Multi-segment foot kinematics and ground reaction forces during gait of individuals with plantar fasciitis

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A B S T R A C T

Background: Clinically, plantar fasciitis (PF) is believed to be a result and/or prolonged by overpronation and excessive loading, but there is little biomechanical data to support this assertion. The purpose of this study was to determine the differences between healthy individuals and those with PF in (1) rearfoot motion, (2) medial forefoot motion, (3) first metatarsal phalangeal joint (FMPJ) motion, and (4) ground reaction forces (GRF).

Methods: We recruited healthy (n=22) and chronic PF individuals (n=22, symptomatic over three months) of similar age, height, weight, and foot shape (p > 0.05). Retro-reflective skin markers were fixed according to a multi-segment foot and shank model. Ground reaction forces and three dimensional kinematics of the shank, rearfoot, medial forefoot, and hallux segment were captured as individuals walked at 1.35 m s⁻¹.

Results: Despite similarities in foot anthropometrics, when compared to healthy individuals, individuals with PF exhibited significantly (p < 0.05) (1) greater total rearfoot eversion, (2) greater forefoot plantar flexion at initial contact, (3) greater total sagittal plane forefoot motion, (4) greater maximum FMPJ dorsiflexion, and (5) decreased vertical GRF during propulsion.

Conclusion: These data suggest that compared to healthy individuals, individuals with PF exhibit significant differences in foot kinematics and kinetics. Consistent with the theoretical injury mechanisms of PF, we found these individuals to have greater total rearfoot eversion and peak FMPJ dorsiflexion, which may put undue loads on the plantar fascia. Meanwhile, increased medial forefoot plantar flexion at initial contact and decreased propulsive GRF are suggestive of compensatory responses, perhaps to manage pain.

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1. Introduction

Plantar fasciitis (PF) is the most common cause of heel pain, yet its aetiology is not well understood (Young et al., 2001). Typically the prognosis of a conservative treatment plan is good, but approximately 10% of cases are recalcitrant (Davis et al., 1994). Numerous factors are thought to contribute to the development of PF; however, biomechanical factors are considered to be the principal contributors (Wearing et al., 2006). Clinicians believe that excessive strain and loading of the plantar fascia (also known as the plantar aponeurosis) occurs concurrently with abnormal subtalar joint overpronation, and flattening of the medial longitudinal arch (often clinically referred to as pes planus, or flat foot) (Kwong et al., 1988; Subotnick, 1981; Taunton et al., 1982). In addition, high ground reaction forces (GRF) during locomotion could also place greater loads on the plantar fascia. Despite that the term “excessive” is commonly used by clinicians to describe certain magnitudes of pronation and loading, it remains difficult to define quantitatively. Nevertheless, clinicians theoretically believe that excessive kinematics and kinetics play a key role in the development and prolongation of recalcitrant PF.

The findings of biomechanical studies, however, are contrary to the clinical assertion that foot overpronation and PF are associated. Research studies in rearfoot motion (Messier and Pittala, 1988; Warren and Jones, 1987), arch kinematics (Wearing et al., 2004), and arch height (Rome et al., 2001; Warren, 1984) have not found a relationship between these characteristics and PF. There are two limitations with these studies. First, there are errors associated with evaluating overpronation, a movement that is three dimensional (3D) in nature, with a two dimensional (2D) measurement. Second, the modeling of the foot as a single rigid segment is problematic. The plantar fascia attaches to the rearfoot, forefoot and toes, and therefore, the plantar fascia can become elongated.
with intrinsic foot motion. It has long been shown that the movements of the medial arch and the hallux are strongly related to the dynamics of the plantar fascia (Hicks, 1954). Therefore, modeling of the foot as a single rigid segment provides no insight regarding the deformation and loading of the plantar fascia, and limits our understanding of how the plantar fascia may become injured. These two limitations can be overcome using 3D multi-segment foot models (Pohl and Buckley, 2008; Rao et al., 2007) and can potentially shed some light on the foot kinematics pertinent to PF (Chang et al., 2008).

Moreover, there is disagreement in the literature concerning the extent to which vertical GRF are affected in individuals with PF in comparison to healthy controls. Some researchers have shown that vertical GRF are unchanged in individuals with PF during gait (Liddle et al., 2000; Wearing et al., 2003), while others have shown reductions in the peak magnitudes (Katoh et al., 1983). These previous studies were conducted at subject-selected walking speeds, however, peak GRFs are directly related to walking speed (Andriacchi et al., 1977). Plantar fasciitis individuals may have selected a slower walking speed to compensate for pain. Controlling walking speed may provide additional insights as to whether GRFs are altered in individuals with PF.

Therefore, the purpose of this study was to determine whether healthy and PF feet are different with respect to multi-segment foot kinematics and GRF. Compared to healthy controls, we hypothesized that individuals with PF would exhibit greater rearfoot, forefoot, and hallux motion (i.e. greater maxima, total excursions, and maximum angular velocities). Additionally, we hypothesized that the peak vertical GRF at loading and at propulsion would differ between PF and healthy controls.

2. Methods

2.1. Participants

Twenty-two healthy controls and 22 chronic PF individuals gave their informed consent to participate. Individuals qualified if they were 30–60 years of age. Participants were limited to 60 years to minimize the potential confounding influence of age-related changes to plantar soft tissue (Kwan et al., 2010). All potential participants underwent a clinical examination by a Canadian Certified Pedorthist with previous clinical experience with PF and various other foot pathologies. The examination consisted of a clinical history, functional tests, palpations, and range of motion tests of the major joints of the foot and shank. Participants in the control group qualified if they had no history of injury or foot pain, and had no pain elicited during the exam. Individuals with PF were included if they had heel pain upon palpation of the plantar fascia’s insertion point, persistent symptoms for at least three months leading up to the study, experienced a minimum of five episodes of first-step pain within the last month (a hallmark of PF), and were otherwise healthy. Potential PF participants were screened with awareness that there are other forms of heel pain with presentations similar to PF (e.g. Achilles tendinitis, heel fat pad syndrome, calcaneal stress fracture). Exclusion criteria included a history of a local steroid injection within the last 2 months, arthritis (self-report), local traumatic injury, and a body mass index greater than 35. Foot posture was quantified via the standing arch ratio (Williams and McClay, 2000) and the foot posture index (Bedmond et al., 2006). Due to their purported mechanical differences, we excluded individuals with a high arch foot type (Schuster, 1977) (a standing arch ratio one standard deviation above our laboratory’s mean).

2.2. Protocol

Spherical markers (8 mm diameter) were fixed to the skin according to a multi-segment foot model (Leardini et al., 2007). The foot model included a rearfoot, a medial forefoot, and a hallux (Fig. 1). All foot markers remained on the skin for both the standing calibration trials and the dynamic trials.

The shank was defined and tracked using an existing shank model and marker set (Manal et al., 2000). The shank segment was defined by four segment

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**Fig. 1.** (a) Multi-segment foot model consisting of three dimensional medial forefoot and rearfoot segments, and a two dimensional hallux line segment. A laboratory coordinate system and segment coordinate systems are provided for medial forefoot and rearfoot segments constructed from anatomically placed skin markers (Leardini et al., 2007) (first metatarsal (FM), second metatarsal (SM), head (H), base (B), peroneal tubercle (PT) sustentaculum tali (ST), calcaneus (CA)). The rearfoot’s origin was located at CA. The rearfoot’s Y-axis was aligned to a midpoint between the ST and the PT. The rearfoot’s X-axis was aligned to a transverse plane defined by rearfoot Y-axis and the ST. The rearfoot’s Z-axis was orthogonal to the rearfoot’s XY plane. The medial forefoot’s origin was located at the SMB. The medial forefoot’s Y-axis was a projection of the line joining SMB and SMH on the transverse plane passing through the origin and FMH and VMH. The medial forefoot X-axis was orthogonal to the rearfoot X-axis and resides in the transverse plane. The medial forefoot Z-axis was orthogonal to the medial forefoot XY plane. Hallux line segment was defined by the marker on the proximal phalans of the hallux (PM) and the FMH. Lower image provides a sagittal view of the foot model with sagittal medial forefoot angle inscribed. (B) Lateral view of the foot and shank. A rigid set of four markers on a plate was fixed to the lateral shank (lateral malleolus (LM)). (C) A medial view of the foot and shank with medial malleolus marker (MM) shown.
definitions markers. There were two proximal markers (medial and lateral femoral epicondyles), and two distal markers (the medial and lateral malleoli). A rigid cluster of four markers located at the distal–lateral shank, whose static position was associated with the segment definition markers, was used to track shank movements. The four shank segment definition shank markers were removed during movement trials. The movements of the shank were tracked by movements of the rigid cluster.

Kinematic data were collected in 3D (240 Hz) using eight motion capture cameras (Qualisys Oqus 3. AB, Gothenburg). Kinetic data (1920 Hz) were synchronized with kinematic data and collected using a force platform (AMTI Inc., Watertown, MA). Each participant performed a neutral standing calibration trial (feet, shoulders, and hips pointed straight in the walking direction), practice walking trials, and at least five acceptable walking trials at 1.35 ms\(^{-1}\) ± 5% on a straight 10 m walkway.

Data processing and model building were performed in Visual 3D\(^\text{TM}\) (C-Motion Inc., Germantown). In bilaterally symmetrical PF participants (n = 12), data for the more symmetrical limb was processed. If a PF participant was affected equally on both limbs, the affected right or left limb was randomly selected. For the control group, data were collected bilaterally, but once the data collection stage was completed for the entire study, the number of right and left data sets used from the control group was randomly matched to the number obtained from the PF group. Marker histories and analog signals were smoothed with a 4th order, low-pass Butterworth filter at 8 Hz and 70 Hz, respectively. Cut off frequencies were chosen from a residual analysis (Winter, 2004).

Coordinate system constructions for the rearfoot, medial forefoot, and hallucus line segment were right-handed, and were made according to the original model (Leardini et al., 2007) (Fig. 1). Since the publication of the foot model (Leardini et al., 2007), it has been shown that fifth metatarsal (lateral foot) kinematics are different from the medial foot (Lundgren et al., 2008), and therefore, we chose to create a medial forefoot segment (as opposed to having a foot segment which combines both medial and lateral markers). The only distinction between the present model and the original model was the tracking markers of the foot. In the original model, forefoot motion was tracked with SMB, FMH, and VMH, while the present medial forefoot model tracked motion via MBM, SMB, FMH, and SNBH (Fig. 1 contains abbreviations and landmarks). The coordinate system for the shank was right-handed and its origin was located at the mid-point between the medial and lateral malleoli (X-axis oriented to lateral malleolus; Z-axis oriented vertically to mid-point between medial and lateral femoral epicondyles; Y-axis was orthogonal to X and Z, and oriented to the anterior direction). Joint angles were calculated using a Cardan XYZ sequence of rotations with six degrees of freedom; the distal segment relative to the proximal segment (i.e. rearfoot angle: rearfoot to shank; forefoot angle: medial forefoot to rearfoot; first metatarsal–phalangeal joint angle (FMJP); hallucis to forefoot) (Cole et al., 1993; Leardini et al., 2007). Medial arch motion was inferred from the medial forefoot angle (dorsiflexion was representative of a flattening of the medial arch, and plantarflexion was representative of a rising/higher medial arch). FMJP angles were 2D (flexion-extension in the sagittal plane), and were computed as the angle formed by the hallucus line segment relative to the medial forefoot segment (Leardini et al., 2007). Stance was identified using the vertical GRF data and a 15 N threshold. Joint and segment angles were normalized to angles obtained in the standing calibration position and time scaled to 100% of stance.

Kinematic curves and discrete kinematic variables were averaged across five trials for each participant. Discrete stance-phase kinematic variables of interest included (1) maxima, (2) value at initial contact, (3) total motion (difference between (1) and (2)), and (4) maximum angular velocity in stance. These variables included (1) maxima, (2) value at initial contact, (3) total motion (difference between (1) and (2)), and (4) maximum angular velocity in stance. These variables were calculated using a Cardan XYZ sequence of rotations with six degrees of freedom; the distal segment relative to the proximal segment (i.e. rearfoot angle: rearfoot to shank; forefoot angle: medial forefoot to rearfoot; first metatarsal–phalangeal joint angle (FMJP); hallucis to forefoot) (Cole et al., 1993; Leardini et al., 2007). Medial arch motion was inferred from the medial forefoot angle (dorsiflexion was representative of a flattening of the medial arch, and plantarflexion was representative of a rising/higher medial arch). FMJP angles were 2D (flexion-extension in the sagittal plane), and were computed as the angle formed by the hallucus line segment relative to the medial forefoot segment (Leardini et al., 2007). Stance was identified using the vertical GRF data and a 15 N threshold. Joint and segment angles were normalized to angles obtained in the standing calibration position and time scaled to 100% of stance.

Qualitatively, the overall movement patterns of the medial forefoot were similar between groups (Fig. 3). From initial contact initial contact to mid- and late stance the medial forefoot was pronated; namely it was dorsiflexed, everted and abducted, with the greatest motion occurring in the sagittal plane, followed by the frontal and transverse planes. Into late stance, this movement

### Table 1

<table>
<thead>
<tr>
<th>Variable</th>
<th>Control</th>
<th>Plantar fasciitis</th>
<th>p-Value</th>
<th>Effect size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>44.0 (10.0)</td>
<td>42.9 (7.6)</td>
<td>0.69</td>
<td></td>
</tr>
<tr>
<td>Height (cm)</td>
<td>171.0 (7.2)</td>
<td>165.6 (7.2)</td>
<td>0.47</td>
<td></td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>72.5 (13.0)</td>
<td>74.3 (11.8)</td>
<td>0.62</td>
<td></td>
</tr>
<tr>
<td>Standing arch ratio</td>
<td>0.327 (0.019)</td>
<td>0.318 (0.022)</td>
<td>0.15</td>
<td></td>
</tr>
<tr>
<td>Foot posture index</td>
<td>2.6 (3.0)</td>
<td>4.0 (3.8)</td>
<td>0.20</td>
<td></td>
</tr>
<tr>
<td>Preferred walking speed (m/s(^{-1}))</td>
<td>1.31 (0.17)</td>
<td>1.28 (0.16)</td>
<td>0.60</td>
<td></td>
</tr>
</tbody>
</table>

### Table 2

<table>
<thead>
<tr>
<th>Variable</th>
<th>CON</th>
<th>PF</th>
<th>p-Value</th>
<th>Effect size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rearfoot</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Inversion IC (°)</td>
<td>2.7 (1.9)</td>
<td>3.6 (2.5)</td>
<td>0.10</td>
<td>0.39</td>
</tr>
<tr>
<td>Eversion Max (°)</td>
<td>3.5 (1.4)</td>
<td>3.8 (1.8)</td>
<td>0.29</td>
<td>0.17</td>
</tr>
<tr>
<td>Total (°)</td>
<td>6.2 (1.4)</td>
<td>7.4 (2.9)</td>
<td>0.049</td>
<td>0.51</td>
</tr>
<tr>
<td>Eversion Max Vel (°/s(^{-1}))</td>
<td>43.3 (20.0)</td>
<td>56.7 (38.0)</td>
<td>0.08</td>
<td>0.44</td>
</tr>
<tr>
<td>Medial forefoot</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantarflexion IC (°)</td>
<td>2.7 (1.7)</td>
<td>3.7 (2.0)</td>
<td>0.04</td>
<td>0.55</td>
</tr>
<tr>
<td>Dorsiflexion max (°)</td>
<td>6.7 (1.4)</td>
<td>6.6 (2.6)</td>
<td>0.46</td>
<td>0.03</td>
</tr>
<tr>
<td>Total (°)</td>
<td>9.4 (1.9)</td>
<td>10.3 (1.9)</td>
<td>0.049</td>
<td>0.50</td>
</tr>
<tr>
<td>Dorsiflexion max vel (°/s(^{-1}))</td>
<td>75.7 (30.1)</td>
<td>75.1 (27.0)</td>
<td>0.48</td>
<td>0.01</td>
</tr>
<tr>
<td>FMJP</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dorsiflexion max</td>
<td>49.0 (73)</td>
<td>53.3 (8.0)</td>
<td>0.04</td>
<td>0.56</td>
</tr>
<tr>
<td>Kinetic</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>GRF1 (3BW)</td>
<td>1.080 (0.073)</td>
<td>1.056 (0.063)</td>
<td>0.25</td>
<td>0.35</td>
</tr>
<tr>
<td>GRF2 (3BW)</td>
<td>1.100 (0.056)</td>
<td>1.059 (0.077)</td>
<td>0.046</td>
<td>0.62</td>
</tr>
</tbody>
</table>

The foot function index revealed that PF participants reported significantly (p < 0.001) more: pain, stiffness, disability, activity limitation and social/emotional issues (Budiman-Mak et al., 2006).

3.2. Rearfoot motion

Plantar fasciitis individuals had a significantly greater total rearfoot motion than healthy individuals (p = 0.049) and tended toward having a greater maximum evasion velocities (p = 0.08) (Table 2). No significant differences were found in the rearfoot positions at initial contact or at maximum evasion of stance phase. Overall the movement pattern of both groups consisted of landing in an inverted position at initial contact, evertig to mid-stance (~60%), and finally inverting toward push-off (Fig. 2).

3.3. Medial forefoot motion

Some significant differences were found in the medial forefoot motion (Table 2). PF participants demonstrated significantly greater total plantar-dorsiflexion motion (p = 0.049, ES = 0.50). At initial contact, the medial forefoot of the PF group was more plantar flexed than controls (p = 0.04, ES = 0.55). No group mean differences were found in: maximum medial forefoot dorsiflexion, maximum evasion, total inversion–eversion motion, and maximum abduction angle.
pattern reversed in the sagittal and transverse planes, with the medial forefoot plantarflexing and adducting. In contrast, the medial forefoot continued to evert.

3.4. First metatarsophalangeal joint motion

PF individuals exhibited significantly greater maximum FMPJ dorsiflexion in late stance (Table 2, $p=0.04$, ES=0.56). For both PF and healthy participants, the FMPJ was in approximately 18° of dorsiflexion at initial contact and rotated to a neutral position (~0°) at mid-stance. From mid-stance to push-off, the FMPJ primarily dorsiflexed until late stance (95%), followed by slight plantar flexion (Fig. 4).

3.5. GRF

In the vertical direction, PF individuals demonstrated significantly lower peak vertical forces at propulsion ($p=0.046$, ES=0.64) (Table 2). There was no significant differences in peak forces associated with impact ($p=0.25$, ES=0.35).

4. Discussion

Clinicians believe that PF is an overuse injury of the plantar fascia and that biomechanical factors, in particular overpronation and increased GRF, play a significant role in its development (Kwong et al., 1988; Subotnick, 1981; Taunton et al., 1982). However, difficulties in measuring intrinsic foot kinematics have resulted in a lack of in vivo data to support this clinical assumption. The purpose of this study was to determine whether individuals with PF have different kinematics and kinetics compared to healthy individuals while controlling foot anthropometrics.

4.1. Rearfoot motion

Our hypothesis that individuals with PF would exhibit greater rearfoot motion was partially supported. There were no significant group differences in maximum rearfoot eversion, but total rearfoot (inversion–eversion) motion was significantly greater in PF individuals. These data suggest that those with chronic PF exhibit a greater magnitude of foot pronation in comparison to healthy individuals, and clinically, a greater degree of pronation is thought to put a greater stress and strain upon the plantar fascia. There is a dearth of data on what is a clinically meaningful difference when evaluating kinematic data. In this study, the mean group difference
was 1.2°, which may be considered small at first glance, however, previous studies and statistics suggest a meaningful result. These present data agree with other previous kinematic studies using skin markers (Hunt et al., 2001; Ratanapraser et al., 1999) and bone-pinned markers (Lundgren et al., 2008) which have shown that there is < 10° of rearfoot eversion from initial contact to the maximum at midstance. Statistically, the difference was a medium effect size, and it represents a 19.3% relative increase of the total motion exhibited by PF individuals. Planter fasciitis is an overuse injury in which a 19.3% increase in motion and/or loading with every step may be injurious to soft tissue repeated periods of prolonged walking. Planter fasciitis individuals exhibited greater maximum eversion velocity (30.9%), but the difference did not meet the a priori level of significance. Within the rearfoot motion variables, only total motion was significant, therefore, we concluded that there was some support for the notion that individuals with PF exhibit pronounced rearfoot motion.

Previous rearfoot motion studies in healthy individuals and with PF conducted in 2D have not detected differences (Messier and Pittala, 1988; Warren and Jones, 1987). It has been shown that 2D techniques maybe prone to error and overestimation, and therefore, the methodological limitations of the past may have masked the differences which were detected using a 3D measurement technique (Areblad et al., 1990).

4.2. Medial forefoot motion

By examining the medial forefoot motion with respect to the rearfoot, we gained insight into the dynamics of the medial arch, and the plantar fascia. Consistent with the clinical concept linking overpronation and injury, we hypothesized that individuals with PF would exhibit greater medial forefoot kinematics. We detected significantly greater total sagittal forefoot motion in PF individuals compared to healthy individuals. However, it was surprising that the greater total motion appeared not due to greater medial forefoot dorsiflexion (flatter medial arch), rather it was attributable in large part to the significant difference in the position of the medial foot at initial contact. The PF forefoot was in a greater degree of plantar flexion (higher arch position) at initial contact, and then achieved a similar peak dorsiflexion magnitude in stance. We speculate that the increased medial forefoot plantarflexion at initial contact may be a pain guarding mechanism, potentially a result of increased plantar intrinsic foot muscles activity.

Although the differences in medial forefoot motion between the PF and healthy individuals may be considered small at first glance, several studies have shown that the intricate movements of the foot can have a profound effect on the stress and strain of the plantar fascia. For instance, surgical implants restricting overpronation can significantly reduce plantar fascia elongation < 1 mm, but can also significantly reduce elongation of the plantar fascia by 33% (Graham et al., 2011). Similarly, an arch support can significantly reduce plantar fascia strain by 34.8% by supporting the arch and reducing the distance between the rearfoot and forefoot (1 mm) (Ferber and Benson, 2011). Also, a change in the arch angle of ~ 1° can result in a concurrent increase in plantar fascia tension from 0.4 BW to 0.7 BW during 0% to 50% of stance (Caravaggi et al., 2010). These previous studies together suggest that the greater medial forefoot motion exhibited by the PF group albeit 1.0° may result in a higher magnitude of strain, and prolong injury.

It is possible that PF individuals compensated to reduce plantar fascia strain. It has been shown that activation of the plantar intrinsic foot muscles work alongside the plantar fascia to increase the height of the medial arch, reduces medial arch collapse, and reduces tension on the plantar fascia (Fiolkowski et al., 2003). Therefore, the forefoot differences that we detected did not appear to be associated with what clinicians would consider causational factors of PF, rather they appeared to be compensatory responses.

We compared the present forefoot kinematic data to a previous study that also examined PF individuals (Wearing et al., 2004). The previous study reported larger total sagittal plane motion (11.4–13.3°) in comparison to these (7.7–8.5°), and they detected no difference in sagittal forefoot plane motion in individuals with PF compared to healthy controls (Wearing et al., 2004). However, a comparison of the data is difficult in light of several key methodological differences, such as different rearfoot and forefoot segment definitions, the use of 2D fluoroscopy, and a lower sampling rate (15 Hz). Furthermore, the previous study’s data capture period was limited to the first 80% of stance phase which may present measurement difficulties if the maximum forefoot angle occurs at around 80% stance or later, which is seen here and in other data (Chang et al., 2008; Hunt et al., 2001).

Frontal and transverse plane forefoot movements during the loading phase of gait were small in comparison to the sagittal plane (approximately one half). In these planes, differences did not reach statistical significance nor exceeded a medium effect size to support the hypotheses. Given the small ranges of motion, it was concluded that frontal and transverse forefoot motion are not characteristically different in those with PF.

4.3. First metatarsal phalangeal joint motion

Plantar fasciitis individuals exhibited greater peak FMPJ dorsiflexion, a movement pattern that might predispose an individual to PF, or prolong the injury. Cadaver models (Flanagan et al., 2007; Hicks, 1951), and finite element analyses have confirmed that tension in the plantar fascia rises directly with the magnitude of toe dorsiflexion (Cheng et al., 2008). Furthermore, there is a stress concentration in the plantar fascia under the first ray and medial calcaneal tubercle (Cheng et al., 2008), supporting the tenet that the FMPJ contributes relatively more than the lesser toes to the windlass mechanism (Hicks, 1951). These locations of high stress also coincide with sites of pain which PF patients typically report. Elevated FMPJ dorsiflexion over multiple cycles of gait could put undue strain on the plantar fascia thereby predisposing or prolonging a state of PF.

The measured FMPJ movement patterns and initial contact values agree with previous literature (Mann and Hagy, 1979). Some differences were noted in present peak dorsiflexion values in comparison to the literature. This study mean of 51.2° was greater than others who have reported 39–42° (Halstead et al., 2005; Nawoczenski et al., 1999; Nawoczenski and Ludewig, 2004). Those previous studies used a 3D electromagnetic system and their transmitters are relatively larger, tethered, and heavier than the present skin markers. Our results reside within the normative range of motion for the FMPJ (Sherreff et al., 1986), and may represent a more unobstructed movement pattern than some previous literature.

4.4. GRF

Plantar fasciitis individuals exhibited a reduced peak vertical GRF during propulsion (second vertical peak) in comparison to healthy individuals. Anecdotally, push off is a moment during stance when many patients report an increase in pain. Instrumented dynamic cadavers and computer simulations have reported that the plantar fascia tension peaks at approximately one body weight at ~80% stance (Erdemir et al., 2004; Scott and Winter, 1999). Therefore, it is plausible that some individuals made gait compensations to reduce this propulsive peak in attempt to manage pain.
Previous findings in peak plantar fascia GRF have been inconsistent, but previous studies have not controlled walking speed, which may mask group differences. It has been reported that PF was associated with reduced peak vertical GRF, but their participants walked at a slower speed (mean 1.19 ms\(^{-1}\)) than controls (mean 1.38 ms\(^{-1}\)) (Katoh et al., 1983), and a higher walking speed elicits increases in peak GRF (Andriacchi et al., 1977). Two other studies reported no differences in the magnitudes of the vertical GRF when comparing symptomatic feet, asymptomatic contra-lateral feet, and healthy individuals (Liddle et al., 2000; Wearing et al., 2003). Neither study addressed the possibility of walking speed as a confounding factor. Future studies could also consider controlling stride length and stride rate (Martin and Marsh, 1992).

4.5. Limitations

The current study was a case-control design, and therefore, its retrospective nature is an overriding limitation. The PF individuals included in this study were considered chronic cases (symptomatic > 3 months). Therefore, we cannot state with a high-level of certainty whether the observed differences were causational, perpetuated the injury, or compensatory responses due to prolonged injury, or a combination. We can only speculate applying the current state of knowledge of foot biomechanics. Skin markers were pursued over bone pinned markers since they are non-invasive, but estimations of bone poses using skin markers have limitations. It has been shown that skin markers oscillate (Karlsson and Tranberg, 1999), approximate the underlying bone position (Reinschmidt et al., 1997), and there is variability in identification of anatomical landmarks. Nevertheless, it has been shown that skin markers on the foot have high levels of correlation with corresponding bony landmarks in their movement patterns (Wrbaskic and Dowling, 2007). Errors associated with skin markers were minimized to the best of our abilities. The marker set was designed to avoid tendon elevation artifacts (Leardini et al., 2007), and the markers used in this study were a relatively small size in comparison to other protocols. Last, the variability due to inter-tester marker placement was circumvented by having an experienced Canadian Certified Pedorthist prepare all participants.

4.6. Summary

In summary, there were biomechanical differences between healthy and PF individuals despite similar foot anthropometrics. Group differences were observed in kinematics of the rearfoot, medial forefoot, FMPJ, and we also observed differences in GRF. Due to the retrospective nature of the study, it was difficult to determine whether these differences caused the injury, prolonged the injury, or were compensatory responses due to injury, however, we speculate that we observed biomechanical differences representing a combination thereof.

Conflict of interest statement

The authors declare no commercial interests which would benefit from the publication of this manuscript.

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