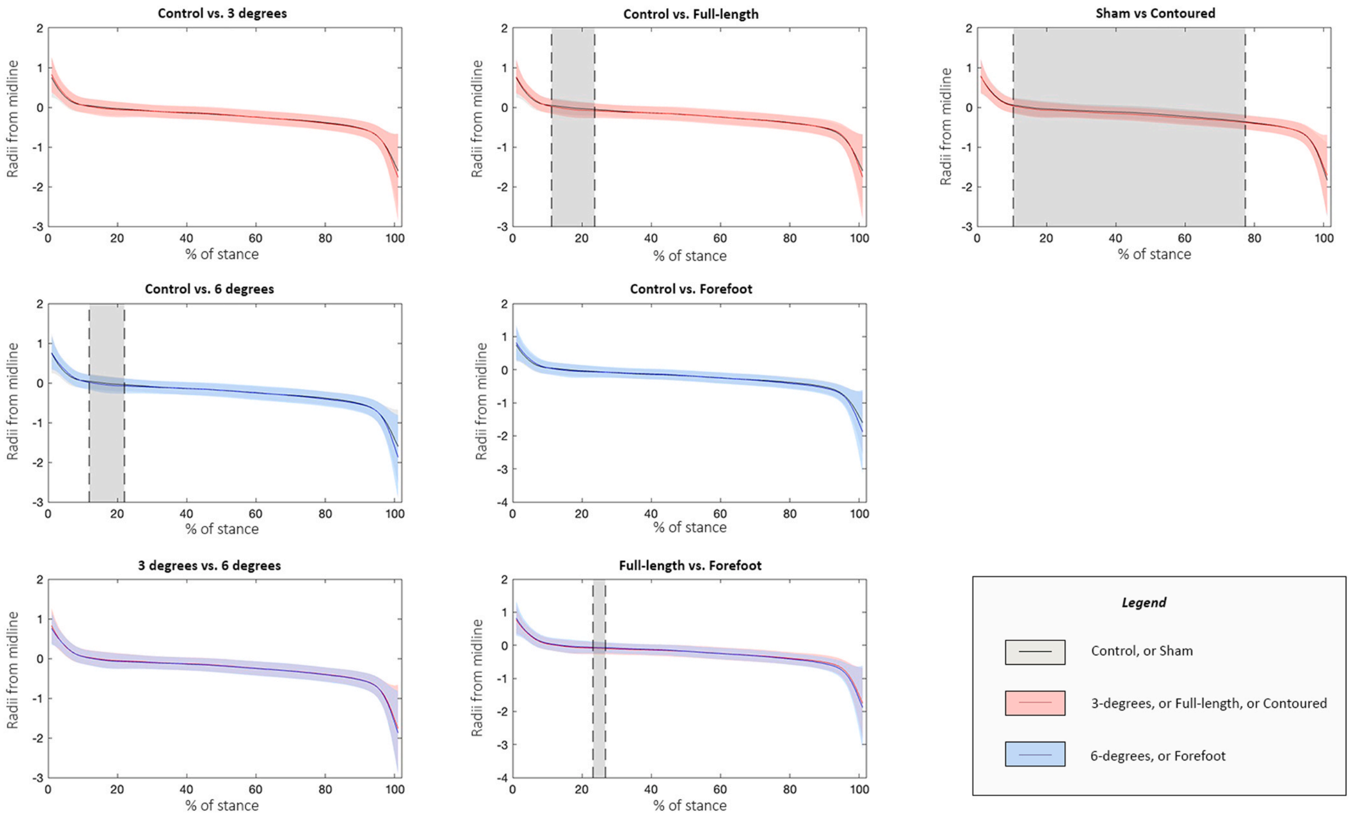

Fig. 3. COP trace and SPM analysis of walking gait. X axis represents 0–100% stance phase. The y-axis represents COP normalised to foot width and presents the number of radii from the midline. A positive value indicates lateral deviation, and a negative value indicates medial deviation. Vertical shading represents periods of (SPM) significance. Coloured shading represents standard deviation. Note, the purple shadowing is a result of red being overlaid on blue.

lower inclination may therefore be preferrable for healthy individuals. Importantly, these results should not be generalised to clinical populations; individuals with pathological lower limb conditions may exhibit altered joint mobility or pain-related inhibition and could respond differently to higher wedge inclinations, potentially showing larger or more clinically meaningful changes in response to increased wedge angles.

Based on the current data, wedges placed under the forefoot did not influence COP and reduced first MPJ extension over a shorter period of stance, compared to full-length wedges. This supports previous findings, where it has been reported that forefoot wedges did not influence pressure distribution in the rearfoot [57]. Regarding insole contour, this study found a greater reduction in the first MPJ extension when wedges were placed on sham insoles during walking. The authors posit that the contoured arch may elevate the proximal base of the first metatarsal and lead to an associated plantar movement of the metatarsal head, evoking engagement of the Windlass mechanism and encouraging first MPJ extension. Interestingly, for both walking and running, wedges placed on contoured insoles shifted the COP medially, compared to when placed on a sham insole. This effect only occurred for a short period of stance, in the case of running (13–19% of stance). Research to understand the impact of sham versus contoured insoles is currently unclear and conflicting [45,58,59]. Jones et al. [2013] found no difference between lateral wedges placed on a contoured insole and those on a sham insole for a range of kinetic outcomes, including COP. It was also recently reported that contoured insoles were no more effective than sham insoles at reducing pain or self-reported physical function and quality of life in patients with first MPJ osteoarthritis (OA) [58].

Despite the biomechanical distinction between walking and running, there is limited evidence contrasting the effect of lateral wedges between gait types [18,19]. The present study found that lateral wedges exerted an effect on COP earlier in stance during walking (12–27%) compared to running (26–52%). Forefoot wedges compared to unwedged insoles reduced first MPJ extension over a greater period of stance phase for walking (2–84%) compared to running (77–94%). These differences may relate to the foot spending longer on the ground during walking, or the broader range of strike patterns observed in running [19,60].

Full-length wedges and wedge inclinations (3° and 6°) influenced first MPJ kinematics over similar periods of stance for walking and running. However, the reduction in extension was greater during running. This supports previous evidence that orthoses exert a larger effect in running compared to walking [61]. Mager et al. [62] identified lower first MPJ joint stiffness in running compared to walking, despite participants displaying no kinematic differences. The lower joint stiffness during running may create a greater capacity for the wedge to make a difference. Previous reports have also indicated that the influence of shoes on foot and lower limb kinematics may be greater during walking than during running [63].

This work must be considered in the context of its limitations. Notably, reflective markers were placed on the shoe rather than directly on the foot. Both skin- and shoe-mounted markers provide an approximation of joint motion and may not be directly comparable [64], however, shoe-mounted markers tend to underestimate motion compared to skin-mounted markers [65–67]. To minimise this potential issue, shoe markers were not removed between conditions and were checked for accuracy by an experienced biomechanical podiatrist following each insole replacement. Additionally, the study did not control foot strike pattern, which may influence lower limb biomechanical responses to lateral wedging during running.

Although lateral wedges are frequently used in the management of conditions such as medial knee OA, chronic ankle instability, and plantar heel pain, the current study considered only healthy individuals [1,68]. Therefore, these findings may not be transferable, and the clinical implications remain unclear. Some small differences in first MPJ

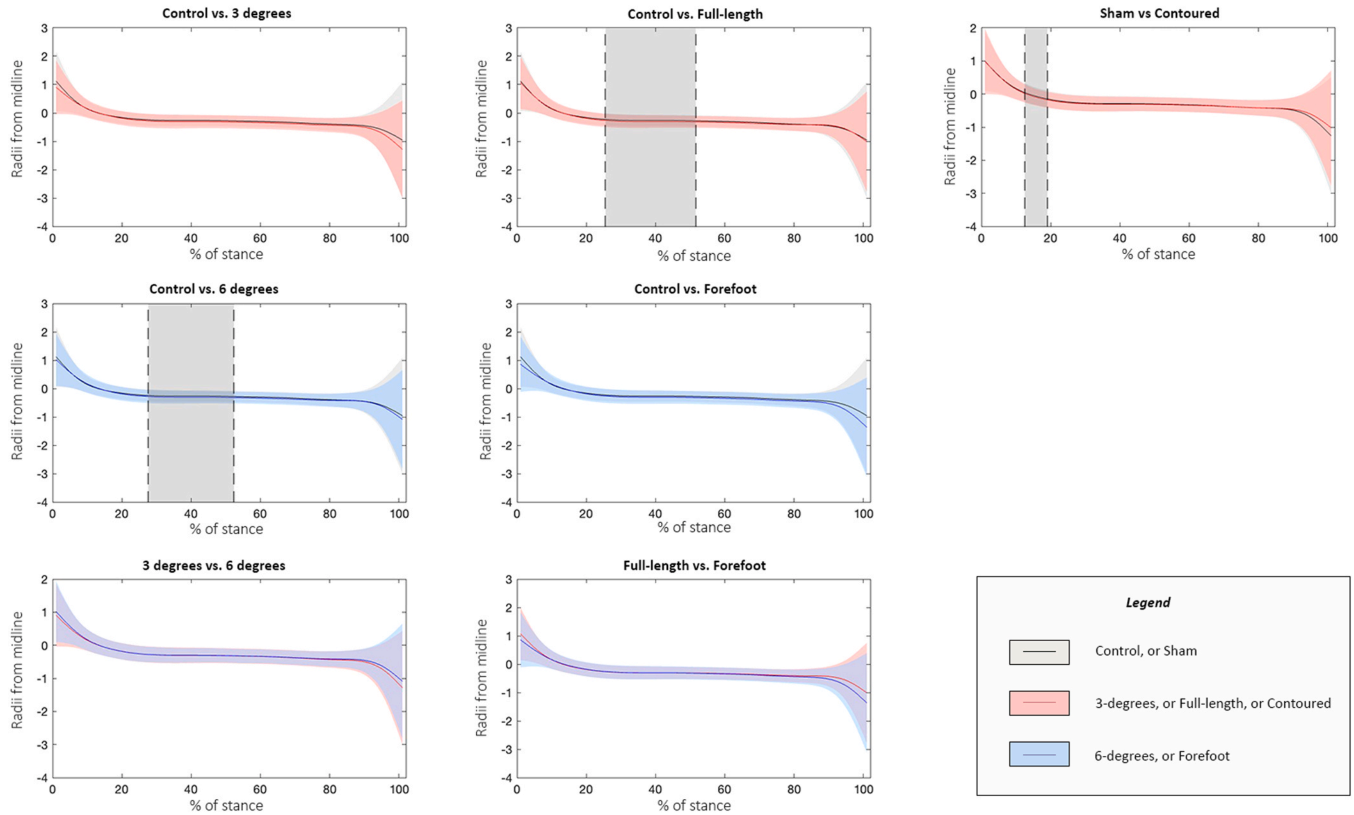

Fig. 4. COP trace and SPM analysis of running gait. X axis represents 0–100% stance phase. The y-axis represents COP normalised to foot width and presents the number of radii from the midline. A positive value indicates lateral deviation, and a negative value indicates medial deviation. Vertical shading represents periods of (SPM) significance. Coloured shading represents standard deviation. Note, the purple shadowing is a result of red being overlaid on blue.

dorsiflexion observed between wedge conditions (e.g., 0.43°–0.56°) may fall below clinically meaningful thresholds. While these changes were statistically detectable, their functional relevance should be interpreted with caution, given the absence of established minimal clinically important difference (MCID) values for first MPJ motion. Future research should seek to understand the MCID for first MPJ motion and establish if these effects are seen in symptomatic populations frequently prescribed lateral wedges. There have been previous indications that the effect of lateral wedging may depend on individual foot posture [50]. It would be clinically useful to know if this applies to COP and first MPJ extension. Lastly, this study did not elucidate the effects of design combinations (such as the effect of 3-degree inclination of forefoot, compared with full-length, wedges), each variable was considered in isolation.

5. Conclusion

The current study demonstrated that lateral wedges reduce first MPJ extension during walking and running. First MPJ extension was reduced regardless of inclination or placement; however, in walking, 6° wedges and those placed full-length reduced joint extension more than other designs. Lateral wedges caused a medial shift in COP, consistent for both 6° and full-length wedges. Data demonstrated that the influence of lateral wedges varied between walking and running. The outcomes were more pronounced and occurred over a greater portion of stance while running than when walking. These findings provide new insights into the biomechanical function of lateral wedges, offering a biomechanical baseline for the effects of lateral wedge design. The results emphasise the importance of lateral wedge placement and design, as insole contour, inclination, and length affect first MPJ extension and COP differently during walking and running.

Abbreviations

COP	Centre of pressure
MPJ	Metatarsophalangeal joint
OA	Osteoarthritis
SPM	Statistical parametric mapping

Ethics approval

The Auckland University of Technology Ethics Committee (AUTEC) granted ethical approval for application reference 22/121.

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CRediT authorship contribution statement

Hannah Wyatt: Resources, Methodology, Formal analysis. **Kelly Sheerin:** Writing – review & editing, Supervision, Resources, Methodology, Conceptualization. **SangHoon Yoon:** Visualization, Software, Formal analysis. **Aaron Jackson:** Writing – review & editing, Writing – original draft, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Duncan Reid:** Supervision, Methodology, Conceptualization. **Matthew R. Carroll:** Writing – review & editing, Supervision, Methodology, Conceptualization.

Declaration of Competing Interest

All authors declare no competing interests.

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.gaitpost.2026.110152.

Availability of data and material

Requests for further details of the data set and queries relating to data-sharing arrangements may be submitted to Aaron Jackson (aaron.jackson@aut.ac.nz).

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