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A NOVEL SIMPLIFIED BIOMECHANICAL ASSESSMENT OF THE HEEL PAD DURING FOOT PLANTARFLEXION

ABSTRACT

The heel pad (HP) which is located below the calcaneus comprises a composition of morphometrical and morphological arrangements of soft tissue that are influenced by factors such as gender, age and obesity. It is well known that HP pain and Achilles tendonitis consist of discomfort, pain and swelling symptoms that usually develop from excessive physical activities such as walking, jumping and running. The purpose of this study was to develop biomechanical techniques to evaluate the function and characteristics of the HP. Ten healthy participants (five males and five females) participated in this laboratory-based study, each performing a two-footed heel raise to mimic the toe-off phase during human locomotion. Twenty-six (3mm) retroreflective markers were attached to the left and right heel (thirteen markers on each heel). Kinematic data was captured using three-dimensional motion analysis cameras synchronised with force plates. Descriptive and multivariate statistical tests were used in this study. In addition, a biomechanical technique that utilises only six markers from twenty-six markers to assess HP deformation and function has been developed and used in this study. Overall HP displacement was significantly higher in males on the most lateral part of the right heel ($p < 0.05$). No significant differences were evident when comparing the non-dominant and dominant heels during the baseline, unloading and loading phases ($p > 0.05$). Findings from this study suggested that biomechanical outputs expressed as derivatives from tracked HP marker movements can morphologically and morphometrically characterise HP soft tissue deformation changes. The outcome of this study highlights the importance of 3D motion analysis being used as a potential prospective intervention to quantify the function / characteristics of the heel pad soft tissues.

***KEY WORDS*:** Heel Raise, Heel Pad, Toe-off, Unloading Phase, Loading Phase, Gait, Marker Displacement

INTRODUCTION

The visco-elastic behaviour of a healthy heel pad (HP) has been found to be beneficial in providing a positive rebound during walking by absorbing kinetic energy and restricting the displacement of the HP when the foot makes contact with the ground[1]. Several studies have indicated that factors such as gender, age and obesity may influence the composition of the soft tissue within the HP[2]. Research by Rome et al (2001) reported that runners with a higher body mass index (BMI) may have a lower HP stiffness which could enhance the heel pads ability to absorb higher mechanical shock associated with the increased mass[3]. In addition, age related-changes have been shown to increase the thickness and decrease the elasticity of the HP which may dampen the soft tissues response to high impact velocities during locomotion[4]. The influence of gender on the biomechanical properties of the HP is controversial with research suggesting that males may have thicker and stiffer heels compared to females[5]. In contrast, research by Taş (2018) highlighted that gender may only have a minimal impact on any structural changes (e.g., stiffness and thickness) of the HP[6].

If the HP is insufficient at absorbing impact forces during walking and other impact activities due to degenerative changes within the fibrous structure, individuals may be more vulnerable to injuries such as plantar fasciitis[7]. Pain associated with plantar fasciitis typically occurs in the medial part of the HP and has been shown to negatively impact on daily functional movements such as walking which could have detrimental effects on health and quality of life[8]. Research has indicated that individuals with plantar fasciitis may develop thicker and stiffer heels[9]. Moreover, it is proposed that the dominant leg tends to displace / deform the HP and surrounding soft tissues less than the non-dominant

leg and that the lack of strength within muscles surrounding the non-dominant foot may contribute to plantar fasciitis[10, 11]. Therefore, a larger tissue displacement / deformation within the non-dominant leg may influence displacement in the inferior parts of the kinetic chain such as the HP, however, no research has investigated this hypothesis. Furthermore, measuring the mechanical function of a healthy and diseased HP using motion analysis may provide a non-invasive screening technique targeted towards the clinical management of patients with plantar heel pain.

Previous studies have examined the properties of the HP using *in vitro* methods on cadaveric human feet[12, 13] and have analysed the physical properties of the HP within a controlled loading environment by utilising measurements such as pendulum tests[13]. However, these studies did not elucidate the structural differences in the planes of measurement of the HP. More recently, ultrasound and indention devices have been used to measure the thickness and stiffness of the HP[2, 14]. Whereas, motion analysis has only been used in one previous study to investigate HP thickness[15]. A smaller degree of HP deformation can be measured objectively when using motion analysis in comparison, to traditional methods like pendulum testing (Chi and Schmitt, 2005). It has been suggested by Aerts et al. (1995) that methods such as pendulum testing may be more likely to overestimate results because of the inclusion of other tissues when HP deformation is taking place[16]. At present, no studies have investigated the displacement / deformation of the HP using motion analysis.

Despite most of the research measuring properties such as stiffness, elasticity and thickness, there is still a lack of studies investigating the displacement of the HP using

non-imaging techniques during human locomotion. Therefore, the purpose of this present study was to develop biomechanical techniques to assess the function and characteristics of the HP. The function of the HP will be investigated using the following techniques (a) morphometrically and morphologically assess the HP, (b) calculate the soft tissue compliance and strain of the HP, (c) determine how gender and the dominant versus non-dominant heel influences HP displacement during foot plantarflexion (replicating the toe-off phase of gait); and (d) assess how the findings in (a, b and c) affect the baseline unloading and loading phases during foot plantarflexion.

METHODS

Participants

After receiving ethical approval from the university ethics committee, five males (age: 26.4 ± 13.0 years, height: 177.7 ± 3.8 cm, body mass: 73.5 ± 8.3 kg; mean ± SD) and five females (age: 27.2 ± 15.2 years, height: 161.9 ± 3.0 cm, body mass: 59.8 ± 6.6 kg; mean ± SD) were recruited for this laboratory-based study. Participants were required to have no Achilles injury, plantar heel pain or surgery within the heel in the 12 months before data collection. Prior to testing, the researchers explained the procedure and purpose of the study to all the participants. All participants completed a consent form and physical readiness questionnaire before testing commenced.

Apparatus

Two force plates, BP400600 AMTI Optima Human Performance System (AMTI, Watertown, United States) sampling at 1000 Hz were embedded in concrete and used in the analysis. Eight Vicon Nexus Bonita Motion Analysis (Oxford Metrics Ltd, United

Kingdom) cameras were placed on tripods and positioned in a semi-circle surrounding the force plates. Kinematic data capture was sampled at 250 Hz. The height of the cameras on tripods ranged from 44 cm to 77 cm. This choice of camera height, position and configuration provided an optimal position for the cameras to capture and record the trajectories of the retroreflective markers during the data capture session. Both kinetic and kinematic output data were captured simultaneously.

Experimental Design

This laboratory-based experiment required one visit to the biomechanics laboratory. Prior to collecting kinematic and kinetic data, the Vicon Bonita Motion Analysis system was statically and dynamically calibrated. On calibration, the origin was set to allow the software to identify the location of the cameras with respect to the three-dimensional axes X (anterior/ posterior), Y (medial/lateral) and Z (vertical). Prior to testing, the AMTI force plates were reset. Participants age (years) and anthropometric measurements such as height (cm), and body mass (kg) were recorded. Body mass and height were recorded using calibrated clinical scales (Seca 803, Seca GmbH, Hamburg, Germany) and a stadiometer (Seca 213, Seca GmbH, Hamburg, Germany), respectively. The participants were asked which leg was dominant and was defined as their preferred leg when kicking a ball. All participants were dominant on their right side.

Labelling Procedure

Twenty-six retroreflective markers (3 mm) were attached to the left and right heel (thirteen markers on each heel) using double-sided toupee tape (30m, Loughborough, UK) which was cut into 2 mm individual squares. Participants were asked to stand

barefoot with their weight distributed equally on both feet while a custom designed template was placed on the left and right heels. A fine and ultra-fine Sharpie® permanent marker (Newell Brands, Atlanta, Georgia, USA) was used to mark the location for the 3 mm retroreflective markers as double-sided tape was transferred to the marked areas on the skin. Retroreflective markers were placed at three levels; upper layer, middle layer and lower layer. Seven markers were placed along the lower circumference of the HP, three markers were positioned on the middle layer and three markers were attached to the upper layer of the heel. Only three markers from each heel (L1, L3 and L7) were assessed in this study (Figure 1).

Following the retroreflective markers being securely attached to the heel, participants were asked to stand on the force plates with their heels facing the cameras and hands on their hips. The left foot was positioned in the centre of Force Plate 1 and the right foot was positioned in the centre of Force Plate 2, with both feet shoulder width apart. Each participant was given a 10-min familiarisation period to practice moving into a plantarflexed position (two-footed heel raise) at a controlled speed. The entire movement comprised of three continuous phases: foot flat (baseline phase), plantarflexion (unloading phase) and foot flat (loading phase). This required participants to stand still then slowly unload both heels by moving in to a two-footed heel raise (plantarflexion). Participants then loaded the HP by slowly placing both heels back on the ground. Each phase lasted two seconds and data were recorded by a researcher. Three trials were recorded.

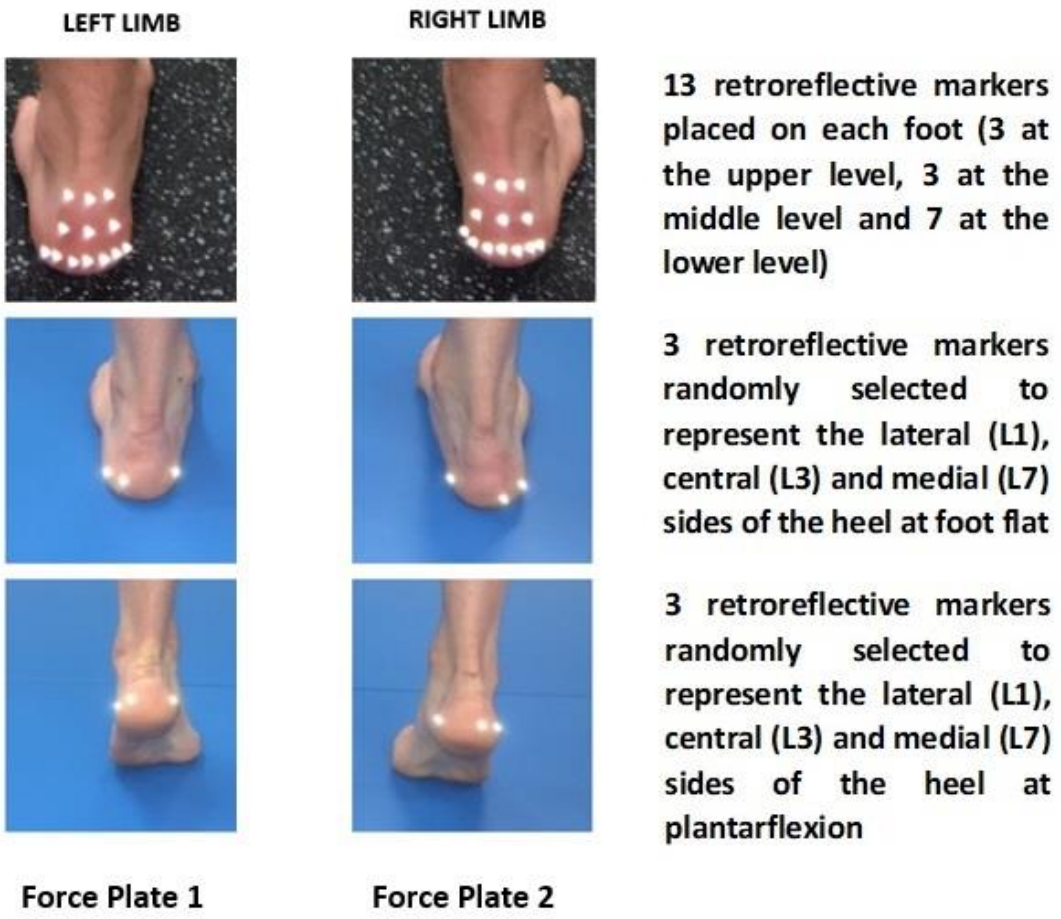

Figure 1: Retroreflective marker configuration during dynamic standing heel raise task.

Data Analysis

Kinematic data were transferred as an ASCII file and then imported to Microsoft Excel 2017 version 16.10 (Microsoft Corporation, Redmond, Washington) for analysis. The displacement of the heel was evaluated based on the motion of the HP during dynamic activity with respect to each phase; baseline (phase 1), unloading (phase 2) and loading (phase 3). Change in displacement (Δ Displacement) between phases was reported. Displacement of the unloaded HP was analysed from phase 1 to phase 2. Displacement of the loaded HP was evaluated from phase 2 to phase 3 and overall total displacement was assessed from phase 1 to phase 3. Displacement (***d***) was calculated using the following equation:

$$\boldsymbol{d} = \sqrt{((x_2 - x_1)^2 + (y_2 - y_1)^2 + (z_2 - z_1)^2)} \quad (1)$$

Where x_1, y_1, z_1 are the coordinates of the first marker position and x_2, y_2, z_2 are the coordinates of the second marker position.

Other outputs reported were expressed as derivatives from the captured biomechanical data. The markers located at L1, L3 and L7 were joined to form a scalene triangle. Using the displacement formula, the lengths ((L3 – L1), (L7 – L3) and (L7 – L1)) were calculated using the *x*, *y* and *z* marker orientations. Morphometric and morphological datasets were calculated as the perimeter and cross-sectional area (CSA) respectively. The perimeter (***p***) was calculated as

$$\boldsymbol{p} = (L3 - L1) + (L7 - L3) + (L7 - L1) \quad (2)$$

While the CSA (***a***) was calculated as

$$\boldsymbol{a} = \sqrt{\left(\boldsymbol{s}(\boldsymbol{s} - (L3 - L1))(\boldsymbol{s} - (L7 - L3))(\boldsymbol{s} - (L7 - L1)\right)} \quad (3)$$

Where $\boldsymbol{s} = \frac{(L3-L1)+(L7-L3)+(L7-L1)}{2}$ (4)

Compliance measurements at the medial and lateral sides respectively were expressed as a ratio of the medial and lateral displacements to the resultant force. Medial and lateral strains were calculated and the medial-lateral strain ratio was determined. Multivariate statistical analysis techniques were applied to the dataset. The independent variables were gender, sidedness and phases. The dependent variables were perimeter, CSA, lateral displacement, medial displacement, lateral compliance, medial compliance, lateral strain,

medial strain, and medial-lateral strain ratio. Bonferroni post-hoc multiple comparisons for observed means were applied as the estimates of the effect sizes expressed as partial eta squared statistic (η_p^2) were determined. The values of 0.0099, 0.0588, and 0.1379 were considered *small, medium,* and *large* effect sizes respectively[17]. Significance was established at $p < 0.05$.

RESULTS

Heel Pad Biomechanical Output Measures

The datasets for the morphometric and morphological output measures, and compliance and strain output measures are displayed in Table 1 and Table 2 respectively. The independent variable, gender, only showed significant differences for lateral compliance ($F = 6.527$, $p = 0.014$, $\eta_p^2 = 0.120$, *medium*), and medial compliance ($F = 13.489$, $p = 0.001$, $\eta_p^2 = 0.219$, *large*). No significant differences with respect to gender were found for perimeter ($F = 3.545$, $p = 0.066$, $\eta_p^2 = 0.069$, *medium*), CSA (F = 0.156, $p = 0.695$, $\eta_p^2 = 0.003$, *small*), lateral displacement (F = 0.602, $p = 0.442$, $\eta_p^2 = 0.012$, small), medial displacement (F = 0.984, $p = 0.326$, $\eta_p^2 = 0.020$, *small*), lateral strain ($F = 0.142$, $p = 0.709$, $\eta_p^2 = 0.004$, *small*), medial strain ($F = 0.144$, $p = 0.706$, $\eta_p^2 = 0.004$, *small*), and medial-lateral strain ratio ($F = 1.421$, $p = 0.242$, $\eta_p^2 = 0.043$, *small*).

Except for gender, no significant main effect ($p > 0.05$) was observed after following a multivariate statistical test. In particular, with respect to the following dependent variables (perimeter, CSA, lateral displacement, medial displacement, lateral compliance and medial compliance) the multivariate test revealed no significant differences for sidedness ($F = 0.691$, $p = 0.658$, $\eta_p^2 = 0.088$, *medium*), phases ($F = 0.580$, $p = 0.853$, η_p^2

= 0.073, *medium*), combined gender and sidedness effects (F = 2.005, p = 0.086, η_p^2 = 0.219, *large*), combined gender and phases effects (F = 0.266, p = 0.993, η_p^2 = 0.035, *small*), combined sidedness and phases effects (F = 0.589, p = 0.846, η_p^2 = 0.074, *medium*) and combined gender, sidedness and phases effects (F = 0.506, p = 0.906, η_p^2 = 0.064, *medium*).

Further multivariate tests using the dependent variables lateral strain, medial strain and medial-lateral strain ratio also showed no significant main effects (p > 0.05) for gender (F = 1.641, p = 0.201, η_p^2 = 0.141, *large*), sidedness (F = 1.312, p = 0.289, η_p^2 = 0.116, *medium*), strain (F = 2.859, p = 0.053, η_p^2 = 0.222, *large*), combined gender and sidedness effects (F = 1.489, p = 0.237, η_p^2 = 0.130, *large*), combined gender and strain (F = 0.247, p = 0.863, η_p^2 = 0.024, *small*), and combined sidedness and strain effects (F = 0.894, p = 0.456, η_p^2 = 0.082, *medium*). Significant main effects were observed for the combined gender, sidedness and strain effects (F = 4.670, p = 0.009, η_p^2 = 0.318, *large*).

Table 1. Heel Pad Morphometric and Morphological Output Measurements.

Heel Positional Phase		Biomechanical Output Measure	All Participants	Female Participants	Male Participants
			Mean (SD)	Mean (SD)	Mean (SD)
	Baseline	Left Heel Pad Morphometric Measurement (Perimeter in mm)	109.96 (9.24)	105.52 (10.55)	114.40 (5.60)
		Left Heel Pad Morphological Measurement (Cross Sectional Area in mm^2)	401.17 (93.98)	370.86 (117.60)	431.49 (61.20)
		Right Heel Pad Morphometric Measurement (Perimeter in mm)	109.29 (7.18)	109.30 (6.95)	109.27 (8.22)
		Right Heel Pad Morphological Measurement (Cross Sectional Area in mm^2)	387.64 (48.74)	403.72 (59.28)	371.56 (34.43)
	Unloading	Left Heel Pad Morphometric Measurement (Perimeter in mm)	106.90 (8.24)	105.34 (11.06)	108.47 (4.94)
		Left Heel Pad Morphological Measurement (Cross Sectional Area in mm^2)	370.86 (83.67)	367.21 (116.96)	374.51 (45.14)
		Right Heel Pad Morphometric Measurement (Perimeter in mm)	106.55 (5.49)	105.24 (5.55)	107.87 (5.71)
		Right Heel Pad Morphological Measurement (Cross Sectional Area in mm^2)	382.13 (52.45)	379.56 (58.53)	384.69 (52.42)
	Loading	Left Heel Pad Morphometric Measurement (Perimeter in mm)	110.54 (8.94)	106.14 (10.59)	114.95 (4.37)
		Left Heel Pad Morphological Measurement (Cross Sectional Area in mm^2)	392.36 (91.68)	377.00 (116.43)	407.71 (69.02)
		Right Heel Pad Morphometric Measurement (Perimeter in mm)	110.95 (6.85)	111.48 (5.38)	110.41 (8.71)
		Right Heel Pad Morphological Measurement (Cross Sectional Area in mm^2)	392.55 (64.02)	404.79 (65.28)	380.32 (67.71)

Table 2. Heel Pad Compliance and Strain Output Measurements.

Heel Positional Phase		Biomechanical Output Measure	All Participants	Female Participants	Male Participants
			Mean (SD)	Mean (SD)	Mean (SD)
	Baseline	Left L3-L1 (Lateral Compliance Measurement (mm/N))	0.07 (0.02)	0.07 (0.03)	0.06 (0.01)
		Left L3-L7 (Medial Compliance Measurement (mm/N))	0.12 (0.03)	0.13 (0.03)	0.11 (0.02)
		Right L3-L1 (Lateral Compliance Measurement (mm/N))	0.07 (0.01)	0.07 (0.01)	0.06 (0.01)
		Right L3-L7 (Medial Compliance Measurement (mm/N))	0.12 (0.02)	0.13 (0.02)	0.12 (0.01)
	Unloading	Left L3-L1 (Lateral Compliance Measurement (mm/N))	0.07 (0.02)	0.07 (0.03)	0.06 (0.01)
		Left L3-L7 (Medial Compliance Measurement (mm/N))	0.12 (0.03)	0.14 (0.03)	0.10 (0.02)
		Right L3-L1 (Lateral Compliance Measurement (mm/N))	0.07 (0.01)	0.07 (0.01)	0.06 (0.01)
		Right L3-L7 (Medial Compliance Measurement (mm/N))	0.12 (0.02)	0.12 (0.01)	0.12 (0.02)
	Loading	Left L3-L1 (Lateral Compliance Measurement (mm/N))	0.07 (0.02)	0.07 (0.03)	0.06 (0.01)
		Left L3-L7 (Medial Compliance Measurement (mm/N))	0.12 (0.03)	0.13 (0.03)	0.10 (0.01)
		Right L3-L1 (Lateral Compliance Measurement (mm/N))	0.07 (0.02)	0.08 (0.02)	0.07 (0.02)
		Right L3-L7 (Medial Compliance Measurement (mm/N))	0.13 (0.03)	0.14 (0.02)	0.12 (0.03)
	Unloading	Left L3-L1 (Lateral Strain)	-0.02 (0.02)	-0.02 (0.02)	-0.03 (0.03)
		Left L3-L7 (Medial Strain)	-0.04 (0.07)	0.00 (0.02)	-0.08 (0.08)
		Right L3-L1 (Lateral Strain)	-0.01 (0.04)	-0.01 (0.02)	0.00 (0.05)
		Right L3-L7 (Medial Strain)	-0.02 (0.08)	-0.05 (0.03)	0.01 (0.11)
		Left Medial : Lateral Ratio	1.16 (3.66)	1.32 (2.40)	1.00 (4.93)
		Right Medial : Lateral Ratio	-0.27 (4.02)	0.65 (5.28)	-1.20 (2.53)
	Loading	Left L3-L1 (Lateral Strain)	0.01 (0.05)	0.01 (0.02)	0.01 (0.07)
		Left L3-L7 (Medial Strain)	-0.01 (0.04)	0.01 (0.02)	-0.02 (0.05)
		Right L3-L1 (Lateral Strain)	0.02 (0.06)	0.03 (0.07)	0.01 (0.05)
		Right L3-L7 (Medial Strain)	0.01 (0.10)	0.00 (0.02)	0.01 (0.15)
		Left Medial : Lateral Ratio	0.58 (3.22)	-1.10 (2.15)	2.27 (3.40)
		Right Medial : Lateral Ratio	-0.03 (0.96)	0.15 (0.82)	-0.34 (1.27)

Negative strains suggest compressive strains. Positive strains suggest tensile strains.

Heel Pad Displacement Between Genders

During the unloading phase, there was no significant differences between males and females in the left heel for markers L1 ($p = 0.591$), L3 ($p = 0.448$) and L7 ($p = 0.589$) (Figure 2a). Similarly, there was no significant differences between genders in the right heel for markers L1 ($p = 0.888$), L3 ($p = 0.755$) and L7 ($p = 0.814$) (Figure 2a). Within the loading phase, there was no significant differences between genders in left HP displacement for markers L1 ($p = 0.585$), L3 ($p = 0.495$), L7 ($p = 0.553$) and right HP displacement for markers L1 ($p = 0.977$), L3 ($p = 0.785$), L7 ($p = 0.805$) (Figure 2b). There were no significant differences in overall left HP displacement for markers L1 ($p = 0.167$), L3 ($p = 0.154$) and L7 ($p = 0.177$) from the baseline to loading phase (Figure 2c). Likewise, there were no significant differences in overall right HP displacement for markers L1 ($p = 0.245$) and L3 ($p = 0.282$) (Figure 2c). However, a significant difference in overall right HP displacement occurred in marker L7 ($p = 0.034$) from the baseline to loading phase.

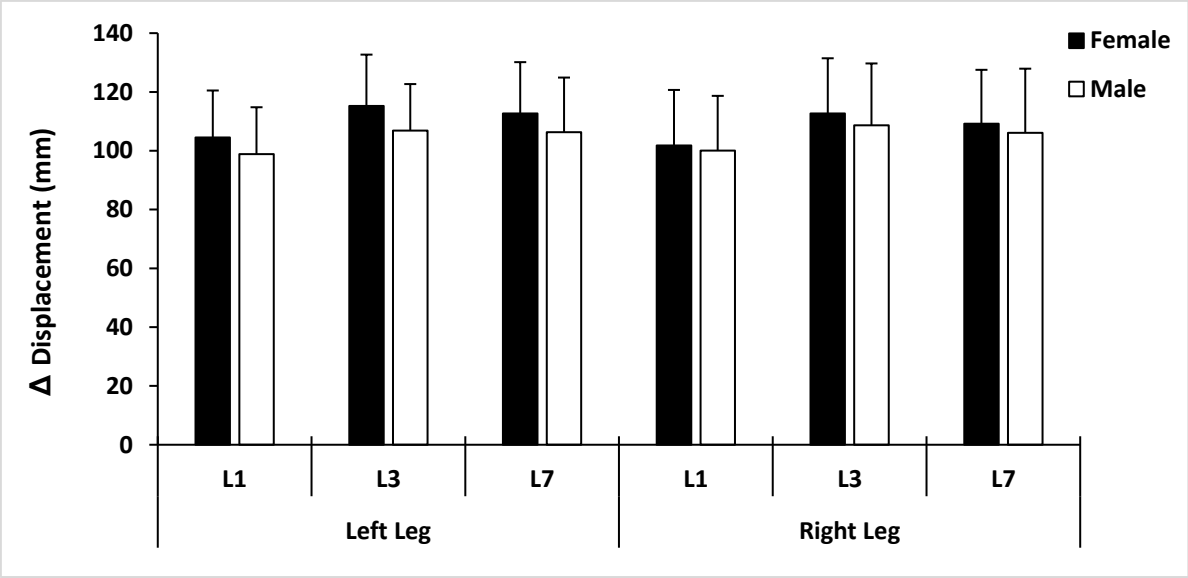

Figure 2a: Mean gender differences in the displacement of the lower heel at each marker location (L1, L3 and L7) during the unloading phase with error bars (± standard deviation) (n=10).

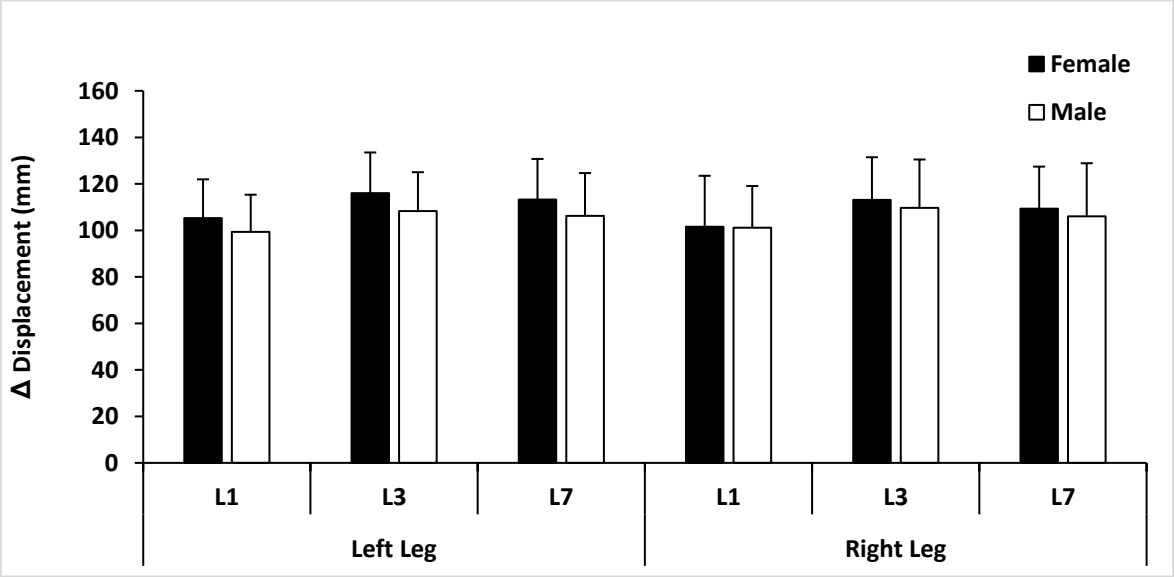

Figure 2b: Mean gender differences in the displacement of the lower heel at each marker location (L1, L3 and L7) during the loading phase with error bars (± standard deviation) (n=10).

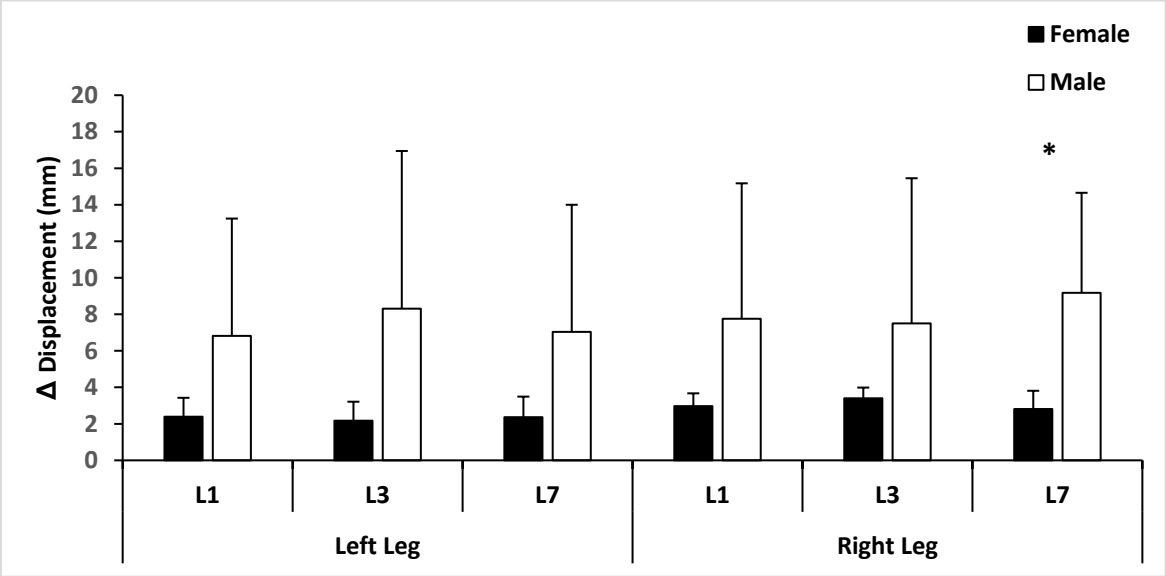

Figure 2c: Mean gender differences in the displacement of the lower heel at each marker location (L1, L3 and L7) from the baseline to loading phase with error bars (± standard deviation) (n=10). * indicates statistical significance ($p < 0.05$).

Displacement between the Dominant and Non-Dominant Heels

When the heel was unloaded, there were no significant differences for markers L1 (p = 0.812), L3 (p = 0.829) and L7 (p = 0.764) when comparing the dominant heel (right) to the non-dominant heel (left) in females (Figure 3a). Likewise, there were no significant differences for males between the unloaded dominant heel (right) and unloaded non-dominant heel (left) in markers L1 (p = 0.915), L3 (p = 0.884) and L7 (p = 0.987) (Figure 3a). Within the loading phase, there were no significant differences when comparing the dominant heel (right) against the non-dominant heel (left) for markers L1 (p = 0.780), L3 (p = 0.804) and L7 (p = 0.736) in females (Figure 3b). Similarly, no significant differences were found in males between the loaded dominant heel (right) and loaded non-dominant heel (left) for markers L1 (p = 0.874), L3 (p = 0.914) and L7 (p = 0.987) (Figure 3b).

There were no significant differences found for overall heel displacement during the baseline to loading phase when comparing females dominant (right) and non-dominant (lefts) heels in markers L1 (p = 0.372), L3 (p = 0.052) and L7 (p = 0.527) (Figure 3c). Also, no significant differences were found within the baseline to loading phase when comparing the males dominant and non-dominant legs in markers L1 (p = 0.835), L3 (p = 0.883) and L7 (p = 0.604) (Figure 3c).

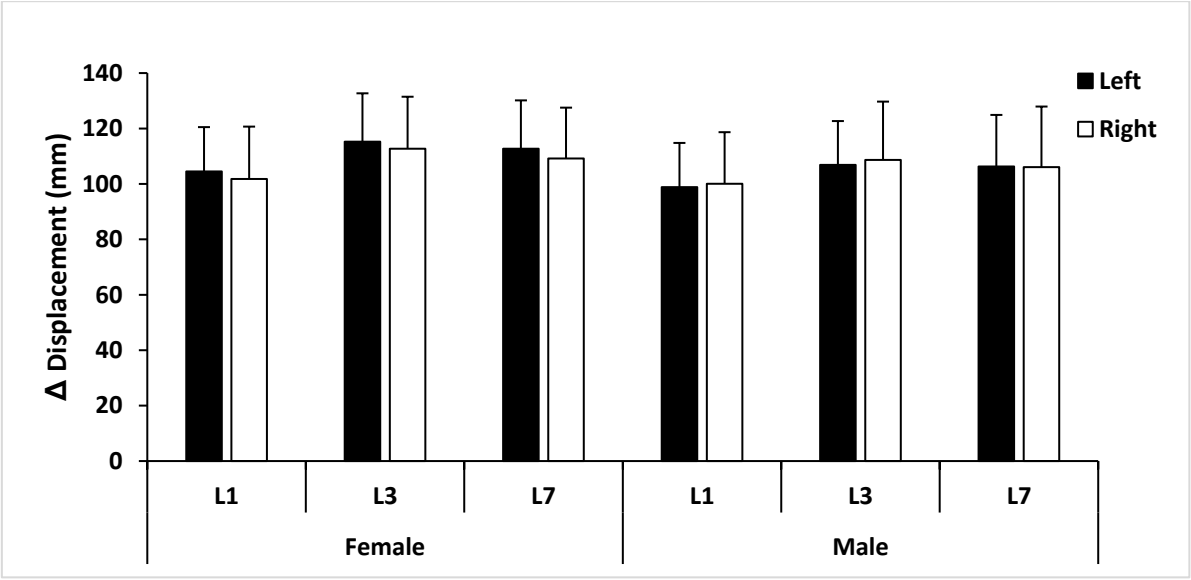

Figure 3a: Mean differences in lower heel displacement each marker location (L1, L3 and L7) between the dominant and non-dominant leg during the unloading phase at with error bars (± standard deviation) (n=10).

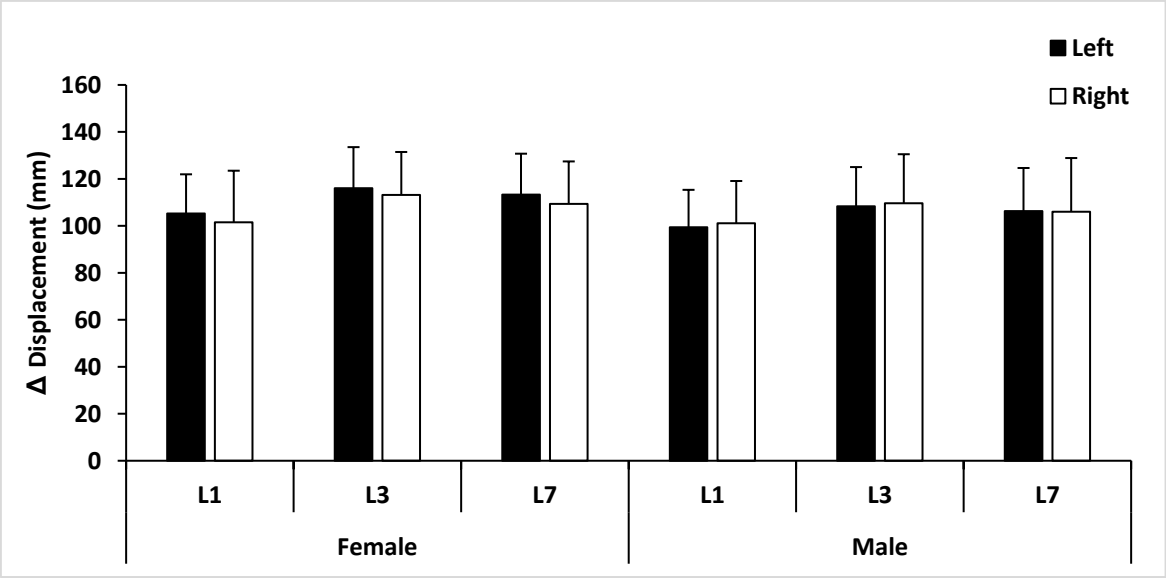

Figure 3b: Mean differences in lower heel displacement at each marker location (L1, L3 and L7) between the dominant and non-dominant leg during the loading phase with error bars (± standard deviation) (n=10).

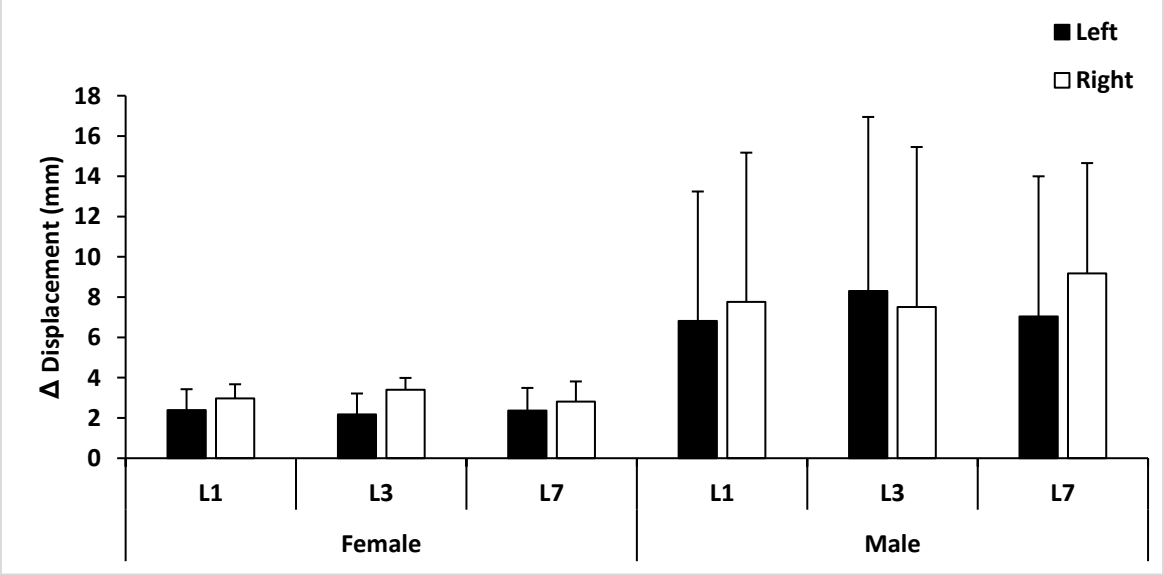

Figure 3c: Mean differences in lower heel displacement at each marker location (L1, L3 and L7) between the dominant and non-dominant leg from the baseline to the loading phase with error bars (± standard deviation) (n=10).

Gender Differences in X, Y and Z Displacement Planes

Unloading Phase

There were no significant differences between genders in the left heel on the anterior/ posterior (X) plane in markers L1 (p = 0.146), L3 (p = 0.214) or L7 (p = 0.119) (Figure 4). Similarly, there were no significant differences between males and females on the anterior/posterior (X) plane within the right heel for markers L1 (p = 0.285), L3 (p = 0.228) or L7 (p = 0.207) (Figure 4). Also, there were no significant differences in the medial/lateral (Y) plane in the left heel for markers L1 (p = 0.213), L3 (p = 0.258) and L7 (p = 0.213) or the right heel, L1 (p = 0.422), L3 (p = 0.326) and L7 (p = 0.477) (Figure 4). Likewise, no significant differences were found in the left heel, L1 (p = 0.773), L3 (p = 0.505) and L7 (p = 0.759), or the right heel, L1 (p = 0.875), L3 (p = 0.952) and L7 (p = 0.919), on the vertical plane (Z) (Figure 4).

Loading Phase

There were no significant differences when comparing males and females in the left heel on the anterior/posterior (X) plane in markers L1 (p = 0.219), L3 (p = 0.188) and L7 (p = 0.101) (Figure 5). In addition, there were no significant differences on the anterior/ posterior (X) plane for the right heel in markers L1 (p = 0.349), L3 (p = 0.236) and L7 (p = 0.199) (Figure 5). Also, there were no significant differences on the medial/lateral (Y) plane in the left heel for markers L1 (p = 0.234), L3 (p = 0.180) and L7 (p = 0.172) or the right heel, L1 (p = 0.543), L3 (p = 0.193) and L7 (p = 0.197) (Figure 5). Similarly, no significant differences were found in the left heel, L1 (p = 0.680), L3 (p = 0.581) and L7 (p = 0.676), or the right heel, L1 (p = 0.547), L3 (p = 0.943) or L7 (p = 0.955), on the vertical plane (Z) (Figure 5).

Overall Displacement

There were no significant differences between genders in the left heel on the anterior/ posterior (X) plane in markers L1 (p = 0.088), L3 (p = 0.850) and L7 (p = 0.528). Similarly, within the anterior/posterior (X) plane there were no significant differences between genders in the right heel for markers L1 (p = 0.323), L3 (p = 0.381) and L7 (p = 0.909). Also, there were no significant differences on the medial/lateral (Y) plane in the left heel for markers L1 (p = 0.965), L3 (p = 0.842) and L7 (p = 0.883) or the right heel, L1 (p = 0.419), L3 (p = 0.528) and L7 (p = 0.343). Likewise, on the vertical plane (Z) there were no significant differences between male and females within the left heel, L1 (p = 0.395), L3 (p = 0.229) and L7 (p = 0.163), or the right heel, L1 (p = 0.377), L3 (p = 0.896) and L7 (p = 0.675).

Regarding the L1, L3 and L7 markers, no significant differences between the unloading and loading phases (p > 0.367) were observed, for both males and females with respect to the dominant and non-dominant limbs.

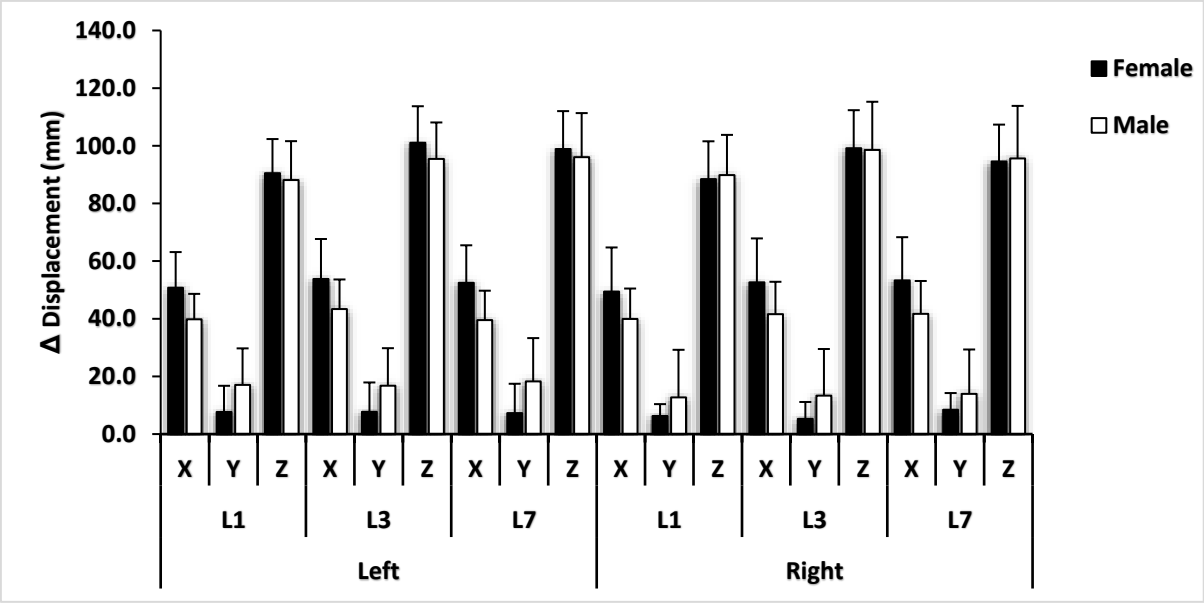

Figure 4: Mean gender differences in lower heel displacement on the vertical (Z), anterior/posterior (X) and medial/lateral (Y) planes at each marker location (L1, L3 and L7) during the unloading phase with error bars (± standard deviation) (n=10).

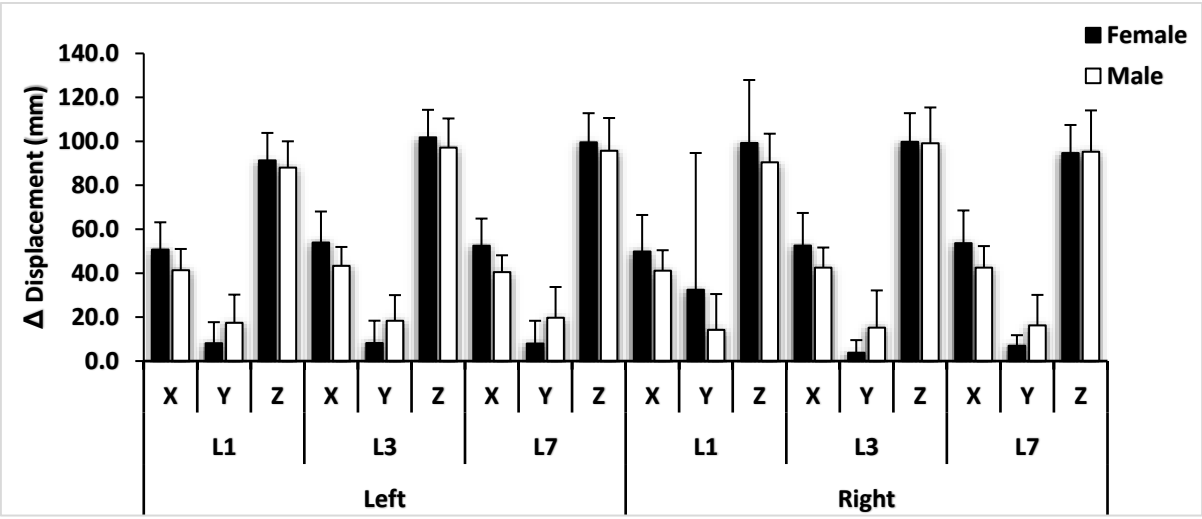

Figure 5: Mean gender differences in lower heel displacement on the vertical (Z), anterior/posterior (X) and medial/lateral (Y) planes at each marker location (L1, L3 and L7) during the loading phase with error bars (± standard deviation) (n=10).

DISCUSSION

The present study aimed to develop biomechanical techniques to evaluate the function and characteristics of the HP deformation using 3D motion analysis and force plate technology. These techniques were used as key experimental drivers to explain changes in soft tissue properties from a HP marker displacement and deformation perspective. Loading and unloading conditions during foot plantarflexion were also assessed using outcome measures representative of the morphometry, morphology and biomechanical responses of the HP. Gender differences together with how the dominant versus non-dominant heel influences HP displacement during foot plantarflexion (replicating the toe-off phase of gait) were also investigated. Although the independent variable (i.e. gender) only showed significant differences in medial and lateral compliance ($p < 0.05$), all other biomechanical outputs showed no significant differences between gender. While previous results have not reported compliance outputs our results showed that all males produced lower HP compliance outputs in comparison to females. The interpretation of these results suggest that males have a stiffer HP in comparison to females. Even though our methodology was different, our results agree with earlier studies by Matteoli et al. 2012 and Tas et al. 2018. Overall, during the unloading phase both the left and right limbs underwent compressive strains on the lateral and medial borders of the HP. Apart from the medial strain on the left HP, the left lateral strain, right lateral strain and right medial strain all exhibited tensile strain patterns during the loading phase. Holistically, the medial-lateral strain ratio from both limbs were similar but dissimilar between genders. These outcomes may account for the structural complexity of the HP and variability in the viscoelasticity of the HP soft tissues.

The HP displacement results between genders indicated that females had a greater mean HP displacement compared to males across all markers (L1, L3 and L7) for both heels during the unloading and loading phases. Additionally, this study highlighted that overall HP displacement was significantly higher in males on the most lateral part of the right heel from the baseline to loading phase. Likewise, a study by Alcántara et al. (2002) demonstrated that females had a lower peak displacement compared to males during a ballistic pendulum test[18]. Furthermore, similar results were illustrated by Jørgensen and Bojsen-Møller (1989) who found that peak HP displacement was reduced in females[19]. However, it is worth considering that the study by Jørgensen and Bojsen-Møller (1989) analysed the HP using an impact velocity test which gradually applied force to the HP[19]. Research suggests that examining the HP externally from the actual / realistic loading condition can yield meaningful information regarding the deformation of the viscoelastic properties which can influence the reliability and validity of the results[20]. Furthermore, studies evaluating the gender differences did not control for the influence of hormones in the female and male subjects[18, 19]. The thicker HP in males has been associated with higher concentrations of growth hormone[21, 22]. Whereas, a greater concentration of oestrogen in females at certain time points might alter the elasticity and increase the laxity of the structures surrounding the fibro-adipose tissue[21]. Subsequently, this may result in females exhibiting an increased elasticity and greater displacement within the HP due to the visco-elastic tissue being softer in correspondence with hormonal influence. This could be the reason why females in the present study had a greater displacement compared to males.

With limited studies that have investigated the displacement of the HP, our output measures showed that the deformation of the lower markers was small and there were no significant differences in HP displacement between genders in the vertical (Z), anterior/posterior (X) and medial/lateral (Y) axes of measurements. These results are similar to research by Hsu et al. (1998) that showed when a load of 3kg was placed on the bottom of the HP there was a displacement greater than 0.8 cm (true value not stated within the study)[4]. In their study, the maximum load utilised was 3 kg whereas this present study loaded the HP by using the natural body weight of the participants. Previous research has suggested that *in vitro* analysis of the HP typically demonstrated higher values of HP deformation under the stimulation of body-weight[13]. In comparison, in *vivo* examination of the HP by Kinoshita et al. (1996) reported lower values of HP deformation when the heel was artificially loaded[23]. Therefore, examining the HP using motion analysis may result in larger HP displacement due to the participants' body weight having a greater effect on the soft tissue within the heel.

In terms of HP displacement between the dominant and non-dominant heels, there were no significant differences in HP displacement when comparing the non-dominant and dominant heels within all three markers (L1, L3 and L7) during the baseline, unloading and loading phases. To date, no research has analysed HP displacement by comparing the dominant and non-dominant heels. Flanagan and Harrison (2007) suggested that both lower limbs are utilized and trained equally through walking and running, however, for activities (e.g., tennis, badminton, baseball) that require one limb to be active the dominant leg tends to become superior in the performance of the task which can result in displacement being reduced[10]. Studies have indicated that greater lateral centre of

pressure (COP) displacement may increase the likelihood of overuse injury such as patellofemoral pain[24]. Despite this, more research in this area is required to establish if HP displacement is associated with similar overuse injuries.

This present study has some limitations. Firstly, the timing of the menstrual cycle was not considered in the female participants in relation to hormone concentrations. Future research should consider and control for this variable. Secondly, a small data set of only ten participants was used to analyse HP displacement. Nevertheless, the 10 participants were healthy and are therefore representative of the general population. With a larger sample size it is envisaged that symmetry and asymmetry medial-lateral strain ratio patterns may be extracted which potentially may provide an insight into potential HP soft tissue balances and/or imbalances experienced during stance and/or during loading and unloading conditions. Thirdly, the age of the participants varied largely (16-54 years). Several studies have examined HP properties across a range of age groups. Despite the use of different equipment, the majority of literature has consistently reported that HP biomechanics are altered by age[25, 26]. Future research should investigate if HP displacement is influenced by different age groups. Lastly, foot posture (supination/pronation), range of motion (ROM) of the ankle joint and scaling for body mass were not accounted for when analysing HP displacement. These factors could have affected the results by changing the mechanical responses of the HP tissues during loading. Although only six out of twenty-six retroreflective markers were examined, further research is underway which examines all makers in the upper, middle and lower regions of the HP.

CONCLUSION

A biomechanical technique using very retroreflective markers that are representative of the HP circumference has been developed in this study. The results from this study highlighted that overall HP displacement was significantly higher in males compared to females on the most lateral area of the right heel from the baseline to loading phase. Nonetheless, there was no significant change in HP displacement when comparing the dominant and non-dominant heels at baseline, unloading and loading conditions. Future studies should evaluate deformation differences between a two-footed and one-footed heel raise task, which are the most typical postures adopted in sports. Lastly, the use of three-dimensional motion analysis can allow the examination of HP deformation / displacements in three directions to understand tissue mechanics during dynamic movements.

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REFERENCES

1. Declercq D, Aerts P and Kunnen M. The Mechanical Characteristics of the Human Heel Pad during Foot Strike in Running - an in-Vivo Cineradiographic Study. *Journal of Biomechanics* 1994; 27: 1213-&. DOI: Doi 10.1016/0021-9290(94)90275-5.
2. Teoh JC and Lee T. Prediction of plantar soft tissue stiffness based on sex, age, bodyweight, height and body mass index. *J Mech Behav Biomed Mater* 2016; 54: 219-222. 2015/10/17. DOI: 10.1016/j.jmbbm.2015.09.015.

3. Rome K, Webb P, Unsworth A, et al. Heel pad stiffness in runners with plantar heel pain. *Clin Biomech (Bristol, Avon)* 2001; 16: 901-905. 2001/12/06. DOI: 10.1016/s0268-0033(01)00081-x.
4. Hsu TC, Wang CL, Tsai WC, et al. Comparison of the mechanical properties of the heel pad between young and elderly adults. *Arch Phys Med Rehabil* 1998; 79: 1101-1104. 1998/09/28. DOI: 10.1016/s0003-9993(98)90178-2.
5. Matteoli S, Madsen M, Virga A, et al. In Vivo Heel Pad Elasticity Investigation: Comparing Males to Females. *In: International Tissue Elasticity Conference*. 2012.
6. Tas S. Effect of Gender on Mechanical Properties of the Plantar Fascia and Heel Fat Pad. *Foot Ankle Spec* 2018; 11: 403-409. 2017/10/17. DOI: 10.1177/1938640017735891.
7. Ozdemir H, Soyuncu Y, Ozgorgen M, et al. Effects of changes in heel fat pad thickness and elasticity on heel pain. *J Am Podiatr Med Assoc* 2004; 94: 47-52. 2004/01/20. DOI: 10.7547/87507315-94-1-47.
8. Irving DB, Cook JL, Young MA, et al. Impact of chronic plantar heel pain on health-related quality of life. *Journal of the American Podiatric Medical Association* 2008; 98: 283-289.
9. Tsai W-C, Chiu M-F, Wang C-L, et al. Ultrasound evaluation of plantar fasciitis. *Scandinavian journal of rheumatology* 2000; 29: 255-259.
10. Flanagan EP and Harrison AJ. Muscle dynamics differences between legs in healthy adults. *Journal of Strength and Conditioning Research* 2007; 21: 67.
11. Kamonseki DH, Gonçalves GA, Liu CY, et al. Effect of stretching with and without muscle strengthening exercises for the foot and hip in patients with plantar fasciitis: a randomized controlled single-blind clinical trial. *Manual therapy* 2016; 23: 76-82.
12. Buschmann WR, Hudgins LC, Kummer F, et al. Fatty acid composition of normal and atrophied heel fat pad. *Foot & ankle* 1993; 14: 389-394.
13. Ker RF. The time-dependent mechanical properties of the human heel pad in the context of locomotion. *J Exp Biol* 1996; 199: 1501-1508. 1996/07/01.
14. Mickle KJ, Steele JR and Munro BJ. Is the foot structure of preschool children moderated by gender? *J Pediatr Orthop* 2008; 28: 593-596. 2008/06/27. DOI: 10.1097/BPO.0b013e318173f782.
15. Chi KJ and Schmitt D. Mechanical energy and effective foot mass during impact loading of walking and running. *Journal of Biomechanics* 2005; 38: 1387-1395. DOI: 10.1016/j.jbiomech.2004.06.020.
16. Aerts P, Ker RF, De Clercq D, et al. The mechanical properties of the human heel pad: a paradox resolved. *J Biomech* 1995; 28: 1299-1308. 1995/11/01. DOI: 10.1016/0021-9290(95)00009-7.
17. Richardson JT. Eta squared and partial eta squared as measures of effect size in educational research. *Educational Research Review* 2011; 6: 135-147.
18. Alcantara E, Forner A, Ferrus E, et al. Influence of age, gender, and obesity on the mechanical properties of the heel pad under walking impact conditions. *J Appl Biomech* 2002; 18: 345-356. DOI: DOI 10.1123/jab.18.4.345.
19. Jørgensen U and Bojsen-Møller F. Shock absorbency of factors in the shoe/heel interaction—with special focus on role of the heel pad. *Foot & Ankle* 1989; 9: 294-299.

20. Spears IR and Miller-Young JE. The effect of heel-pad thickness and loading protocol on measured heel-pad stiffness and a standardized protocol for inter-subject comparability. *Clinical Biomechanics* 2006; 21: 204-212.
21. Rome K. Mechanical properties of the heel pad: current theory and review of the literature. *Foot* 1998; 8: 179-185.
22. Weltman A, Weltman J, Hartman M, et al. Relationship between age, percentage body fat, fitness, and 24-hour growth hormone release in healthy young adults: effects of gender. *The Journal of Clinical Endocrinology & Metabolism* 1994; 78: 543-548.
23. Kinoshita H, Francis PR, Murase T, et al. The mechanical properties of the heel pad in elderly adults. *Eur J Appl Physiol Occup Physiol* 1996; 73: 404-409. 1996/01/01. DOI: 10.1007/bf00334416.
24. Dowling GJ, Murley GS, Munteanu SE, et al. Dynamic foot function as a risk factor for lower limb overuse injury: a systematic review. *Journal of foot and ankle research* 2014; 7: 53.
25. Gefen A, Megido-Ravid M and Itzchak Y. In vivo biomechanical behavior of the human heel pad during the stance phase of gait. *Journal of biomechanics* 2001; 34: 1661-1665.
26. Kwan RL, Zheng YP and Cheing GL. The effect of aging on the biomechanical properties of plantar soft tissues. *Clin Biomech (Bristol, Avon)* 2010; 25: 601-605. 2010/05/12. DOI: 10.1016/j.clinbiomech.2010.04.003.