Foot for Thought: Identifying Causes of Foot and Leg Pain in Rural Madagascar to Improve Musculoskeletal Health

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Spring 2018

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This Graduation with Distinction Honors Thesis is submitted in partial fulfillment of the requirements for Graduation with Distinction in Evolutionary Anthropology and Global Health.

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Abstract

Incidence of musculoskeletal health disorders is increasing in Madagascar. Foot pain in the Malagasy may be related to daily occupational activities or foot shape and lack of footwear. Our study tests hypotheses concerning the cause of foot pain in male and female Malagasy populations and its effects on gait kinematics. The study was conducted in Mandena, Madagascar. We obtained 89 participants’ height, mass, and age from a related study ($n_{\text{male}} = 41$, $n_{\text{female}} = 48$). We collected self-report data on daily activity and foot and lower limb pain. A modified Revised Foot Function Index (FFI-R) assessed pain, difficulty, and limitation of activities because of reported foot pain (total score = 27). We quantified ten standard foot shape measures. Participants walked across a force platform at self-selected speeds while being videorecorded at 120 fps. Females reported higher FFI-R scores ($p = 0.029$), spending more hours on their feet ($p = 0.0184$), and had larger BMIs ($p = 0.0001$) than males. Strong linear relationships were examined between participants’ self-selected speed and force curve peaks and loading rates. No significant differences were found in force curve parameters between participants with foot/ankle/knee pain and lack thereof. Males showed higher values of force curve parameters and steeper slopes when relating velocity to the same parameters. The higher foot pain and lower force peaks in females may be related to the combination of higher BMI, small feet relative to BMI, and the amount of time they are on their feet. Results suggest that a combination of BMI, foot size, and occupational factors influence foot pain in this community leading to long term injury and limitations on work. These results will help guide future interventions that promote engagement in leisure/work activities.
Introduction

The prevalence of musculoskeletal disorders in Madagascar has increased 8.97% since 1990 - from 8,337 to 9,058 cases per 100,000 people (Wang et al., 2016). The increase has been especially prevalent in Malagasy between the ages of 15-49 (5.55%), the prime working age (Wang et al., 2016). It is important to address this increase in musculoskeletal disorders in a population that relies on manual labor as a central part of their economy. The associated long distance walking while carrying heavy loads increases the risk of musculoskeletal injuries. Although the risk of musculoskeletal injuries is high in this region and could result in physical and economic consequences, there is minimal research on musculoskeletal health in Madagascar.

Trentadue, Schmitt, and Nunn (2016) investigated the relationship between vertical ground reaction forces (VGRFs) and joint pain in the knee and hip using Fourier analysis. Although valuable information was found on the association between VGRF and joint pain, risk factors associated with musculoskeletal disorders in the Malagasy are still unknown. As a result, this study investigated how foot morphology is associated with lower limb injuries that impede locomotion.

Evolution of Bipedalism

The origins of bipedalism show that human gait is similar amidst variation in foot morphology. Experimental research indicates that hominid bipedalism originated from an arboreal climbing primate (Schmitt, 2003). Early hominids exhibited a compliant-styled gait, while modern hominids now show a stiff-legged gait that is often modeled by an inverted pendulum (Cavagna, Thys, & Zamboni, 1976; Queen, Sparling, & Schmitt, 2016; Schmitt, 2003).
Figure 1. Application of the inverted pendulum model on human gait cycle. Kinetic energy and potential energy fluctuate throughout stride. If we assume there is 100% energy recovery during gait, the amplitudes for kinetic and potential energy will be the same (Queen et al., 2016).

The string of the pendulum represents the stiff leg, connecting the point of contact of the foot to the body’s center of mass. The model highlights the change in mechanical energy during the gait cycle. Following heel strike and the initial stance phase, the body begins to convert its kinetic energy to potential energy as it approaches midstance, in which the leg is aligned perpendicularly to the foot. During midstance, mechanical energy is mostly potential energy; however, it is converted into kinetic energy during the terminal stance phase as an individual distributes their weight to their phalanges and push off from their toes. Hence, the terminal stance phase is also known as “toe-off” (Cavagna et al., 1976; Lee & Farley, 1998).
Figure 2. Standard curves corresponding to a single step of a run, speed-walk, and walk. Our study will examine three points of interest on the “Walk” curve. The first peak, the valley between the two peaks, and the last peak (Tongen & Wunderlich, 2010).

The first peak of the VGRF curve for walking corresponds with heel strike as the body’s center of gravity is driven towards the ground. Hence, this peak is also referred to as the **loading peak**. The valley following it corresponds the body’s center of mass moving away from the ground, which occurs during **midstance**. The third peak, also known as the **propulsive peak**, corresponds to toe-off. (Tongen & Wunderlich, 2010).

We expect most potential energy to be exchanged for kinetic energy and vice versa during a gait cycle for a healthy human (Queen et al., 2016). With the onset of foot and lower limb joint pain, it is possible that this exchange in energy will be inefficient and will be compromised to reduce the onset of pain. Furthermore, this compensation could be visible on participants’ respective VGRF curves. We will monitor the gait cycles and VGRF points of interest for our Malagasy participants to see how self-reported pain affects gait kinematics and energy exchange while walking.
Foot Morphology and Plasticity

The human foot is plastic and its morphology can be influenced by a variety of factors as we age (Barnett, 1962; D’Août, Pataky, Clercq, & Aerts, 2009; Hoffmann, 1905; Scott, Menz, & Newcombe, 2007). Before discussing these changes, it is important to understand the foot’s morphological components.

Figure 3. Superior view of the left human foot (Morton, 1935).

The human foot can be split into three areas – rearfoot, midfoot, and forefoot. The rearfoot consists of the talus and the calcaneus. The midfoot consists of the navicular, cuboid, and the three cuneiform bones. Lastly, the forefoot is made up of the metatarsals and phalanges (Figure 3). The metatarsals are numbered 1-5 from the most medial to lateral metatarsals. The joints located between the metatarsals and phalanges are the metatarsophalangeal joints (MTPJs) and are numbered based on their respective metatarsals. The four most lateral phalanges can be split into distal, middle, and proximal phalanges. The most medial phalange is the hallux, commonly referred to as the “big toe.”
Foot morphology changes with age and shoewear. With increasing age, human feet become more pronated, have limited flexion in the ankle and first MTPJ, and are more likely to exhibit medial deviation of the first metatarsal and lateral deviation and/or rotation of the hallux (Scott et al., 2007). There are distinct differences between feet in shod and minimally shod populations (D’Août et al., 2009; Hoffmann, 1905). Shod populations are considered groups where, starting at a young age, individuals habitually wear shoes that cover their phalanges (e.g., closed toe shoes such as athletic shoes or boots). In contrast, minimally shod populations are characterized by being habitually barefoot or wearing footwear that does bind the phalanges, such as flip-flops or sandals while outside, but are barefoot indoors (Griffin, Miller, Schmitt, & D’Août, 2013; Hatala, Dingwall, Wunderlich, & Richmond, 2013a).

Moreover, shoes that compress toes in shod populations result in a misalignment between phalanges and metatarsals as an adult (Hoffmann, 1905). In contrast, phalanges are in alignment with their respective metatarsals in unshod populations. Medial deviation of the first metatarsal and lateral deviation and/or rotation of the hallux has also shown a positive relationship with constrictive footwear (Barnett, 1962). Recently, unshod populations were also found to have wider and longer feet than shod populations, hence, being able to distribute plantar pressure more evenly (D’AoUt, Pataky, Clercq, & Aerts, 2009). It is possible that unshod populations receive reduced local loads at points on the foot because of increased surface area and have a lower risk of foot injuries such as plantar fasciitis, which may be associated with high bending moments around parts of the foot.

Changes in foot shape can change the distribution of weight during gait. These variations in foot morphology between shod and minimally shod populations could influence the onset of musculoskeletal injuries because of a change in gait kinematics. The feet of the Malagasy are
unusual in comparison to the Western foot (C. L. Nunn, personal communication, February 8, 2017). This unusual morphology may be associated with the lack of footwear seen in rural areas in developing nations throughout the world. However, with the rapid pace of globalization in Mandena, it is possible that more Malagasy are adopting restrictive footwear, which could be altering foot morphology and making them more susceptible to foot injuries, hence, increasing the rate of musculoskeletal disorders. It will be important to see how footwear history is correlated with foot shape and musculoskeletal injuries in the Malagasy to prevent future foot and lower limb joint pain.

**Evolution of Foot Mechanics**

Research has also considered the evolution of the metatarsophalangeal joint in the foot. Humans dorsiflex their metatarsophalangeal joints significantly more than bonobos during the terminal stance phase (Griffin, D’Août, Richmond, Gordon, & Aerts, 2010). Dorsal flexion of the first metatarsophalangeal joint (MTPJ1) in humans can also vary based on subtalar pronation (Griffin et al., 2013). One study found no significant relationship between the two variables; however, two other studies have found such a relationship (Griffin et al., 2013; Kappel-Bargas, Woolf, Cornwall, & McPoil, 1998; Nakamura & Kakurai, 2003). It is important to continue to consider this topic as variation in MTPJ1 flexion affects tension on the plantar aponeurosis, and hence, can introduce possible risk factors for foot pain, such as plantar fasciitis.

The windlass mechanism describes how the triceps surae uses the calcaneus as a lever point to plantarflex the foot, pulling and increasing tension in the plantar aponeurosis, raising the medial longitudinal arch and dorsiflexing the MTPJ1 (Griffin, Miller, Schmitt, & D’Août, 2015, *Figure 4*). It has evolved to help bipedal humans conserve 15% in energy while walking. The human achilles tendon differs from that of other primates because it has a relatively long length,
which helps store and release more energy while walking. Human proximal phalanges have flattened because of required MTPJ flexion during the windlass mechanism. Increased flexion at the first MTPJ during toe-off would increase tension on the plantar aponeurosis, making an individual more susceptible to pain from overuse. In addition to looking at MTPJ flexion, we can also see how much the navicular drops during gait to assess subtalar pronation and how much the plantar aponeurosis is stretched while walking. We expect to see less of a drop in the navicular, hence less pronation, with individuals with pain below the foot to limit stretching of the plantar aponeurosis and prevent the onset of pain.

Figure 4. Elements of the windlass mechanism. A) triceps surae, B) plantar aponeurosis, C) medial longitudinal arch (striped line), D) metatarsophalangeal joints (MTPJs) (Griffin et al., 2015).

Foot and Ankle Morphology in Males and Females

Studies have also shown variation in foot and ankle shape between males and females (Putti, Arnold, & Abboud, 2010; Sinclair, Greenhalgh, Edmundson, Brooks, & Hobbs, 2012; Wunderlich & Cavanagh, 2001; Zifchock, Davis, Hillstrom, & Song, 2006). When standardized
for foot length, male and female feet have shown significant differences in foot, ankle, and calf morphology (Wunderlich & Cavanagh, 2001). Males also have broader feet than females and a greater contact area with the ground while walking (Putti et al., 2010). Hence, when standardized for speed, we expect males to have greater contact time between their feet and the ground while walking. Moreover, females display greater knee abduction, knee internal rotation, and ankle eversion while males have greater hip flexion while running. These differences in kinematics could also be attributed to differences in foot and ankle morphology, although not examined in this study (Sinclair et al., 2012). Although there is no difference in medial longitudinal arch height between males and females, female arches are significantly less stiff (Zifchock et al., 2006). As part of our comparison of loading on the human foot between shod and minimally shod populations, we will also examine whether some differences may also be due to sex differences.

**Evolution of Foot Strike and Injury**

Research conducted on foot strike often focuses on running; however, its findings on kinematics and vertical grounds reaction forces and their relationship to injury can be applied to walking as well. Lieberman et al. (2010) attracted a large amount of attention to barefoot running because of their study on foot strike patterns and collision forces on habitually shod and unshod runners. Most habitually barefoot runners struck the ground with the front of their foot, while shod runners struck the ground with the heel of the foot. Front-foot striking resulted in smaller collision forces for these unshod runners and that the impact transient typically seen during heel-strike could be attributing to modern-day running injuries. However, this study was replicated with another habitually barefoot population in Africa and it discovered that this population was heel-striking when asked to demonstrate endurance running (Hatala, Dingwall, Wunderlich, &
Richmond, 2013b). This exposed shortcomings to Lieberman et al.’s (2010) study, highlighting how the study recruited recreational runners who ran across a force platform at a speed (5-6 m/s) between what would be considered endurance running and sprinting. When humans sprint, we typically land with the front of our feet, which may have been the case for Lieberman et al.’s (2010) sample. Nonetheless, additional research needs to look at the possible consequences (if any) of heel-striking and the significance of the impact transient of the force curve in terms of diagnosing an injury. Our study will look at the relationship between footstrike and foot and lower limb pain in our minimally shod population in Mandena. Heel-striking in this population during walking can introduce a large impact transient on the foot that could serve as a risk factor for foot and lower limb joint pain.

**Foot Function and Gait Parameters**

A few studies have already assessed the effect of foot shape on gait parameters (Hillstrom et al., 2013; Mootanah et al., 2013). Differences in foot structure between planus, rectus, and cavus feet in asymptomatic individuals have been related to differences in peak pressures, maximum force, and force-time integrals (Hillstrom et al., 2013). When paired with anthropometric and walking speed data, foot structure plays less of a role in variance in gait parameters, however, it can still influence variance in force peaks and impulse (Mootanah et al., 2013). Based on these studies, we can hypothesize that walking speed and anthropometric data will play more of a role in variance in gait parameters between our comparative samples rather than foot shape. However, it is important to note that these studies have only been conducted on asymptomatic individuals and that our sample will contain participants who have complained about lower limb and foot pain. Hence, it is possible that this relationship may not exist in symptomatic individuals.
**Osteoarthritis and Gait**

Osteoarthritis (OA) is a progressive disease involving joint deterioration and inflammation that can inhibit daily activity if found in the lower limb. It poses a risk to the Malagasy because its onset can prevent them from supporting and feeding their families because of their reliance on manual labor. Symptoms include joint pain, stiffness, and locomotor restriction (Hurley, Scott, Rees, & Newham, 1997). OA in any of the three lower-limb joints (ankle, knee, and hip) has resulted in lower peak vertical ground reaction forces (Schmitt, Vap, & Queen, 2015). Symptomatic participants also show abnormal force curves that do not show the expected two peaks. Ankle OA limits flexion of the foot and knee and hip OA limits hip extension (Schmitt et al., 2015). Ankle OA patients use greater range of motion for their hips as well (Schmitt et al., 2015). These gait patterns could be used to diagnose OA in a low-resource setting and can be used to identify joint pain in our participants in Madagascar.

Force plates and acceleration can also be used to identify osteoarthritis (OA) (Queen et al., 2016). Less energy recovery during stance phase means that more work is being done by the muscle, hence, inducing fatigue. End-stage OA in the knee, hip, and ankle can slow down an individual’s walking speed (Queen et al., 2016). Energy recovery is nearly the same for individuals with end stage knee OA in comparison to those without knee OA. Energy recovery is less for individuals with hip and ankle OA in comparison to those with knee OA (Queen et al., 2016). Although these findings are only applicable to individuals with end stage OA, they can still be applied to the current study. We expect participants with joint pain to walk slower across our force platform than those with no pain. Hence, velocity could be used as another variable to help determine joint pain in participants in a low resource setting.
Plantar Fasciitis and Gait Deviation

Plantar fasciitis is a common injury often associated with heavy weight bearing or consistent microtrauma to the foot (Schepsis, Leach, & Gorzyca, 1991; Waclawski, Beach, Milne, Yacyshyn, & Dryden, 2015). It is important to look at plantar fasciitis because the Malagasy in Mandena travel on uneven surfaces and hills while carrying heavy loads of wood and crops.

Plantar fasciitis is the inflammation of the plantar aponeurosis which extends from the calcaneus to the metatarsals in the plantar region of the foot. The ailment causes pain in the plantar region of the foot while walking, which can inhibit daily activity. Risk factors of the disease include prolonged standing or jumping, pes planus (flat feet), and limited dorsiflexion of the foot (Waclawski et al., 2015). Hence, it is important to assess how foot morphology and foot biomechanics impact the onset of plantar foot pain in the Malagasy and see who is at most risk of injury.

Vertical ground reaction forces (the normal force that acts on the foot while walking) is altered in patients with plantar fasciitis (Phillips & McClinton, 2017). Hence, in addition to using a questionnaire, a force plate can be used in Mandena to assess the relationship between foot morphology and plantar fasciitis, typically found in one foot. There is moderate to strong evidence that there is decreased peak VGRF during the first third of stance in association with rear-foot strike (Phillips & McClinton, 2017). To compensate for this decreased force, the midfoot and forefoot contact the ground for a longer period of time, resulting in better pressure distribution in the foot and there is a decreased peak VGRF at the terminal stance (Phillips & McClinton, 2017). We will search for these patterns when subjects claim they have experience plantar foot pain.
The Double Burden of Disease in Madagascar

With the expansion of globalization, Madagascar is currently facing an epidemiological transition because of a rapid increase in life expectancy over the past 30 years (Masquelier, Waltisperger, Ralijaona, Pison, & Ravélo, 2014). Although the threat of communicable diseases has been decreasing, they still contribute to a large portion of disability-adjusted life years, especially outside of the capital city of Antananarivo (Wang et al., 2016). Paired with this decreased impact of infectious diseases, the prevalence of non-communicable diseases has been on the rise because of the increased life expectancy. As a result, the Malagasy now face a double burden of communicable and non-communicable diseases (Masquelier, Waltisperger, Ralijaona, Pison, & Ravélo, 2014).

Unfortunately, Madagascar lacks the financial infrastructure to sustain its health care system, leaving this double burden of disease unaddressed. In 2014, health expenditures only consisted of 3% of the nation’s GDP and there were only 0.16 physicians for every 1,000 Malagasy (“The World Factbook — Central Intelligence Agency,” 2017). This is alarming because 75.3% of the population in 2010 lived in poverty and were already struggling to address their health needs (“The World Factbook — Central Intelligence Agency,” 2017).

Health outcomes influence an individual’s ability to work and support their family/community. For instance, 71.1% of Madagascar is agricultural land and the agricultural industry consumes 25% of the nation’s GDP (“The World Factbook — Central Intelligence Agency,” 2017). Indeed, 80.4% of the employed population works in agriculture and 93% of the employed population is in poverty (United Nations Development Programme, 2015). As a result, morbidities that incapacitate the Malagasy, such as musculoskeletal disorders, can play a detrimental role in an individual’s physical and economic well-being.
In recent years, there has been a priority to address “poverty traps” in Africa through communicable diseases ("AIDS, Other Diseases Create Poverty Trap in Africa," 2006; Bloom & Canning, 2000; Bonds, Keenan, Rohani, & Sachs, 2010). By treating and preventing infectious diseases, a population would be economically productive and invest more in physical capital. Unfortunately, these discussions have not been had for noncommunicable diseases and the greater implications they have on a community beyond the patient.

**Summary of Goals**

Studies show a clear relationship between foot morphology and pathology. The effect of foot injuries, such as plantar fasciitis, and osteoarthritis can be detrimental to the well-being of the Malagasy and it is important to establish risk factors for these morbidities. By establishing these associations, we can contribute to preventative measures to improve musculoskeletal health outcomes and promote economic sustainability. The study aims to identify the relationship between foot morphology and pain, assess how this pain affects gait, and how these associations vary between males and females within the Malagasy population.

**Methods**

**Study Site and Subjects**

The study was conducted in Mandena, Madagascar, a small village near Marojejy National Park in the northeast region of the island. Participants were recruited at a community meeting led by the village president. Unshod participants (n = 89, male = 41, female = 48, average age = 39 years) of ages 18-60 were enrolled following informed consent. Prior foot surgery or a known neuromuscular disorder were the sole exclusion criteria.
Data Collection

We first obtained participants’ height, mass, and age from a related study (Appendix A). Working with translators, we collected self-report data on daily activity and foot and lower limb pain. The Revised Foot Function Index (FFI-R) is a validated self-report measure that assesses foot-function related to health (Budiman-Mak, Conrad, Stuck, & Matters, 2006). We used nine questions from the sixty-eight questions typically used in the FFI-R and used a three-tier ranking system for the index instead of the typical six (Appendix B). We altered some of the questions to make them more applicable to our participants’ lifestyles. For instance, one question on the official FFI-R asked about difficulty “climbing stairs.” Because there are very few stairs in Mandena and our participants hike hills to harvest crops, we changed the wording to “ascending hills.” Our modified FFI-R assessed pain, difficulty, and limitation of activities because of reported foot/ankle pain (total score = 27).
To test whether foot morphology plays a role in injury or force exertion, we measured ten standard foot and ankle characteristics (Figure 6). These measurements were standardized in a previous study and were shown to be significantly different between males and females in a shod population (Wunderlich & Cavanagh, 2001).

Figure 6. We recorded 10 foot shape measures: Ankle Height (2), Medial Malleolus Height (3), Hallux Height (8), Ankle Circumference (12), Instep Circumference (14), Ball of Foot Breadth (17), Ankle Length (18), Outside Ball of Foot Length (22), Bimalleolar Breadth (24), Foot Length (26) (Wunderlich & Cavanagh, 2001).

To test variation in vertical ground reaction forces, participants walked barefoot across a PASCO-PS-2142-PASPORT 2-Axis Force Platform at self-selected speeds while being videorecorded at 120 fps by two Casio EX-ZR1000 cameras (Figure 7). Participants walked on sandy terrain before and after approaching the force plate by at least three steps. The force platform recorded vertical ground reaction force at 1 kHz. We only recorded trials in which the entirety of a participant’s foot landed on the platform. The cameras were placed perpendicular to the platform and recorded videos of the full body and foot (Figure 8). Markers were placed on the hallux, navicular, calcaneus, and medial malleolus (Figure 8b).
The validity of normal VGRFs recorded through the PASCO Force Platform was confirmed by comparing it to a more sophisticated AMTI Force Plate prior to travel. The force platform was extensively calibrated in Mandena using a 0.9 kg water bottle and a 61.7 kg person. It reported an error of only 6.87% with the bottle and 2.86% with the person. Lastly, four western participants walked over the PASCO Force Platform in Mandena in addition to another PASCO platform in good condition in the United States after travel. The force platform in Mandena recorded similar force curves to the one in the United States, confirming its validity (Figure 9).

Figure 7. Experimental set-up in Mandena.

Figure 8. Screenshots of full-body (a) and foot (b) videos.
Figure 9. Screenshots of force curves recorded from the same participant in the United States (a) and in Madagascar (b) with different PASCO Force Platforms. Both force platforms recorded lower loading peaks than propulsive peaks, impact peaks, and undefined valleys at midstance. Differences in the two curves can be attributed to the different surfaces on which the participant walked on: hard wood (a) and sandy terrain (b).

Data Analysis

We assessed force peaks, impulse, contact time, loading rate, and unloading rate from participants’ force curves. Using DLTdv6, we digitized videos to record self-selected speeds of each participant (Hedrick, 2008) (Figure 10, Appendix C).
Figure 10. Markers placed through DLTdv6 tracked the participant’s body to record velocity.

Markers were also placed on each end of the middle of the force platform to calibrate the video. Self-selected speeds were calculated as change in distance over change in time (Appendix D). Velocity was recorded at 5 points throughout the video and the average of the values was recorded in the dataset.

Two sample T-tests compared foot shape, FFI-R Index (higher values indicate more pain), activity between males and females, and gait kinematics. Foot shape measures were standardized based on foot length and mass using Wunderlich and Cavanagh (2001)’s method to make comparisons for each of the 10 measurements between males and females within the Malagasy population.

ANOVA tests compared the force curves of participants who reported foot/ankle/knee pain on one leg, both legs, and no legs. We used a strict exclusion criterion for our force curve analysis. Participants who demonstrated forefoot striking or had significant lower limb pain elsewhere besides the knee/foot/ankle were excluded from this analysis. Trials that exhibited participant acceleration while walking across the force platform were also excluded.
Using MATLAB, we collected the contact time of the stance phase (length of the curve on the x-axis), impulse (area under the curve), impact peak, timing of the impact peak, loading peak, loading rate, minimal value at midstance, propulsive peak, and unloading rate. Loading and unloading rates were calculated as the average change of force over time. These measurements were standardized by the participant’s body weight (Figure 11).

![Figure 11. Labelled force curve characteristics calculated for the study.](image)

Values of self-selected speed and force curve characteristics were averaged for each participant. As a result, we assessed the force curves of 43 participants (male = 18, female = 25, average age = 38 years). Two sample T-tests compared the force curves of participants who reported pain versus participants who did not report pain. Force curves were also compared between males and females using this method.
Results

Foot Morphology

Like Wunderlich and Cavanagh (2001), we compared our nine foot measurements between males and females in the Malagasy sample after standardizing for foot size. We only found three characteristics that were significantly different between males and females in our sample. Interestingly, all three characteristics were associated with the ankle: height, circumference, and length. In contrast, Wunderlich and Cavanagh found nine differences in their western sample, highlighting the similar foot characteristics between males and females in our Malagasy sample when standardized for foot length (Table 1).

<table>
<thead>
<tr>
<th>Foot Characteristic</th>
<th>Greater Average</th>
<th>Male Mean (%)</th>
<th>Female Mean (%)</th>
<th>P Value</th>
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<tr>
<td></td>
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<td>$\bar{x}$ (%)</td>
<td>$s$ (%)</td>
<td>$\bar{x}$ (%)</td>
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<td>Instep Circumference</td>
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Table 1. Comparison of foot characteristics standardized by foot length between males (n = 41) and females (n=48) in Mandena.
However, when we standardized our measures by mass, we found more differences between the males and females in our sample. Overall, these results show that the females in our sample had smaller feet than the males (Table 2).

<table>
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<tr>
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<th>Female Mean (%)</th>
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<td></td>
<td>$\bar{x}$ (%)</td>
<td>s (%)</td>
<td>$\bar{x}$ (%)</td>
</tr>
<tr>
<td>Foot Length</td>
<td>Male</td>
<td>45.31</td>
<td>4.95</td>
<td>41.42</td>
</tr>
<tr>
<td>Hallux Height</td>
<td>Male</td>
<td>3.53</td>
<td>0.47</td>
<td>3.11</td>
</tr>
<tr>
<td>Ankle Height</td>
<td>Male</td>
<td>20.31</td>
<td>2.60</td>
<td>17.68</td>
</tr>
<tr>
<td>Ball of Foot Breadth</td>
<td>Male</td>
<td>17.6657</td>
<td>1.95</td>
<td>16.1084</td>
</tr>
<tr>
<td>Outside Ball of Foot Length</td>
<td>Male</td>
<td>30.95</td>
<td>3.67</td>
<td>28.69</td>
</tr>
<tr>
<td>Medial Malleolus Height</td>
<td>Male</td>
<td>15.3691</td>
<td>1.77</td>
<td>13.7655</td>
</tr>
<tr>
<td>Ankle Length</td>
<td>Male</td>
<td>12.3123</td>
<td>1.47</td>
<td>12.1215</td>
</tr>
<tr>
<td>Bimalleolar Breadth</td>
<td>Male</td>
<td>12.4494</td>
<td>1.50</td>
<td>11.26</td>
</tr>
<tr>
<td>Ankle Circumference</td>
<td>Female</td>
<td>36.99</td>
<td>3.45</td>
<td>37.01</td>
</tr>
<tr>
<td>Instep Circumference</td>
<td>Male</td>
<td>45.0857</td>
<td>4.39</td>
<td>41.7802</td>
</tr>
</tbody>
</table>

**Table 2.** Comparison of foot characteristics standardized by mass between males (n=40) and females (n=43) in Mandena. Only for medial malleolus height was n=42 for females.
Figure 12. Foot/Ankle shape measures standardized by mass in which females were significantly smaller than males are marked in red. Differences in green measures were nonsignificant. Image adapted from (Wunderlich & Cavanagh, 2001).

Clinical Data

It was important to standardize our footshape measures by mass because our study showed that females had significantly higher BMIs ($n = 43, \bar{x} = 23.1634, s = 3.056$) than males ($n = 40, \bar{x} = 20.6451, s = 2.0563$), ($p = 0.0001$) (Figure 13).
Additionally, our survey collected data on our sample’s daily activity. Our participants self-reported walking an average of 5.24 km a day and spending an average of 7.85 hours a day on their feet. Females reported spending significantly more hours on their feet (n = 48, $\bar{x} = 8.25$, $s = 1.2463$) than males (n = 41, $\bar{x} = 7.39$, $s = 1.922$), ($p = 0.0184$) (Figure 14). Lastly, 79.7% (n=71) of participants reported that they primarily walk on dirt roads on a daily basis.

Figure 13. Female and male BMI comparison.

Figure 14. Comparison of hours spent on feet between males and females.
Most of our participants reported a variation of lower limb pain (Figure 1). 11% (n=8) of our participants reported that they have visited the doctor for their pain and 12.3% (n=9) mentioned that they have treated their pain with a massage.

Figure 15. Number of participants reporting pain in specific locations of the leg.

Lastly, we used the modified FFI-R to assess the severity of self-reported foot/ankle pain and to what extent it made activities difficult and made participants limit their activities. We wanted to validate the use of the index in our study and compared our scores between participants who did not report foot/ankle pain (n=16) to those who did (n=47). Those who reported pain ($\bar{x} = 14,809, s = 3.905$) reported significantly higher total index scores than those who did not report pain ($\bar{x} = 11.56, s = 2.16$), ($p=0.0009$) (Figure 16).
Figure 16. Comparison of total FFI-R index score between participants that did and did not report foot/ankle pain.

The reported index scores show that there is some pain in this population, but it is not severe, nor does it make activities difficult for our population. However, our sample reported that their foot/ankle pain results in moderate activity limitation (Figure 17). Females (n = 34, $\bar{x} = 14.9706$, $s = 3.6472$) reported higher FFI-R scores than males (n = 29, $\bar{x} = 12.8276$, $s = 3.7136$), ($p = 0.029$) (Figure 18). All other comparisons of FFI-R were non-significant.
Figure 17. Average index score proportion for each subsection of the Modified FFI-R (n=63).

Figure 18. Comparison of total FFI-R Index scores between males and females.
Gait Kinematics

After using a strict exclusion criterion, we compared the force curves of our participants (n=43). Participants were between the ages of 18-60 and had an average BMI of 21.7.

Participants were asked to walk at self-selected speeds across our force platform. The average velocity of the participants was 1.03 m/s (s = 0.18 m/s). Participants also varied in different force curve parameters (Table 3 – “Prop. BW” = Proportion of Body Weight).

<table>
<thead>
<tr>
<th>Force Curve Parameter</th>
<th>Mean</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Time (s)</td>
<td>0.750</td>
<td>0.110</td>
</tr>
<tr>
<td>Impulse (Prop. BW-second)</td>
<td>0.562</td>
<td>0.065</td>
</tr>
<tr>
<td>Impact Peak (Prop. BW)</td>
<td>0.509</td>
<td>0.178</td>
</tr>
<tr>
<td>Impact Time (s)</td>
<td>0.056</td>
<td>0.033</td>
</tr>
<tr>
<td>Loading Peak (Prop. BW)</td>
<td>1.085</td>
<td>0.123</td>
</tr>
<tr>
<td>Loading Rate (Prop. BW/s)</td>
<td>6.253</td>
<td>2.074</td>
</tr>
<tr>
<td>Midstance (Prop. BW)</td>
<td>0.793</td>
<td>0.089</td>
</tr>
<tr>
<td>Propulsive Peak (Prop. BW)</td>
<td>1.007</td>
<td>0.060</td>
</tr>
<tr>
<td>Unloading Rate (Prop. BW/s)</td>
<td>4.785</td>
<td>1.031</td>
</tr>
</tbody>
</table>

Table 3. Summary statistics of force curve parameters for selected participants. n=43 for all force curve parameters except Impact Peak and Impact Time (n = 30).

As each participant was asked to walk across the force platform at self-selected speeds, velocity varied between participants. Hence, we assessed how self-selected speed was related to each force curve parameter in our sample (Table 4, Figure 19).

<table>
<thead>
<tr>
<th>Force Curve Parameter</th>
<th>Linear Relationship</th>
<th>Adjusted R²</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Time (s)</td>
<td>Negative</td>
<td>0.706</td>
<td>&lt;&lt;0.001</td>
</tr>
<tr>
<td>Impulse (Prop. BW-second)</td>
<td>Negative</td>
<td>0.591</td>
<td>&lt;&lt;0.001</td>
</tr>
<tr>
<td>Impact Peak (Prop. BW)</td>
<td>Positive</td>
<td>0.532</td>
<td>&lt;&lt;0.001</td>
</tr>
<tr>
<td>Impact Time (s)</td>
<td>Negative</td>
<td>0.004</td>
<td>0.2967</td>
</tr>
<tr>
<td>Loading Peak (Prop. BW)</td>
<td>Positive</td>
<td>0.519</td>
<td>&lt;&lt;0.001</td>
</tr>
<tr>
<td>Loading Rate (Prop. BW/s)</td>
<td>Positive</td>
<td>0.714</td>
<td>&lt;&lt;0.001</td>
</tr>
<tr>
<td>Midstance (Prop. BW)</td>
<td>Negative</td>
<td>0.744</td>
<td>&lt;&lt;0.001</td>
</tr>
<tr>
<td>Propulsive Peak (Prop. BW)</td>
<td>Positive</td>
<td>0.160</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>Unloading Rate (Prop. BW/s)</td>
<td>Positive</td>
<td>0.744</td>
<td>&lt;&lt;0.001</td>
</tr>
</tbody>
</table>

Table 4. Linear relationship between participants’ velocity and force curve parameters.
As shown in **Table 4**, there is a very strong linear relationship between participant velocity and force curve parameters. The relationship between velocity and propulsive peak, although weaker, is still significant. The relationship between impact time and velocity is not significant.

![Figure 19](image)

**Figure 19.** Linear relationship between our sample’s force parameters and self-selected speed. Row 1 (L to R): Contact Time, Impulse, Impact Peak. Row 2 (L to R): Impact Time, Loading Peak, Loading Rate. Row 3 (L to R): Midstance, Propulsive Peak, Unloading Rate.

Because self-selected speed heavily influenced most of the force curve parameters, it was important to assess if there were significant differences in velocity between our comparative groups. We conducted 3 comparisons in force curve parameters: 1) Between participants that
reported pain on both legs, one leg, and no legs, 2) between participants that reported pain on at least one leg vs. no pain, 3) between male and female participants. ANOVA and pairwise t-tests showed that differences in velocity between groups was not significant for all three comparisons (Table 5).

Table 5. Average velocity of comparative groups.

Our results showed that there was no significant difference in self-selected speed between our comparative groups. Hence, we continued ANOVA and t-test analyses of force parameters between groups without the consideration of self-selected speed.

**Pain on Legs Comparison**

We conducted an ANOVA test to compare force curve parameters between participants with pain on both legs, pain on one leg, and no pain in their legs. Our results showed that there was no significant difference in the examined force curve parameters between the comparative groups (Table 6, Figure 20, Figure 21).
<table>
<thead>
<tr>
<th>Force Curve Parameter</th>
<th>Location of Pain</th>
<th>Mean</th>
<th>Standard Deviation</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Time (s)</td>
<td>Both Legs</td>
<td>0.790</td>
<td>0.127</td>
<td>0.413</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>0.735</td>
<td>0.106</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>0.775</td>
<td>0.110</td>
<td></td>
</tr>
<tr>
<td>Impulse (Prop. BW-second)</td>
<td>Both Legs</td>
<td>0.582</td>
<td>0.031</td>
<td>0.523</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>0.554</td>
<td>0.070</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>0.574</td>
<td>0.071</td>
<td></td>
</tr>
<tr>
<td>Loading Peak (Prop. BW)</td>
<td>Both Legs</td>
<td>1.071</td>
<td>0.140</td>
<td>0.756</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>1.094</td>
<td>0.126</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>1.059</td>
<td>0.104</td>
<td></td>
</tr>
<tr>
<td>Loading Rate (Prop. BW/s)</td>
<td>Both Legs</td>
<td>6.369</td>
<td>2.527</td>
<td>0.818</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>6.336</td>
<td>2.115</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>5.789</td>
<td>1.596</td>
<td></td>
</tr>
<tr>
<td>Midstance (Prop. BW)</td>
<td>Both Legs</td>
<td>0.799</td>
<td>0.069</td>
<td>0.934</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>0.790</td>
<td>0.099</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>0.801</td>
<td>0.072</td>
<td></td>
</tr>
<tr>
<td>Propulsive Peak (Prop. BW)</td>
<td>Both Legs</td>
<td>1.004</td>
<td>0.077</td>
<td>0.717</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>1.012</td>
<td>0.062</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>0.991</td>
<td>0.031</td>
<td></td>
</tr>
<tr>
<td>Unloading Rate (Prop. BW/s)</td>
<td>Both Legs</td>
<td>4.698</td>
<td>1.100</td>
<td>0.576</td>
</tr>
<tr>
<td></td>
<td>One Leg</td>
<td>4.890</td>
<td>1.084</td>
<td></td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>4.438</td>
<td>0.746</td>
<td></td>
</tr>
</tbody>
</table>

**Table 6.** Comparison of force curve parameters between participants whom reported pain on both legs, pain on one leg, and no pain.
Figure 20. Comparison of force curve parameters between participants whom reported pain on both legs, pain on one leg, and no pain. Error bars represent standard deviation.

Figure 21. Comparison of loading rate and unloading rate between participants whom reported pain on both legs, pain on one leg, and no pain. Error bars represent standard deviation.

Our results imply that rather than patterns of foot/ankle/knee pain, velocity and other features could drive variation in force curve parameters among our participants. Hence, we
assessed how self-selected speed in each of the three comparative groups affected variability in force curve parameters (Figure 22).

Figure 22. Linear relationship between self-selected speed (m/s) and force curve parameters when comparing participants who reported pain on both legs (Red), pain on one leg (Blue), and no pain (Green). Row 1 (L to R): Contact Time, Impulse, Loading Peak. Row 2 (L to R): Loading Rate, Midstance, Propulsive Peak. Row 3: Unloading Rate.

The linear relationship between self-selected speed and force curve parameters follow similar trends between our comparative samples. The slopes of these relationships are in similar magnitude as well, except for normalized propulsive peak. The slope for normalized propulsive
peak affected by self-selected speed was steeper for participants with pain on both legs (Figure 22).

**Pain vs. No Pain Comparison**

We conducted two-sided t-tests between participants who reported pain in the foot/ankle/knee and those who reported no pain. Although we observed some differences in force curve parameters between the two groups, none of these comparisons were significant (Table 7, Figure 23, Figure 24).

<table>
<thead>
<tr>
<th>Force Curve Parameter</th>
<th>Explanatory Variable</th>
<th>Mean</th>
<th>Standard Deviation</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Time (s)</td>
<td>Pain</td>
<td>0.745</td>
<td>0.111</td>
<td>0.547</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>0.775</td>
<td>0.110</td>
<td></td>
</tr>
<tr>
<td>Impulse (Prop. BW-second)</td>
<td>Pain</td>
<td>0.559</td>
<td>0.065</td>
<td>0.639</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>0.574</td>
<td>0.071</td>
<td></td>
</tr>
<tr>
<td>Impact Peak (Prop. BW)</td>
<td>Pain</td>
<td>0.517</td>
<td>0.194</td>
<td>0.476</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>0.476</td>
<td>0.088</td>
<td></td>
</tr>
<tr>
<td>Impact Time (s)</td>
<td>Pain</td>
<td>0.056</td>
<td>0.036</td>
<td>0.992</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>0.056</td>
<td>0.021</td>
<td></td>
</tr>
<tr>
<td>Loading Peak (Prop. BW)</td>
<td>Pain</td>
<td>1.090</td>
<td>0.127</td>
<td>0.511</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>1.059</td>
<td>0.104</td>
<td></td>
</tr>
<tr>
<td>Loading Rate (Prop. BW/s)</td>
<td>Pain</td>
<td>6.343</td>
<td>2.162</td>
<td>0.461</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>5.789</td>
<td>1.596</td>
<td></td>
</tr>
<tr>
<td>Midstance (Prop. BW)</td>
<td>Pain</td>
<td>0.791</td>
<td>0.093</td>
<td>0.758</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>0.801</td>
<td>0.072</td>
<td></td>
</tr>
<tr>
<td>Propulsive Peak (Prop. BW)</td>
<td>Pain</td>
<td>1.010</td>
<td>0.064</td>
<td>0.278</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>0.991</td>
<td>0.031</td>
<td></td>
</tr>
<tr>
<td>Unloading Rate (Prop. BW/s)</td>
<td>Pain</td>
<td>4.853</td>
<td>1.073</td>
<td>0.260</td>
</tr>
<tr>
<td></td>
<td>No Pain</td>
<td>4.438</td>
<td>0.746</td>
<td></td>
</tr>
</tbody>
</table>

**Table 7.** Comparison of force curve parameters between participants who reported/did not report pain in their foot/ankle/knee.
Figure 23. Averages of force curve parameters between participants who reported/did not report foot/ankle/knee pain. Error bars represent standard deviation.

Figure 24. Comparison of loading rate and unloading rate between participants who reported/did not report foot/ankle/knee pain. Error bars represent standard deviation.
Like our previous comparison, differences in force curve parameters between participants who reported pain and those who did not were not significant. Hence, we were also interested in examining the difference between self-selected speed and force curve parameters between our comparative groups (Figure 25).

Figure 25. Linear relationship between self-selected speed (m/s) and force curve parameters when comparing participants who reported pain (Teal) and no pain (Red). Row 1 (L to R): Contact Time, Impulse, Impact Peak. Row 2 (L to R): Impact Time, Loading Peak, Loading Rate. Row 3 (L to R): Midstance, Propulsive Peak, Unloading Rate.
Interestingly, the relationship between velocity and force curve parameters are similar in slope and magnitude between the comparative groups.

**Male vs. Female Comparison**

We conducted two-sided t-tests to compare force curve parameters between males and females in our sample. All comparisons were nonsignificant (Table 8, Figure 26, Figure 27).

<table>
<thead>
<tr>
<th>Force Curve Parameter</th>
<th>Gender</th>
<th>Mean</th>
<th>Standard Deviation</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Time (s)</td>
<td>Female</td>
<td>0.739</td>
<td>0.108</td>
<td>0.431</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>0.767</td>
<td>0.115</td>
<td></td>
</tr>
<tr>
<td>Impulse (Prop. BW-second)</td>
<td>Female</td>
<td>0.551</td>
<td>0.065</td>
<td>0.639</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>0.576</td>
<td>0.065</td>
<td></td>
</tr>
<tr>
<td>Impact Peak (Prop. BW)</td>
<td>Female</td>
<td>0.520</td>
<td>0.187</td>
<td>0.745</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>0.498</td>
<td>0.173</td>
<td></td>
</tr>
<tr>
<td>Impact Time (s)</td>
<td>Female</td>
<td>0.059</td>
<td>0.043</td>
<td>0.588</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>0.052</td>
<td>0.021</td>
<td></td>
</tr>
<tr>
<td>Loading Peak (Prop. BW)</td>
<td>Female</td>
<td>1.074</td>
<td>0.098</td>
<td>0.537</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>1.100</td>
<td>0.152</td>
<td></td>
</tr>
<tr>
<td>Loading Rate (Prop. BW/s)</td>
<td>Female</td>
<td>5.898</td>
<td>1.740</td>
<td>0.224</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>6.744</td>
<td>2.431</td>
<td></td>
</tr>
<tr>
<td>Midstance (Prop. BW)</td>
<td>Female</td>
<td>0.800</td>
<td>0.077</td>
<td>0.567</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>0.783</td>
<td>0.105</td>
<td></td>
</tr>
<tr>
<td>Propulsive Peak (Prop. BW)</td>
<td>Female</td>
<td>0.998</td>
<td>0.050</td>
<td>0.285</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>1.020</td>
<td>0.072</td>
<td></td>
</tr>
<tr>
<td>Unloading Rate (Prop. BW/s)</td>
<td>Female</td>
<td>4.760</td>
<td>0.864</td>
<td>0.865</td>
</tr>
<tr>
<td></td>
<td>Male</td>
<td>4.820</td>
<td>1.253</td>
<td></td>
</tr>
</tbody>
</table>

**Table 8.** Comparison of force curve parameters between males and females.
Figure 26. Averages of force curve parameters compared between males and females. Error bars represent standard deviation.

Figure 27. Comparison of loading and unloading rate between males and females. Error bars represent standard deviation.
We were also interested in examining the relationship between self-selected speed and force curve parameters between males and females (Figure 28).

**Figure 28.** Linear relationship between self-selected speed (m/s) and force curve parameters when comparing male (Teal) and female (Red) participants. Row 1 (L to R): Contact Time, Impulse, Impact Peak. Row 2 (L to R): Impact Time, Loading Peak, Loading Rate. Row 3 (L to R): Midstance, Propulsive Peak, Unloading Rate.

Males had longer contact time and higher normalized impulses than females at the same velocity. On the other hand, females had higher normalized impact peaks than males at the same velocity.
speed. Males also exhibited higher positive slopes than females for the relationship between loading peak, loading rate, propulsive peak, and unloading rate to velocity. Males had negative slopes of greater magnitude when comparing change in midstance peak to velocity.

**Discussion**

When standardized by foot length, female and male feet in our unshod sample in Madagascar were more similar to each other (only 3 statistically significant differences) than were western shod feet examined between males and females by Wunderlich and Cavanagh (8 statistically significant differences). The only significant differences we identified were in the ankle, in which female ankles were larger than those of males. This larger size could be associated with the higher BMIs we observed in females. Future research should specifically examine ankle pain and mobility in this population and its effects on daily activity.

Male and female feet were more similar in our Malagasy population than the western sample in Wunderlich and Cavanagh’s study when using the same standardization method. This variation in differences between the two sexes can be attributed to modern footwear. It is possible that males and females wear different shoewear that influence the shape of their feet differently in the western sample. For instance, women shoewear are often more constrictive than shoewear for men in the West, which can contribute to smaller foot size (Hoffman, 1905; D’Août et al., 2009). In contrast, the general health survey we conducted in Mandena showed that our participants seldom wore close-toed shoes. This could be a reason why female and male feet were more similar in Madagascar than in the West when standardized for foot length.

When standardized for mass, female feet were shorter and narrower in our Malagasy sample. This smaller size of the foot would limit the surface area over which force could be
distributed, potentially increasing the strain put upon the foot and ankle and increasing the risk of developing pathology that might cause pain and mobility limitations. This could be one of the reasons why females reported higher amounts of pain through the Modified FFI-R in comparison to the males in our sample. Moreover, females reported spending more time on their feet than males, increasing the daily amount of stress put on their feet and the risk of pathology.

The observed correlation between self-selected speed (velocity [m/s]) and the nine force curve parameters further validated the use of the PASCO force platform and the digitization required to record self-selected speed. Increased velocity would result in the foot spending less time on the ground during stance, hence, having a negative relationship with contact time and impulse. As expected, velocity also had a positive linear relationship with the peaks associated with the force curve in addition to the loading and unloading rates. With increased speed, we expect individuals to make sharper and more pronounced contact with their feet during stance, but for shorter amounts of time. Hence, the transition from loading response to terminal stance is rapid, with less force being applied during midstance, which we also examined through the negative linear relationship between velocity and midstance peaks.

Although comparison of force curve parameters between individuals who reported pain on both legs, pain on one leg, and no pain were insignificant, there is value in assessing the relationship between velocity and force curve parameters between the comparative groups. Although most relationships were similar, it was interesting that participants who reported pain on both legs exhibited a stronger positive relationship between self-selected speed and propulsive peak. It is possible that this may have been a method to compensate for heel pain through exaggerated foot plantarflexion and more force being applied during propulsion. However, it should be noted that this interpretation is being made with a sample of 7 participants and future
studies should assess this relationship with a greater sample size of individuals that report pain on both legs.

The comparison of the linear relationship of velocity and our selected force curve parameters between males and females showed that males had a greater magnitude of slope for 5 of the relationships. As mentioned earlier, females reported significantly higher total scores on the FFI-R, hence, having higher self-reports of pain. This pain may have prevented our female samples from spending more time during stance and from applying more force. For instance, our results showed that at the same speed, males spent longer time in stance and had greater impulse.

We did not examine any significant difference in force curve parameters between comparative groups when foot/ankle/knee pain was used as an explanatory variable. However, when comparing the relationship between velocity and force curve parameters between our comparative groups, notable differences (although they were not tested statistically), were observed. Velocity, along with other features, could influence force curve parameters more than pain. Factors such as foot shape, joint flexion, and pressure distribution could also affect force curve parameters and should be examined in future investigations.

This study had room for improvement. The force platform we used in Mandena was not validated with a more sophisticated force plate, such as an AMTI; however, it was still calibrated and detected the same trends in force curves as did a similar platform back in the United States. The amount of trials between participants also varied because of how difficult it was to land one foot entirely on the force plate. We had individuals needing to walk nearly 15 times and did not want to take up too much of their time; hence, we only collected one of two trials from them. Lastly, participants who reported pain on one leg had the option to land on the force platform with their pathologic leg, or their leg with no pain. Although this data was recorded during our
study, we did not include it in our analysis because we calculated the mean from all trials for a participant who landed on the platform with either leg.

However, the study had many notable strengths. It was the first of its kind to collect biomechanical data through videos and surveys in Mandena. A large enough sample size was obtained during the short stay in the village to make valuable comparisons between males and females as well as individuals who reported/did not report foot and leg pain. This study is an important first step towards addressing musculoskeletal health in Mandena and low-resource communities alike.

**Conclusion**

Our results showed that females are more susceptible to foot/ankle and leg pain than males. Interventions that address the risk factors we identified through this study can not only decrease the double burden of disease in the region, but also promote economic productivity by reducing the risk of musculoskeletal disorders.

Interventions that reduce time spent standing while working could be valuable to address foot and ankle pain in women. Being able to sit while conducting agricultural and domestic duties, as well as division of labor within the household, could encourage women to spend less time on their feet and reduce the onset of pain in the foot/ankle. Although this may seem to promote a sedentary lifestyle, it is important to note that the Malagasy are active individuals that travel far by feet and can afford to sit down more. A shoe could additionally be developed for this population, however, this type of shoewear may be a cultural deviation and may not be adopted by the Malagasy. Nevertheless, our results showed that a handful of our participants wore hiking shoes and soccer cleats and developing a shoe that could address sources of pain while looking like shoes that would be seen in Mandena may be the most feasible solution to
address musculoskeletal disorders, especially in women. Once future studies further assess the relationship the effects of pain, velocity, and foot shape on joint angles and force curves, more insight can be used to develop a more sustainable solution to address musculoskeletal pain in low-resource settings.
References


*Journal of Anatomy, 96*(Pt 4), 489–494.1.


Appendix A: General Health Survey

General Health and Foot Function Survey Questions

**General Health Survey**

1. Name ______________ ID______________

2. Do you live in Mandena?
   
   Yes   No
   
   If no, where do you live? ________________________________

3. Weight (kg) ______________

4. Height (m) ______________

5. Age (years) ____________

6. Gender ______________

7. What is your occupation(s)?

________________________________________________________________________

8. How often do you wear the following footwear? How old were you when you started wearing it?

<table>
<thead>
<tr>
<th>Usage (Daily, Weekly, Monthly, less than once a month)</th>
<th>BP</th>
<th>Pulse</th>
<th>Temperature</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age When Starting to Wear Footwear (0-10, 11-20, 21-30, 31-40, 41-50, 50+ Years)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

9. Have you ever had surgery on your feet?
   
   Yes   No

10. When walking, do you have trouble keeping your balance or walking in a straight line because of problems unrelated to your feet?
    
    Yes   No
Appendix B: Foot Function Survey

ID ____________________________

1. How much distance do you walk each day? __________

2. On what type of surface do you walk on each day
   Grass    Sand    Asphalt

3. How long are you standing or walking while working at home or outside each day? __________

4. Do you lift, transfer, and carry heavy loads while working?
   Yes    No

5. Have you ever dropped these loads onto your foot?
   Yes    No

6. If yes, how many times has this happened in the past month?
   1 or 2 times    Once a week    On most days    Daily

7. Do you have any pain, aches or injuries to the foot?
   Yes    No
   If yes, which foot? (Side:   Left   Right)
   Can you circle where the pain is?

8. Have you seen a doctor for your foot pain?
   Yes    No
Modified Revised Foot Function Index (FFI-R)

### Pain
During the past week, how severe was your foot pain:
- When you first stood without shoes in the morning
- At the end of the day
- While working at home or outside

<table>
<thead>
<tr>
<th></th>
<th>No Pain (1)</th>
<th>Mild Pain (2)</th>
<th>Severe Pain (3)</th>
</tr>
</thead>
<tbody>
<tr>
<td>When you first stood without shoes in the morning</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>At the end of the day</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>While working at home or outside</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

### Difficulty
During the past week, how much difficulty did your foot problems cause you:
- Walking on uneven surfaces (sand, grass)
- Ascending hills
- Getting out of a chair
- Carrying weight

<table>
<thead>
<tr>
<th></th>
<th>No Difficulty (1)</th>
<th>Mild Difficulty (2)</th>
<th>Severe Difficulty (3)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking on uneven surfaces (sand, grass)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ascending hills</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Getting out of a chair</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Carrying weight</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

### Activity Limitation
During the past week, how much of the time did you:
- Limit outdoor work activities because of foot problems
- Limit leisure/sport activities because of foot problems

<table>
<thead>
<tr>
<th></th>
<th>None of the time (1)</th>
<th>Some of the time (2)</th>
<th>Most of the time (3)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Limit outdoor work activities because of foot problems</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Limit leisure/sport activities because of foot problems</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
Appendix C: MATLAB Code to Analyze Force Curves

%% Presets
CutOffFreq = 60;
RecordingSpeed = 1000;
%% Actual Program
[File Path] = uigetfile('*.*');
Data = dlmread([Path File], '	', 13);
Time = Data(:, 1);
zForce = Data(:, 6);
clear Data
format long g
%% Filter
FilteredZ = real(lowpassFourier(Time, zForce, CutOffFreq));
%% Plot for Cutoffs
plot(FilteredZ)
[xbounds, ybounds] = ginput(2);
start = xbounds(1);
stop = xbounds(2);
start = fix(start);
stop = fix(stop);
close
N = input('How many ranges of interest?
');
W = input('What is their weight in kilograms?
');
FinalZ = FilteredZ(start:stop);
Time = 1:(stop - start) / RecordingSpeed;
%% Calculate Contact Time, Impulse, and Maximum
DataSet = cell(N, 3);
for ii = 1:(N)
plot(FilteredZ(start:stop));
[xbounds, ybounds] = ginput(2);
start2 = xbounds(1);
stop2 = xbounds(2);
start2 = fix(start2);
stop2 = fix(stop2);
close
FinalZ = FilteredZ(start2 + start:stop2 + start);

ZPeak = max(FinalZ);
ZPeakNorm = (ZPeak) / (W * 9.806);
ZPeakX = find(FilteredZ == ZPeak);
ZPeakX = ZPeakX / RecordingSpeed;
ImpactTime = (ZPeakX - (start / RecordingSpeed)) / ((stop - start) / RecordingSpeed);
LoadingRate = ZPeakNorm / (ZPeakX - (start / RecordingSpeed));
UnloadingRate = ZPeakNorm / ((stop / RecordingSpeed) - ZPeakX);
ZMin = min(FinalZ);
ZMinNorm = (ZMin) / (W * 9.806);
Impulse = trapz(0: (stop2 - start2), FinalZ). / RecordingSpeed;
ImpulseNorm = (Impulse) / (W * 9.806);
ContactTime = length(FinalZ)./RecordingSpeed;

DataSet((ii+1),1) = {ZPeakNorm};
DataSet((ii+1),2) = {ZPeakX};
DataSet((ii+1),3) = {ImpactTime};
DataSet((ii+1),4) = {LoadingRate};
DataSet((ii+1),5) = {UnloadingRate};
DataSet((ii+1),6) = {ZMinNorm};
DataSet((ii+1),7) = {ImpulseNorm};
DataSet((ii+1),8) = {ContactTime};
end

DataSet(1,1) = {'ZPeakNorm'};
DataSet(1,2) = {'ZPeakX'};
DataSet(1,3) = {'ImpactTime'};
DataSet(1,4) = {'LoadingRate'};
DataSet(1,5) = {'UnloadingRate'};
DataSet(1,6) = {'ZMinNorm'};
DataSet(1,7) = {'ImpulseNorm'};
DataSet(1,8) = {'Contact Time'};

%% Create Excel File
fid = fopen([File(1:end-4) '_PASCO.csv'], 'w');
fprintf(fid, '%s,
', DataSet{1,1:end-1});
fprintf(fid, '%s
', DataSet{1,end});
fclose(fid);
dlmwrite([File(1:end-4) '_PASCO.csv'], DataSet(2,:), '-append');
Appendix D: MATLAB Code to Calculate Velocity

%% Presets
%Force Plate length is .364 meters
%Recording speed was set to 120 frames per second
ForcePlate = .364;
RecordingSpeed = 120;

%% Actual Program
[File Path] = uigetfile('*.csv');
Data = csvread([File Path],1,0);
Xcal1 = Data(:,1);
Xcal2 = Data(:,3);
pixelpermeter = abs(max(Xcal2)-max(Xcal1))/ForcePlate;
Path = Data(:,5);
clear Data
format long g

%% Selecting 5 Instances for Velocity
DataSet = cell(6,1);
for ii = 1:(5)
plot(Path);

[xbounds,ybounds]=ginput(2);
timeinitial=xbounds(1);
timefinal=xbounds(2);
start=ybounds(1);
stop=ybounds(2);
close

DeltaX = (stop-start)/pixelpermeter;
DeltaTime = (timefinal-timeinitial)/RecordingSpeed;
Velocity = DeltaX/DeltaTime;

DataSet((ii+1),1) = {Velocity};
end
DataSet(1,1) = {'Velocity'};

%% Averaging the Velocities
numericcells = DataSet(2:6,1);
numericvectors = cell2mat(numericcells);
AverageVelocity = mean(numericvectors);
DataSet(2,2) = {AverageVelocity};
DataSet(1,2) = {'AvgVel'};