TOWARDS AN UNDERSTANDING OF PROLONGED PRONATION: IMPLICATIONS FOR MEDIAL TIBIAL STRESS SYNDROME AND ACHILLES TENDINOPATHY

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DISSERTATION ABSTRACT

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Title: Towards an Understanding of Prolonged Pronation: Implications for Medial Tibial Stress Syndrome and Achilles Tendinopathy

Epidemiologic data suggest 25% to 75% of all runners experience an overuse injury each year. Commonly cited biomechanical factors related to overuse injuries such as Achilles tendinopathy or medial tibial stress syndrome include excessive amounts or velocities of foot pronation. However, there is conflicting evidence in the literature supporting this theory. An alternative hypothesis suggests it is not necessarily the amount or velocity of pronation which is important for injury development; rather it is the duration the foot remains in a pronated position throughout stance that is the important variable.

This project examined this hypothesis by first identifying biomechanical markers of prolonged pronation. Second, it assessed whether individuals currently symptomatic with injuries typically attributed to excessive pronation instead demonstrate the biomechanical markers of prolonged pronation. Finally, musculoskeletal modeling techniques were used to examine musculotendinous kinematics in injured and healthy runners, as well as healthy runners with prolonged pronation.

The results suggest the two most robust measures for identifying individuals with prolonged pronation are the period of pronation and the eversion of the rear foot at heel

off. Individuals with prolonged pronation can also be identified with a set of clinically feasible measures including higher standing tibia varus angles, reduced static hip internal rotation range of motion, and increased hip internal rotation during stance phase. Finally, individuals with prolonged pronation display a more medially located center of pressure trajectory during stance. Compared to healthy controls, individuals currently symptomatic with Achilles tendinopathy or medial tibial stress syndrome did not differ in the amount or velocity of pronation. However, they did demonstrate the biomechanical markers of prolonged pronation. Injured individuals also demonstrated greater average musculotendinous percent elongation than healthy controls, especially through mid and late stance. Currently healthy individuals demonstrating prolonged pronation exhibited musculotendinous percent elongations intermediate to the healthy and injured groups.

As a whole, the results from this study suggest prolonged pronation may play a role in the development of common overuse running injuries. It is suggested future studies on injury mechanisms consider pronation duration as an important variable to examine.

This dissertation includes unpublished co-authored material.

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CHAPTER I

INTRODUCTION

Running Injury Incidences and Associated Costs

Both the American College of Sports Medicine's "Exercise is Medicine" campaign and the Federal Government's Healthy People 2020 initiative promote the importance of aerobic exercise in maintaining overall health. One way many Americans choose to obtain this exercise is through participation in running. The 2011 Sporting Goods Manufacturer's Association survey suggested there were roughly 19 million individuals in the United States who ran at least 100 days per year [1]. According to the industry monitoring group Running USA, participation in road races has nearly tripled from 1989 to 2010, with 10.4 million Americans participating in a road race during the 2010 calendar year [1]. Looking at marathons in particular, similar trends can be seen with approximately 465,000 Americans completing a marathon in 2009 [1].

While the number of runners continues to increase, one factor remains disturbingly consistent: the injury rate among these individuals. Runners get injured at incredibly high rates. Depending on the exact definition of injury used, epidemiologic studies have consistently reported 25% to 75% of all competitive and recreational runners will sustain a running related injury in any given year [2–12]. Using the estimate of 19 million runners in the United States [1], this suggests anywhere from 4.75 million to 14.25 million individuals will sustain a running related injury each year.

Even more troubling is that the actual injuries these individuals are experiencing have not changed despite over thirty years of research. For instance, in 1978 James et al.

[13] published one of the first papers detailing the most common running injuries based on observations of 262 patients treated in Dr. James's clinical practice (Table 1.1). Twenty five years later Taunton et al. [14] detailed similar parameters in a group of 2002 recreational runners (Table 1.1). The specific injuries and the frequencies with which they occurred were nearly identical to those reported by James et al. [13]. The most recent epidemiologic review, published in late 2012 by Lopes et al. [15] suggests some progress may have made in the in the intervening ten years with regards to knee injuries. However, for the other common running injuries the incidence remains unchanged from previous reports. Thus, overall, this suggest that despite over thirty years of advancement in sports medicine and thousands of research articles investigating biomechanical factors contributing to running injuries, understanding of the etiology of these common injuries is still lacking.

While few running injuries are life threatening, these injuries are not without associated costs. One study on the etiology of over 2000 running injuries reported that 31% of runners sought medical treatment, with most cases requiring an average of 3.8 medical consultations [2]. These authors also reported that 5% of observed running

Table 1.1. The most common running injuries and their frequency as reported by James et al. [13] and Taunton et al. [14]. Note how the same injuries occur with roughly the same frequency.

Injury (James et al. 1978)	N	%	Injury (Taunton et al. 2002)	N	%
Knee pain	67	29%	Patella femoral pain syndrome	331	16.5%
			Iliotibial band syndrome	168	8.4%
Plantar fasciitis	17	7%	Plantar fasciitis	158	7.8%
Posterior tibial syndrome (shin splints)	30	13%	Tibial stress syndrome (MTSS)	99	4.9%
Achilles tendinitis	25	11%	Achilles tendinitis	96	4.7%
Stress fractures	14	6%	Stress fractures – tibia	67	3.3%

related injuries were serious enough to lead to an absence from work, with the average duration of absence being 10.1 days. Van Middelkoop et al. [7] reported 41% of injured runners sought medical treatment and that 25% of these individuals were still symptomatic up to 3 months later. Jacobs and Berson [9] reported that 70% of injured runners in their study sought medical treatment for their injuries, with treatments ranging from corrective exercises, orthotic prescriptions, to surgical interventions in some cases. Thus, while the exact cost of running related injuries is not known, based on the sheer number of individuals sustaining these injuries, the number seeking medical treatment because of them, and the repeated visits often involved with treating these injuries, we can reasonably conclude it is not insignificant.

In addition to the health care costs, another troubling finding reported in the epidemiologic literature is that the most consistent risk factor for sustaining a running injury is having previously sustained a running related injury. For instance, Marti et al. [2] reported that previously injured runners had a 74% risk of sustaining a second injury. Similarly, Walter et al. [4] reported that the presence of an initial running related injury meant men and women were 1.69 and 2.35 times more likely to sustain a subsequent injury, respectively, while Buist et al. [11] reported males who had previously sustained an injury were 2.7 times more likely to sustain a second injury. A prospective study on 844 runners training for a 10 km road race reported 50% of runners who sustained and injury during the study period had previously sustained some form of running injury [5].

Powell et al. [16] presented three main arguments for why a previous injury appears to increase the likelihood of a subsequent injury: the previous injury did not heal completely before activity was resumed, the repaired tissue did not function as well or have the strength of the original tissue, or the fundamental underlying cause of the injury was not addressed leading to re-injury upon resuming activity. It is in this third area where biomechanical analysis can be particularly helpful. As such there has been an enormous volume of literature examining biomechanical factors contributing to common running injuries. The majority of these studies focus in on three key areas: anatomic and anthropometric variables, kinematic variables, and kinetic variables. This series of studies will primarily focus on the first two areas.

Anthropometric and Anatomic Factors Related to Running Injuries

The common running related injuries detailed in Table 1.1 are all overuse injuries, meaning they result from the repetitive application of loads rather than from one single traumatic incident. Hreljac [6] suggest the occurrence of running injuries should be considered in terms of a stress-frequency curve. At low frequencies, the body is capable of tolerating fairly high levels of stress before an injury occurs. However, the tolerance decreases as the frequency of stress application increases. Runners certainly live at the high frequency end of this curve. For instance, an average individual who runs 5 km per day at a ten minute mile pace will experience approximately 2,435 foot contacts per run [17]. The external loads applied to the musculoskeletal system on each contact are generally between two to three times and individuals body weight [18,19] while the internal forces generated by the muscles may be several times higher [20,21]. While the body is capable of tolerating such high forces and application frequencies when everything is aligned and working properly, any deviations in alignment may increase the susceptibility to injury.

Specific structural anatomic variables that have been linked to running injuries include leg length discrepancies [22], femoral neck anteversion, varus or valgus alignment of the calcaneus relative to the forefoot and tibia, pes planus or pes cavus foot structure under either static or dynamic conditions [23–25], squinting patella and high Q angle [26], and genu varus or valgus alignment at the knee [13,26–29]. However, despite the number of studies there is no clear agreement in the literature. For instance, Walter et al. [4] reported there were no differences in femoral neck anteversion, patella or rearfoot alignment between injured and uninjured runners. Montgomery et al. [30] reported prospective measures of knee varus valgus alignment did not predict running related injuries in military recruits. Based on both retrospective and prospective studies Wen et al. [27,31] concluded that static lower limb alignment did not appear to play a role in the development of overuse running injuries. Further complicating the issue is a study by Lun et al. [29] that reported relationships between genus varus and forefoot varus alignments and injury, but only for individuals with patellofemoral pain syndrome. There were no relationships between alignment measures and individuals with other injuries, suggesting the role of alignment may be injury specific.

While many of these anatomical structures are fixed and cannot be changed, other factors intrinsic to runners, such as flexibility or joint range of motion, are modifiable, and as such have received significant attention. However, as with the structural factors, there is little agreement regarding exactly how or if flexibility relates to injuries. For instance, while some authors have reported that runners who stretch infrequently are at a higher risk of injury [4], others have reported that runners who stretch regularly are actually at a higher risk of injury [9]. Similarly, Kaufman et al. [23] found reduced dorsiflexion range of motion at the ankle to be a significant risk factor for Achilles tendinitis while Van Mechlen et al. [32] reported there were no differences in ankle dorsiflexion range of motion between injured and uninjured individuals. However, they did find injured runners displayed reduced hip internal rotation range of motion compared to healthy control subjects. Other authors have reported reduced hip internal rotation range of motion as a potential predictor variable for the development of medial tibial stress syndrome [33]. In summary, when the literature in this area is viewed as a whole, there does not appear to be consistently observed relationships between running injuries and anatomic alignment. Thus, the connection between anatomic malalignment and running injuries is an area especially in need of further study.

However, what is clear from Table 1.1 is that the most common running injuries tend to occur at or below the knee. Thus, before discussing suspected biomechanical mechanisms responsible for these injuries, a detailed discussion on foot kinematics during running is required.

General Foot Kinematics and Anatomy of the Foot

Out of all the studies on limb kinematics, by far the most heavily investigated parameter is motion of the rearfoot. At initial contact runners will generally strike on the lateral aspect of their foot. For rearfoot striking runners this occurs on the lateral aspect of the heel while for midfoot striking runners first contact generally occurs halfway down the foot [18]. Since initial contact occurs slightly lateral to the ankle joint, the ground reaction force at this instant will cause the calcaneus to evert. On average, the longitudinal axis of the subtalar joint is tilted 42° from horizontal and rotated 23°

medially [34] (Figure 1.1). Given this orientation, calcaneal eversion cannot take place in the cardinal body planes and thus calcaneal eversion at the subtalar joint cannot happen without also producing movement at other joints. This coupled motion is often described as a metered hinge [34] and means that during load bearing activities such as walking or running, subtalar eversion is accompanied by dorsiflexion at the talocrural joint and abduction of the forefoot, occurring across the transverse tarsal and tarsometatarsal joints [34]. This combination of motion about the three cardinal body planes is called pronation of the foot.

While pronation is usually initiated by eversion of the calcaneus, as mentioned previously, this motion does not occur in isolation. As the calcaneus everts, maintaining joint congruity at the subtalar joint requires the talus to also evert and internally rotate. However, the talus is constrained by the medial and lateral malleoli of the ankle mortise,

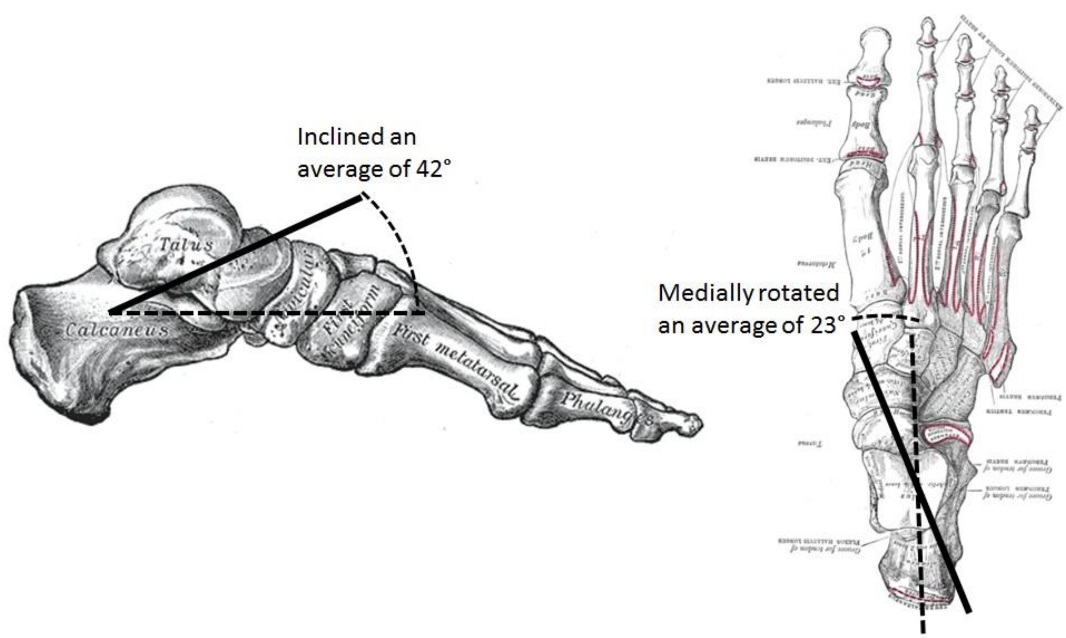

Figure 1.1. Schematic illustration showing the orientation of the subtalar joint axes. Bone images from [198].

and thus has limited ability to rotate in the transverse and frontal planes. Therefore internal rotation of the talus requires internal rotation of the tibia, thus coupling tibial rotation with pronation of the foot [34]. Similarly, as motions reverse, supination of the foot requires external rotation of the tibia. The physical restriction on eversion is solved by the talus plantar flexing slightly relative to the tibia as the calcaneus everts [34]. Thus, the combination of internal rotation and plantar flexion allow the talus enough mobility to maintain congruity at the subtalar joint.

Both the talus and calcaneus bones possess substantial neck regions extending out to the talonavicular and calcanealcuboid joints, respectively (Figure 1.2). These two joints together are also referred to as the transverse tarsal joint. While the talus internally rotates and plantar flexes during the first half of stance, the forefoot does not follow similar motions. This is largely due to the fact that the transverse tarsal joint serves to

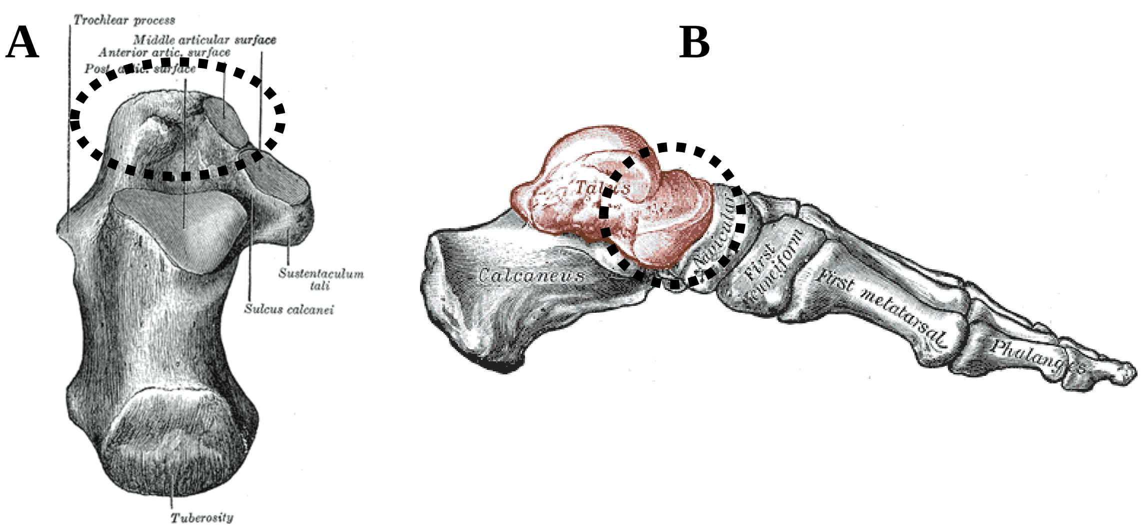

Figure 1.2. Views of the calcaneus (A) and talus (B) with the neck region of each bone circled. The neck of the calcaneus articulates with the cuboid bone at the calcaneocuboid joint while the neck of the talus articulates with the navicular at the talonavicular joint. Images from [198].

convert and transmit the movements of the calcaneus and talus to the distal foot bones [34,36]. A recent *in vivo* study examining intrinsic foot motion during running using intracortical bone pins inserted into nine bones of the foot reported that while substantial amounts of motion were present at both joints, the primary planes of movement were different [37]. At the talonavicular joint the most motion occurred in the frontal plane (13.5° of eversion on average) with about half that amount of motion observed in the sagittal and transverse planes (6.5° of dorsiflexion and 8.7° of abduction, on average) [37]. However, at the calcanealcuboid joint there were approximately equal amounts of motion in the sagittal, frontal, and transverse planes (7.2° of dorsiflexion, 7.1° of eversion, and 6.9° of abduction on average) [37]. These large amounts of motion reinforce the importance of the transverse tarsal joints for transmitting the movement of the calcaneus and talus to the distal foot bones. As additional *in vivo* work emerges our understanding of the motion and demands placed on these joints during dynamic activities such as running will continue to develop.

In addition to providing a link between the rearfoot and the forefoot, the motion of the talus, calcaneus, cuboid and navicular causes the axes of the transverse tarsal, cuneonavicular joints, and tarsometatarsal joints to align , "unlocking" the midfoot [34]. The functional consequence of this configuration is that the foot assumes a soft, flexible structure. Thus, one function of the foot during this period is to help dissipate and cushion the impact forces arising from the foot contacting the ground [38,39]. Indeed, it has been demonstrated that artificially reducing the amount of available pronation increases tibial acceleration, the magnitude of the first peak in the vertical ground reaction force, and the magnitude of the vertical loading rate [40]. In addition to the cushioning benefits, a second function of the soft and flexible foot is that it allows accommodation for difference surface textures and can adapt to uneven terrain. Compared to a rigid structure this increases overall stability of the foot, and subsequently the rest of the body, at foot contact.

In healthy, uninjured individuals, the foot typically strikes the ground in a slightly supinated position and then pronates between 10° to 16° [35,41–47], with peak pronation being reached between 35% to 50% of stance [48]. As discussed, the functional structure of the foot during this time allows for cushioning and stability. However, the overall function of the foot changes during the second half of stance. Rather than absorbing shock or adapting to uneven terrain, the foot now becomes a lever with which a successful push off can be generated. This is accomplished by reversing the direction of bone movement that occurred during pronation. As the calcaneus inverts the talus externally rotates and dorsiflexes, movements which, in turn, cause the axes of the transverse tarsal, cuneonavicular, and tarsometatarsal joints to diverge, essentially "locking" up the midfoot [34]. This turns the foot from a soft flexible structure into a rigid lever capable of propelling the body forward into the next flight phase.

While the movement patterns at the bony level are opposite in direction, there is one major difference in the actual causes of the motion. Since the foot is typically supinated slightly when it makes contact with the ground, the initial contact point is lateral to the instantaneous center of rotation. Thus, the ground reaction force at this instant serves to start the eversion of the foot. In other words, without any muscular control, the simple action of an individual's body weight settling onto the foot would cause these bony motions to occur. This is not necessarily true for supination of the foot.

Examinations of the center of pressure trajectory during running shows that the center of pressure finally translates medially to the long axis of the foot between 60% to 80% of stance [49]. After this point the ground reaction force would help invert the foot. However, since maximum pronation is reached between 35% and 50% of stance, the initial supination cannot be caused by the ground reaction force. Rather, this responsibility rests with the intrinsic and extrinsic muscles of the foot.

Depending on how one counts there are twenty individual intrinsic foot muscles, with two on the dorsal aspect, fourteen on the plantar aspect, and four located in between the dorsal and plantar surfaces of the metatarsals (Figure 1.3A; [50]). One function of these muscles is to produce movement at the metatarsophalangeal and interphalangeal joints. Since these muscles are all intrinsic to the foot and fairly small in size they cannot generate large torques needed to initiate supination. However, they serve an important

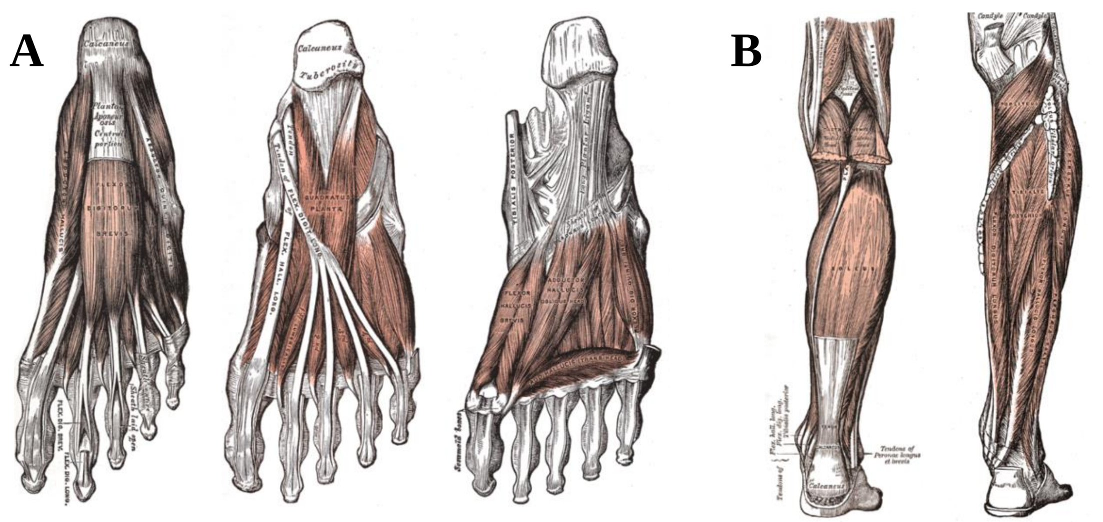

Figure 1.3. Illustration showing most of the intrinsic foot muscles from a plantar view (A) and the extrinsic muscles from the superficial and deep posterior compartments (B) from a posterior view. Images from [198].

role by helping stabilize and support the longitudinal arch of the foot during mid and late stance [34,50].

In addition to the twenty intrinsic muscles, there are an additional thirteen muscles that originate either on the femur, proximal tibia, or proximal fibula, and insert into the foot (Figure 1.3B). Of these muscles, there are two that serve as primary invertors of the foot. First, is the tibialis posterior which originates from the superior aspect of the tibia and fibula and the inerosseous membrane, and inserts on the navicular tuberosity, and cuneiforms [50]. The insertion site and line of action for this muscle make it especially powerful for inverting the subtalar joint, thus locking midfoot joints [34]. This function is supported by the results of several electromyographic studies which all show this muscle has a large burst of activity during midstance while running [51–53].

The second primary invertor of the foot is the soleus muscle. Both the soleus and the gastrocnemius come together to form the Achilles tendon. While the soleus muscle is deep to the gastrocnemius on the leg, their muscle bellies lie mostly in the same plane. However, as the Achilles tendon descends it rotated medially anywhere from 30° to 150° so that the portion of the tendon arising from the soleus constitutes the medial portion of the tendon while portion arising from the gastrocnemius constitutes the lateral portion of the tendon [54–56]. Thus, while the soleus's moment arm relative to the center of rotation is fairly small, the combination of the size of the muscle and its medial insertion into the calcaneus combine to make it a power invertor. As with the tibialis posterior, electromyographic studies have supported this role, finding large bursts of activity during midstance of running [51,52].

Theoretical Relationships Between Foot Kinematics and Common Running Injuries

Given the interplay between foot pronation and normal foot function, it is not surprising that abnormal foot pronation has been suggested to play a role in the development of many common overuse running injuries, with excessive amounts or velocities of pronation being the most commonly cited abnormalities. Specifically, excessive amounts or velocities of pronation have been suggested as a contributing factors in the development of Achilles tendinopathy [46,57,58], plantar fasciitis [59–61], and medial tibial stress syndrome [24,25,33,59,62–66]. While these are all soft tissue injuries, excessive amounts of pronation have also been reported in individuals with a retrospective history of tibial stress fractures [67].

The theoretical relationship between excessive amounts or velocities of pronation makes sense when considering a combination of anatomy and proposed mechanisms for injury occurrence. There are currently two commonly cited theories on the etiology of medial tibial stress syndrome. Some authors have suggested it is a traction based injury involving irritation or even possibly a slight evulsion of the periosteal layer along the posterior face of the tibia [68–71]. The structures most commonly implicated for the development of medial tibial stress syndrome include the flexor digitorum longus [68,72], tibialis posterior [69,73], and soleus [68,72,74] muscles, as well as the deep crural fascia connecting these muscles to the tibia [68,75]. Exactly which of these muscles is most responsible for the injury is debated as some authors have suggested the tibialis posterior does insert around the injury site [73] while others have reported the tibialis posterior does not insert in the region usually affected by medial tibial stress syndrome [72].

An alternative view suggests that medial tibial stress syndrome does not arise from excessive traction, but rather from repeated micro trauma caused by bending stress in the tibia [76]. During running, the tibia is placed under bending stress, with the anterior aspect of the bone being loaded in tension and the posterior aspect in compression. Bone is a dynamic tissue, and as such, according to Wolff's law, adapts by depositing new bone in the area of greatest strain. Thus, the actual pathologic origins of medial tibial stress syndrome may simply result from an imbalance between strain application and bone deposition. In this light, medial tibial stress syndrome could be thought of as lying along a continuum with tibial stress fracture. Experimental support for this theory comes from studies reporting narrower tibial diaphysis widths in individuals with medial tibial stress syndrome or stress fractures when compared to healthy control subjects [77–79].

Though the exact mechanism of injury for medial tibial stress syndrome and the role of the various anatomic structures involved in injury development still need to be clarified, as can be seen in Figure 1.4, both mechanisms and all the anatomic structures thought to be involved would be negatively affected by increased amounts or velocities of pronation. For instance, increased pronation would cause the navicular and cuneiform bones to drop, increasing the strain placed on the tendons for the flexor hallucis and flexor digitorum longus muscles (Figure 1.4A). There is some evidence to suggest strain in these muscles is transmitted through the fascia to the bone [68]. As the muscles resist the elongation, either actively or passively, they will be applying a force on the tibia at their insertion points. This force would cause a bending stress in the tibia. If the muscles

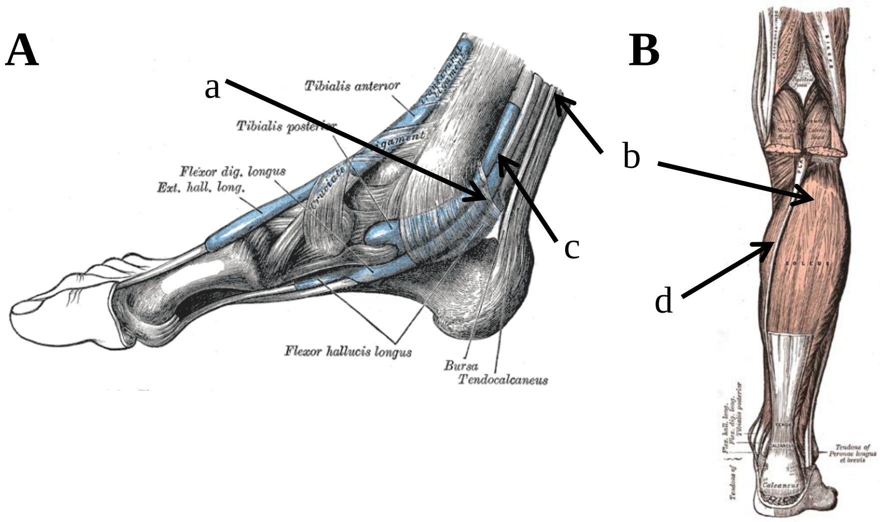

Figure 1.4. Illustrations of the anatomic structures thought to be involved in medial tibial stress syndrome (A) and Achilles tendinopathy (B). The following muscles are labeled a: flexor digitorum longus, b: the soleus, c: the tibialis posterior, and d: the gastrocnemius. Images from [198].

are working eccentrically to resist the lengthening the forces applied to the tibia could be quite high.

A similar thought process could be applied to examine how excessive amounts or velocities of pronation could play a role in the development of Achilles tendinopathies. As pronation increases the amount of calcaneal eversion will increase. Increased calcaneal eversion will place the Achilles tendon under greater strain. Several authors have described the etiology of Achilles tendinopathy as an inflammatory reaction in response to repeated application of high strains to the tendon [80,81]. In support of this hypothesis, an i*n vitro* study subjecting the Achilles tendon to repeated loading cycles reported the amount of strain present was the major factor in determining time to tendon failure [82].

In addition to the amount of strain present, the distribution of the strain also appears to play a role in injury development, with heterogeneously distributed strain being especially important for injury development [83,84]. In this regard, the medial rotation of the Achilles tendon is potentially problematic. The combination of the soleal contributions to the Achilles tendon being on the medial side, and this muscles action as a strong invertor of the foot suggests increased pronation leads to increased heterogeneity of strain distribution within the tendon. This was confirmed in a recent study by Lersch et al. [85] that reported that differences in calcaneal inversion-eversion positioning could change Achilles strains by up to 15%, with higher amounts of eversion leading to higher strain in the medial portion of the tendon.

In addition to the amount of pronation, the velocity of pronation has also been implicated as a possible factor for the development of Achilles tendinopathy [46]. The main hypothesis is that high pronation velocities result in a “whip-like” motion within the tendon. In other words, high pronation velocities and the accompanying eversion of the calcaneus may initially result in a relaxation within the tendon. However, as the tibia dorsiflexes over the talus during stance, this relaxation is quickly replaced by a rapid increase in strain as the tendon catches up with foot motion. While this model includes the previously discussed high amounts of strain, it also suggests the strain rate may be an important factor.

Issues with the Current Theories and an Alternative View

While the previous hypotheses regarding how excessive amounts or velocities of foot pronation play a role in the development of medial tibial stress syndrome and

Achilles tendinopathy make sense based on the anatomy and mechanics involved, they suffer from a lack of support in the epidemiologic research. For instance, in contrast to the previously cited studies which reported higher amounts of pronation or pronation velocities in injured individuals, other authors have reported no differences in either the amount or velocity of pronation in individuals with medial tibial stress syndrome [86–89], or Achilles tendinopathy [90,91]. Thus, there is no agreement on whether or not higher amounts or velocities of pronation are related to common running injuries.

This discrepancy may partly be explained by methodological issues. For instance, many of the studies examining relationships between foot pronation and medial tibial stress syndrome have been performed in clinical settings lacking access to full three dimensional motion capture systems. In these studies static or dynamic measures of navicular drop are commonly used as markers of foot pronation [24,25,65,87]. Other authors have reported the amount pronation based on two dimensional motion analysis by calculating the angle between the vertical bisection of the calcaneus and the vertical bisection of the tibia [59,62]. These measures only measure frontal plane movement and small differences in camera alignment could have a large impact on measured results [92]. However, as previously described, foot pronation includes significant amounts of motion outside of the frontal plane and increased foot abduction has been shown to increase the differences between two dimensional and three dimensional measures of pronation [93]. Currently, studies utilizing full three dimensional motion capture typically measure pronation by calculating an Euler angle rotation sequence describing the motion of the rearfoot relative to the shank [94–96]. This second method for measuring foot pronation has been widely adopted in the biomechanical literature with standard coordinate systems and rotation sequences having been suggested [97]. However, whether or not there is agreement between the two methods has not been described in the literature.

In addition to methodological issues, a second possible explanation involves the definition of what constitutes "excessive" amounts or velocities of pronation. As stated previously, "normal" amounts of pronation tend to include between 10° and 16° of calcaneal eversion after foot contact. Table 1.2 shows the amounts of pronation reported in the injured subjects by several studies, which found injured individuals demonstrate more pronation than healthy controls for either medial tibial stress syndrome or Achilles tendinopathy. If normal ranges of pronation are between 10° to 16°, then from all of these, only the findings by Donoghue et al. [57] could truly be considered "excessive". Thus, while injured individuals in these studies demonstrate greater amounts of pronation than healthy controls, they are still within what could be considered "normal" ranges of motion.

When considered as a whole the conflicting experimental evidence linking

Table 1.2. Average amounts of pronation (rearfoot eversion) reported by several authors finding injured individuals demonstrate greater amounts of pronation than non-injured individuals.

Authors	Injury	Average Pronation in the Injured Group
Messier and Pittala [59]	Medial tibial stress syndrome	16.5° (± 1.3°)
Willems et al. [95]	Exercise related lower leg pain	15.5° (± 5.5°)
Donoghue et al. [57]	Achilles tendinopathy	21.1° (± 5.3°)
Ryan et al. [58]	Achilles tendinopathy	13.0° (± 3.0°)
McCrory et al. [46]	Achilles tendinopathy	13.3° (± 1.4°)

excessive amounts or velocities of pronation and common running injuries suggests an alternative approach may be warranted. One such approach that has not been systematically explored is the idea of prolonged pronation as opposed to excessive amounts or velocities of pronation. As previously discussed, when the calcaneus starts inverting during mid-stance, the axes of the transverse tarsal, cuneonavicular joints, and tarsometatarsal joints diverge, essentially "locking" up the mid-foot. This turns the foot into a rigid lever for use during push off. However, if pronation is prolonged beyond mid-stance then push off will begin while the foot is still in a soft flexible configuration. This configuration may require additional effort from the intrinsic foot muscles to stabilize the foot and from the extrinsic foot muscles to generate the torques required to achieve push-off [13]. First proposed by James et al. in 1978 [13], to date, the idea of prolonged pronation has received little attention in the running injury literature.

Overall Goals and Specific Aims for the Dissertation

The series of studies in this project will examine the concept of prolonged pronation and its possible implications for two of the most common running injuries, medial tibial stress syndrome and Achilles tendinopathy [13–15]. As previously discussed, there is little agreement in the literature regarding the role of excessive amounts of velocities of pronation and the relationship to these injuries. Thus, the overarching working hypothesis for this project is that it is not necessarily the amount or velocity which matter, but rather the duration over which pronation is prolonged throughout stance. The results this study will help clarify mechanisms responsible for these two common injuries as well as provide a basis for future work examining the effects of intervention strategies in preventing or rehabilitating these injuries. Within this overall context, the following specific aims were proposed.

Identify Specific Biomechanical Markers of Prolonged Pronation

The potential implications of prolonged pronation have been discussed numerous times in the literature [13,41,62,98–103]. However, parameters reported in these studies usually reflect the amount of pronation, velocity of pronation, or time until peak pronation is reached, rather than an actual measure of the duration of pronation. The closest thing to an actual measure of pronation duration is found in two studies that use a measure called the period of pronation, a representation of the amount of time the foot remains in a pronated position during stance [104,105]. While these two studies examined how the period of pronation changes with different shoe conditions, they did not provide information on what constitutes excessive values for the period of pronation, nor did they examine whether individuals with longer periods of pronation also demonstrate greater amounts or velocities of pronation. Additionally, there is currently no literature describing any other biomechanical parameters, either those that can be measured in clinical settings or lab settings, which may identify individuals with longer periods of pronation. Therefore the purpose of this study was to identify biomechanical markers of prolonged pronation in healthy runners. Through collaboration with two clinicians, a group of runners with clinically determined prolonged pronation were be identified. This group was then compared to a group clinically determined to have "normal" pronation and biomechanical parameters differentiating the two groups was identified. The hypothesis tested in this study was that individuals who demonstrated prolonged pronation would not demonstrate greater amounts or velocities of pronation than those who did not demonstrate prolonged pronation.

Examine Whether Injured Individuals Display Biomechanical Markers of Prolonged Pronation

The logic behind pronation duration being the important factor for injury development is that by having prolonged pronation, individuals slightly increase the demand placed on the intrinsic and extrinsic foot muscle on each step. The cumulative effect of this increased demand may either cause or increase the likelihood of these individuals sustaining an overuse injury. However, in order for this to be true, then injured individuals should theoretically demonstrate characteristics associated with prolonged pronation. Upon completion of the first specific aim, biomechanical markers identifying individuals with prolonged pronation would have been identified. Therefore, the second specific aim is to examine whether individuals currently symptomatic with Achilles tendinopathy or medial tibial stress syndrome displayed these same biomechanical markers. The biomechanical markers of prolonged pronation from the first study were evaluated in individuals currently symptomatic with these injuries and in healthy control subjects. It was hypothesized that, compared to healthy control subjects, injured subjects would not demonstrate greater amounts or velocities of pronation. Rather, they would demonstrate a greater duration of pronation than the healthy controls.

Examine Musculotendon Kinematics in Healthy, Injured, and Prolonged Pronators

It was hypothesized in the second specific aim that currently injured individuals would demonstrate characteristics of prolonged pronation. However, since these individuals were already injured, and this study only observed their behavior after the injury it could not conclusively determine whether the observed movement patterns were responsible for causing the injury in the first place. However, examining muscular dynamics may give some insights into whether the observed movement patterns may lead to injury. For instance, it is hypothesized with medial tibial stress syndrome that increased strain in the crural fascia leads to greater traction forces at the bony insertion and that higher forces ultimately may lead to the periosteal inflammation thought to cause this injury [68,106]. It has also been shown that as strain increases within the tibialis posterior, flexor digitorum longus, and soleus muscles, strain within the crural fascia also increases [68]. Similarly, higher amounts of strain through both the triceps surae muscles and the Achilles tendon itself have been discussed as potential mechanisms for Achilles tendinopathy development [82–84]. Therefore the goal for this specific aim was to examine whether there were differences in musculotendinous strain and strain rates between healthy individuals with "normal" pronation, healthy individuals with prolonged pronation, and individuals currently symptomatic with either medial tibial stress syndrome or Achilles tendinopathy.

While it can be done, measuring strain in either a tendon or muscle *in vivo* is both a difficult and invasive procedure. Thus for this specific aim OpenSim musculoskeletal modeling software [107] was used to estimate musculotendinous strain and strain rates by calculating the percent elongation and rate of percent elongation for seven extrinsic foot muscles. It was hypothesized that muscle percent elongation and percent elongation rates would fall along a continuum with the injured individuals demonstrating higher musculotendinous percent elongations and percent elongation rates compared to the prolonged pronators, which in turn, demonstrate higher percent elongations and percent elongation rates than healthy individuals with “normal” pronation.

Flow of the Dissertation

This dissertation is structured in a journal format. The studies described in Chapters II-V include co-authored materials and are individual manuscripts prepared for submission to peer-reviewed scientific journals. One of the methods used to quantify prolonged pronation involved examination of the center of pressure trajectories. Following this introduction, Chapter II details the development of this methodology and several examinations of its ability to detect subtle differences between experimental conditions. Chapter III applies this method and other biomechanical analyses to identify biomechanical markers of prolonged pronation. Chapter IV examines whether individuals currently symptomatic with medial tibial stress syndrome or Achilles tendinopathy demonstrate characteristics of prolonged pronation. Chapter V uses a musculoskeletal model to examine differences in musculotendinous percent elongations and percent elongation rates between three groups of individuals: those currently symptomatic with medial tibial stress syndrome or Achilles tendinopathy, those who are currently healthy but have prolonged pronation, and those who are currently healthy and have “normal” pronation. Finally, Chapter VI provides conclusions, a discussion on the dissertation’s limitations, and suggestions for follow up research directions.

CHAPTER II

CENTER OF PRESSURE TRAJECTORY DIFFERENCES BETWEEN SHOD AND BAREFOOT RUNNING

This chapter contains co-authored material and was developed by Dr. Li-Shan Chou, Dr. Louis Osternig, and James Becker. Dr. Chou contributed to the development and refinement of the methodology as well as providing critiques and editing advice for the manuscript. Dr. Osternig also provided critiques and editing advice for the manuscript while Mr. Becker was responsible for conceptual development, development of the protocol, data collection and analysis, as well as all the writing.

Introduction

The location of the center of pressure (COP) provides information on the point of application of the net ground reaction force (GRF) beneath the foot and has become a common tool for assessing dynamic foot function during running. COP trajectories have been used to examine differences in foot function between high and low arched individuals [108], as a measure for quantifying mediolateral stability of the foot [109], and as a tool for prescribing and assessing the effects of shoe or orthotic interventions [110]. The COP trajectory may also provide information on the relative risk for sustaining common overuse running injuries such as Achilles tendinopathy [90] or exercise related lower leg pain [94].

Most studies investigating COP trajectories during running use pressure sensing insoles [110] or a pressure platform [49,111]. However, it is also possible to calculate the location of the COP on a force plate [18] which allows for simultaneous measurement of all three components of the GRF, something that is not possible on pressure plates. However, the COP trajectory obtained from a force plate is referenced to the orthogonal axes of either the force plate or the room. In order to be informative regarding injury risk or performance factors, the COP location should be quantified relative to the anatomic structures of the foot. While pressure sensing insoles or pressure platforms will inherently provide this information, obtaining it from a force plate requires knowledge of the foot's orientation on the plate.

While methods for obtaining this information using chalk [18] or ink [112] outlines of the foot have been reported, these methods are subject to practical limitations such as requiring a coating of material on the force plate. Avoiding these limitations, other authors have reported methods using markers placed on the shoe to align COP trajectories measured from a force plate with those measured from pressure sensing insoles [113,114]. The good agreement between the transformed force plate derived COP trajectories and the pressure insole derived COP trajectories reported by these authors suggests this is a feasible method for using a force plate to quantify COP trajectories relative to the foot's anatomic structures. However, to date, these studies have focused on methods for aligning COP trajectories derived from force plates and pressure sensing insoles and the method has not been applied to examining differences in experimental conditions.

Recently, barefoot (BF) running has seen a resurgence in popularity. When running BF the foot is often in a less supinated position at initial ground contact then when running shod (SH) [115]. It has been suggested this kinematic change reduces the external eversion moment from the GRF, thereby reducing the tendency for the rearfoot to evert [116]. However, whether this actually happens is not clear as Kerrigan et al. [117] reported that there are no differences in the net ankle eversion moment between SH and BF running. This discrepancy may partially be explained by understanding how the location of the COP differs between SH and BF running. For instance, while the COP may be located more laterally when SH then when BF, if the joint center also shifts laterally this will partially or wholly negate any changes in external eversion moment [111]. However, to date, there are no descriptions in the literature regarding how COP trajectories differ between SH and BF running.

Therefore, the purpose of this study was to investigate differences in COP trajectories and resulting external moments between shod (SH) and barefoot (BF) running. It has been previously reported that having habitually SH individuals run BF consistently results in a more plantar flexed ankle at initial contact [115]. However, changes in ankle eversion are less consistent and vary from subject to subject [115]. Therefore, we hypothesized that compared to SH running, the COP during BF running would be more anteriorly located during early stance but that there would not be consistent differences in its mediolateral location. Based on the hypothesized COP trajectory difference, we also hypothesized there would be a larger external dorsiflexion moment in the BF condition but that there would not be consistent differences in the external eversion moment.

Methods

Subjects

Morley et al. [45] reported significant differences in rearfoot eversion when habitually SH runners ran BF (SH: 10.6 ± 1.5°, BF: 7.4 ± 2.5°). Based on these data, a power analysis revealed that 7 subjects would be required to adequately power this study (effect size 1.47, $\alpha = 0.05$, $\beta = 0.05$). Therefore, 10 habitually SH runners participated in this study (sex: 4 female, 6 male; age: 41 ± 8.7 years; weekly mileage: 39.5 ± 10.3 miles). Specific inclusion criteria included currently logging at least 20 miles per week, having no history of injuries within the last 6 months, and no prior experience BF running. Additionally, subjects had to use a rearfoot strike (RFS) while running SH and switch to a midfoot strike (MFS) while running BF since many of the proposed benefits of BF running are tied to utilizing a MFS [116]. Since some amount of kinematic and kinetic asymmetry is present even in uninjured runners [118] subject's left and right feet were analyzed separately. Therefore, the total *n* for the statistical analysis in this study was 20 feet.

Experimental Protocol and Instrumentation

Reflective markers were placed bilaterally on the medial and lateral malleoli, heads of the 2nd metatarsal, the tuberosities of the navicular and 5th metatarsal, two along the vertical bisections of the posterior calcaneus, and one on the lateral aspect of the calcaneus. For the BF condition markers were placed directly on the skin while for the SH condition they were placed on the skin in the identical locations as during the BF condition but visible through holes cut into the shoe (Figure 2.1A). To ensure consistent placement between conditions marker location on the skin was circled with a black pen.

Subjects then ran continuous laps around a short track in the laboratory at self-selected speeds approximating their easy training pace under SH then BF conditions (Figure 2.1B). Marker trajectories were recorded with a 10-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) while ground reaction forces were measured with three force plates (AMTI, Inc., Watertown, MA) sampling at 200 Hz and 1000 Hz, respectively.

Data Analysis

Marker trajectories and ground reaction forces were smoothed using second order, zero lag, low pass Butterworth filters with cutoff frequencies of 8 Hz and 50 Hz [119], respectively. Timing of initial contact and toe off were identified as the first instant where the vertical ground reaction force (vGRF) rose above, then fell below, a 50 N threshold [18]. To ensure subjects were using similar foot strike patterns the strike index (SI; [18]) was calculated for all trials and subjects. The SI is a ratio comparing the location of the first contact point on the COP trajectory relative to the length of the foot.

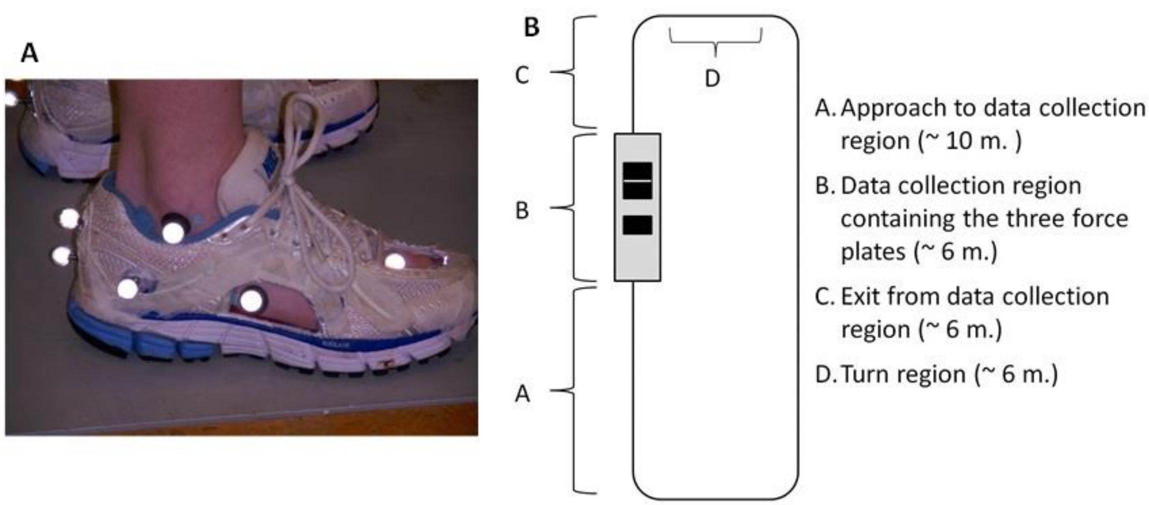

Figure 2.1. Foot marker placements used in the study (A) and an illustration of the track in the laboratory around which subject ran (B). These were consistent through all subsequent studies in the dissertation.

Values under 33% indicate a RFS, values between 33 and 66% indicate a MFS, and values greater than 66% indicate a forefoot strike (FFS) [18].

A local foot coordinate system (FCS) defining the orientation of the foot relative to the global coordinate system (GCS) was established at each instant of stance from initial contact through toe off. To examine the COP trajectory relative to the anatomic structures of the foot the COP was first calculated in the GCS then transformed to the FCS. As a result, the COP trajectories from numerous foot strikes, even if they occurred on different force plates, could be compared and the location of the COP could be related to anatomic structure of the foot (Figure 2.2). Once expressed in the FCS, the anterior-posterior (COP_{AP}) and mediolateral (COP_{ML}) positions at each 10% stance interval were used for comparison between conditions. Additionally, the following discrete variables describing the COP trajectory were extracted for analysis: COP_{AP} and COP_{ML} excursions, the most medial location of the COP, and the percent stance at which the most medial COP location occurred. These variables were specifically selected since prospective studies have observed reduced COP_{AP} excursion in individuals who subsequently sustained Achilles tendon injuries [90] and increased COP_{ML} excursion and more medial COP locations in individuals who subsequently suffered from exercise related lower leg pain [47]. Finally, the percent stance at which the maximum vertical GRF occurred, and the COP_{AP} location at this instant were also extracted.

To examine the external dorsiflexion and eversion moments, the ankle joint center (AJC) was first calculated in the GCS as the midpoint between the medial and lateral malleolus and transformed into the FCS. The GRF was also transformed into the FCS and the external dorsiflexion and eversion moments from the GRF were then calculated.

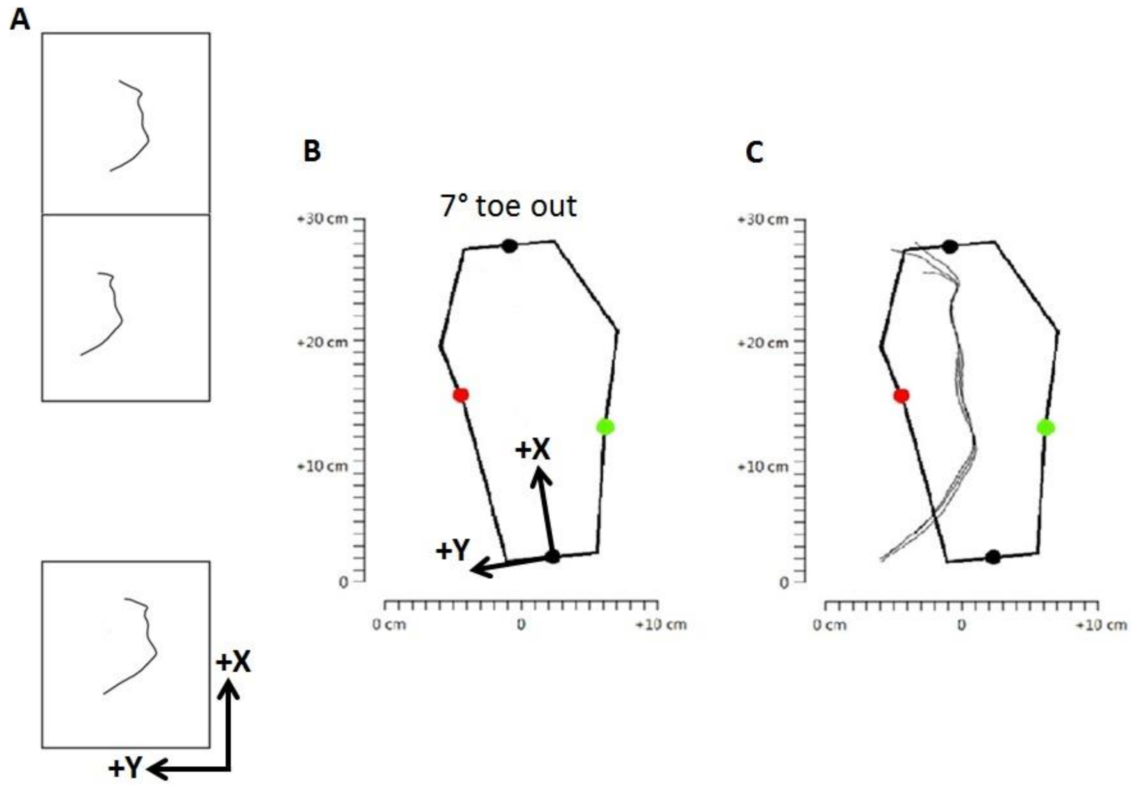

Figure 2.2. COP trajectories are initially calculated in the global coordinate system (GCS) (A). A local foot coordinate system was established at each instant during stance (B). The long axis of the LCS aligns with the longitudinal midline axis of the foot as defined by the heel and toe markers. By rotating the COP trajectory from the GCS to the FCS at each instant, COP trajectories from multiple trials can be compared relative to the anatomic structures of the foot (C).

Moments were normalized by subject's body mass. The AP and ML location of the AJC in the FCS was calculated across stance and at 10% intervals. Three variables were calculated to describe the external eversion moment: the maximum value during the first 15% of stance, the peak value, and the percent stance where the peak value occurred. For the external dorsiflexion moment the maximum value and percent stance at which the maximum value occurred were calculated.

Statistical Analysis

Eight trials per subject per foot were averaged. Differences between SH and BF conditions in COP_{AP} and COP_{ML} excursions, the most medial location of the COP, the percent stance at which the most medial location occurred, the variables describing the external dorsiflexion and eversion moments were compared using paired *t*-tests. For these comparisons a value of $p < .05$ was used to indicate statistical significance. Pair *t*-tests were also used to compare the COP_{AP}, COP_{ML}, and AJC positions at each 10% of stance, however, to reduce the risk of a Type I error, a Bonferroni correction was applied for these comparisons and statistical significance was defined as $p < .0045$ (.05/11).

Results

Running Speed and Foot Strikes

Running speed was not significantly different between SH (3.30 ± 0.41 m/s) and BF (3.27 ± 0.42 m/s) conditions ($p = 0.198$). Mean SI values indicated that when converting from SH to BF conditions the subjects shifted their initial point of contact anteriorly, adopting a MFS pattern while running BF ($p < .001$, Table 2.1).

COP Locations and Excursions

The COP was located more anteriorly in the BF condition than the SH condition at initial contact, 10%, and 20% of stance. Beyond this point there were no differences in the COP_{AP} position (Figure 2.3A). The COP was located more medially in the BF condition than the SH condition at all points except initial contact and 20% of stance (Figure 2.3B). Mean COP_{AP} excursions were smaller in the BF condition than in the SH

Table 2.1. Results for the strike index, COP_{AP} and COP_{ML} positions and excursions, and positioning of the COP at maximal vertical GRF. A negative value for most medial location of COP indicates the COP is medial to the long axis of the foot. * indicates SH is significantly different than BF with $p < .05$.

Variable	Shod	Barefoot
Strike Index Values	19.6 (± 7.3)	48.59 (± 12.04)*
COP_{AP} Excursion (% foot length)	65.8 (± 11.7)	34.79 (± 22.22)*
COP_{ML} Excursion (% foot width)	31.1 (± 18.6)	29.43 (± 17.46)
Most medial COP location (% foot width)	-2.2 (± 7.9)	-13.55 (± 7.63)*
Percent stance most medial location (%)	81.9 (± 21.3)	95.5 (± 9.04)*
Percent stance at maximal vertical GRF (%)	44.8 (± 3.5)	45.4 (± 3.5)
COP_{AP} at instant of maximal vertical GRF (% foot length)	61.5 (± 4.1)	61.7 (± 4.5)

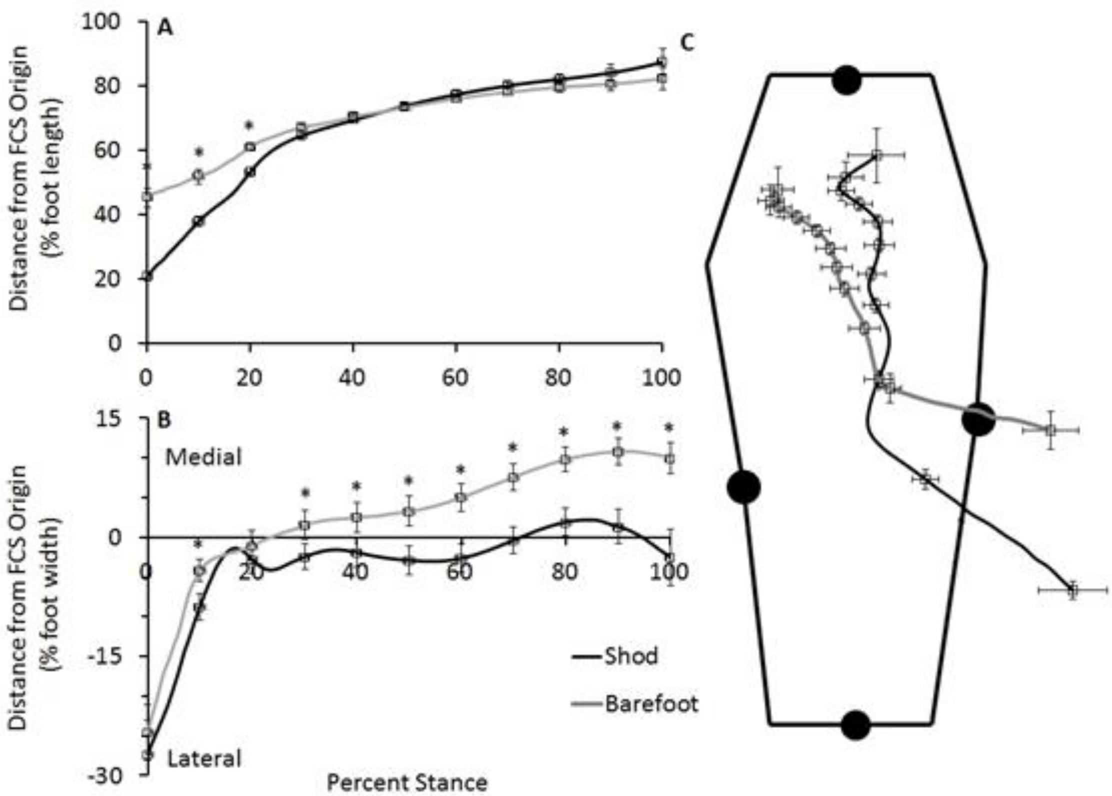

Figure 2.3. Average anterior-posterior (A) and mediolateral (B) positions of the COP across stance along with a graphical illustration showing average COP trajectories within a rough outline of the foot (C). * indicates SH is significantly different than BF at $p < .0045$.

condition ($p < .001$); however there were no differences in COP_{ML} excursion between conditions ($p = .753$; Table 2.1). For the BF condition the most medial location of the COP was further medial from the long axis of the foot than in the SH condition ($p < .001$) and occurred later during stance ($p < .001$; Table 2.1). There were no differences between conditions in the percent stance at which maximal vertical GRF occurred or in the COP_{AP} position at this instant (Table 2.1).

AJC Locations and External Moments

For all time points except initial contact, the AJC was located more posteriorly in the BF condition compared to the SH condition. The AJC was also located more medially in the BF condition at all points across stance (Figure 2.4C) and the most medial location of the AJC was more medial in the BF condition then in the SH condition (BF: -11.47 ± 7.21 % foot width; SH: -7.29 ± 6.95 % foot width; $p < .001$). The peak external dorsiflexion moment was higher in the BF condition than the SH condition (SH: 2.1 ± 0.5 Nm/kg; BF: 2.4 ± 0.4 Nm/kg, $p = .01$) and occurred earlier in stance (SH: 58.6 ± 3.3 % stance; BF: 56.5 ± 4.1 % stance; $p = .001$; Figure 2.4A). The external eversion moment during the first 15% of stance was not different between conditions (SH: 0.266 ± 0.188 Nm/kg; BF: 0.219 ± 0.102 Nm/kg; $p = .274$), however the peak external eversion moment was higher in the SH condition than the BF condition (SH: 0.415 ± 0.178 Nm/kg; BF: 0.329 ± 0.151 Nm/kg; $p = .027$; Figure 2.4B). The timing of the maximal external eversion moment was not different between conditions (SH: 38 ± 18.10 % stance; BF: 35.3 ± 15.31 % stance; $p = .349$).

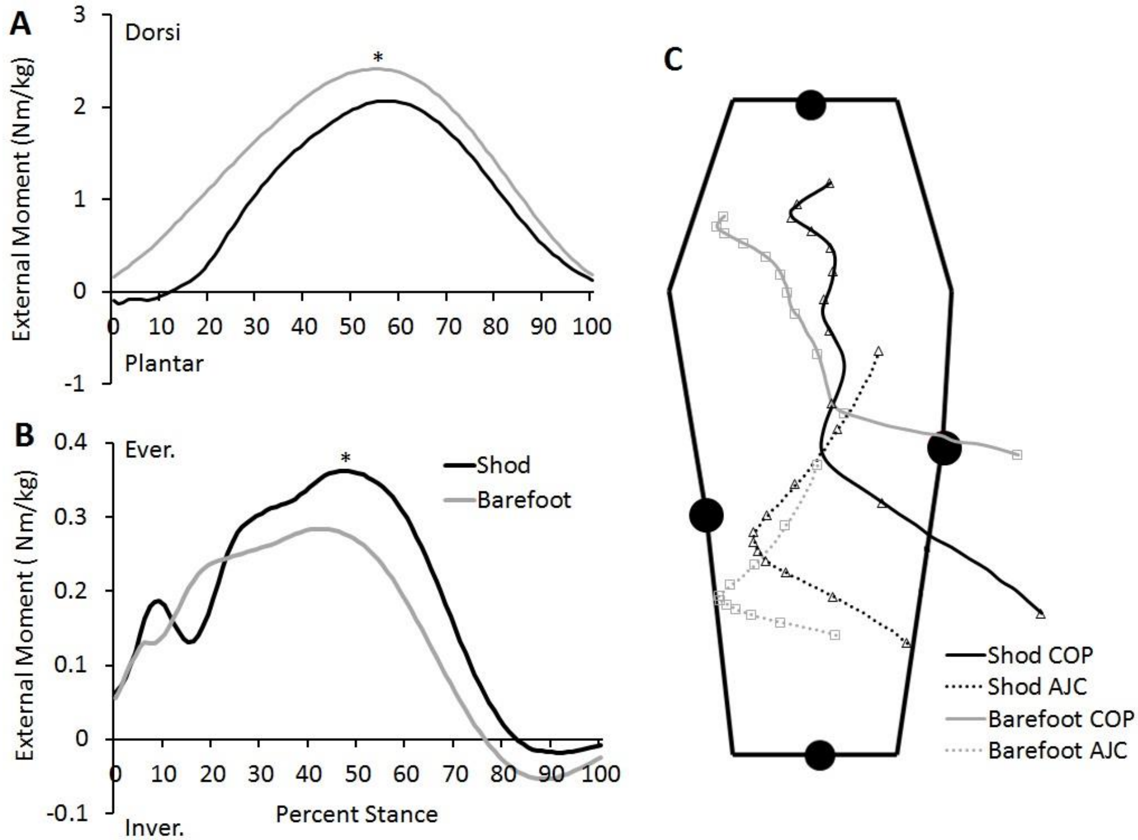

Figure 2.4. External dorsiflexion moment across stance (A), external eversion moment across stance (B), and a graphical illustration of the ankle joint center (AJC) and COP trajectories plotted within a rough outline of the foot (C). * indicates SH is significantly different than BF at $p < .05$.

<u>Discussion</u>

The goal of this study was to examine differences in COP trajectories and the resulting external eversion moment between SH and BF running. The COP_{AP} and COP_{ML} positions in the current study are similar to those previously reported using pressure platforms for both SH [95] and BF [49,94] running. Previous authors have reported methods for using force plates to measure the COP trajectory relative to the anatomic structures of the foot [113,114]. However, these studies have focused on aligning COP trajectories derived from force plates and pressure insoles, and have not examined differences between experimental conditions. The results of the current study provide further support to the force plate method and suggest this method is robust enough to detect small differences in COP trajectories between conditions.

These differences in COP trajectories between SH and BF running suggest COP related measures may be important to consider when examining both basic mechanical differences and relative injury risk between SH and BF running conditions. For instance, it has been reported that BF running is more metabolically efficient than SH running [120]. Divert et al. [121] and Perl et al. [122] have suggested this may be because BF results in greater storage of elastic energy then running SH. The increased external dorsiflexion moment observed during BF running supports this interpretation. In the absence of increased resistance from the triceps surae muscles, a higher external dorsiflexion moment would cause the ankle to move through an increased range of motion, and placing the Achilles tendon under greater stretch. Indeed, Squadrone et al.[123] reported increased ankle dorsiflexion range of motion when subjects ran BF compared to when they ran SH. However, this also raises questions regarding relative injury risk, especially as a habitually SH runner transitions into BF running. It has been reported that running with a forefoot strike results in higher forces in the Achilles tendon than running with a RFS, for both SH [124] and BF running [122]. If the Achilles tendon is not sufficiently resilient, the increased energy storage and force on each step could have a debilitating effect.

In addition to the possible metabolic advantages of BF running, it has also been suggested that BF running may reduce the tendency for the rearfoot to evert [116]. This is primarily based on reports that when running BF the foot makes contact with the ground in a less everted position [115] thereby shifting the COP laterally and reducing the eversion moment arm. While increased amounts or velocities of rearfoot eversion are often cited as biomechanical variables contributing to running injuries, there is little evidence in the literature supporting this relationship [98,125]. Even if eversion amounts or velocities are related to overuse injuries, the lack of difference in external eversion moments early in stance found in the current study suggest BF running may not be protective against these types of injuries. As discussed by Dixon [111] medial or lateral shifting of the COP may also be accompanied by medial or lateral shifting of the joint center, thus offsetting any changes in moment arms. This was observed in the current study, with the AJC being in a more medial position through stance in the BF condition compared to the SH condition. Therefore, even though the COP was located more laterally during the SH condition, there were no differences in the external eversion moment during early stance.

While there were no differences in COP_{ML} excursions between SH and BF conditions, the COP was located significantly more medially in the BF condition at all points except initial contact and 20% stance (Figure 2.3C). As the location of the COP shifts medially, pressures under the medial aspect of the foot increase. High pressures under the medial aspect of the foot have been observed in individuals who developed exercise related lower leg pain in two separate studies, one where subjects ran SH [95] and one where subjects ran BF [94]. In this context, a more medial COP location could be a potential concern for lower limb injury risk while BF running. However, a more medial COP location at push off also implies a greater use of the hallux, and a push off under the hallux has been reported as normal for young adults with no history of injury during both BF jogging [49] and BF walking [126]. Thus the BF COP trajectory observed in the current study may represent a more “natural” use of the foot, something that is not required when running SH.

One limitation to this study was that all subjects were habitually SH runners who were participating in an acute bout of BF running. It is unknown if the differences in COP trajectories would have remained had the subjects been given more time to adapt to BF running. Similarly, it is unknown whether these differences would still have been observed had habitually BF runners been used as subjects. Finally, it should be noted that subjects were included in this study only if they naturally transitioned from a RFS when SH to a MFS when BF. While this transition in foot strike patterns is commonly observed when habitually SH runners are asked to run BF [127,128] other data from our laboratory suggests this might not always be the case. Thus, the differences in COP trajectories between SH and BF running reported in this study may be dependent on whether or not subjects change their foot pattern when switching from SH to BF running.

In conclusion, this study used a force plate to examine differences in COP trajectories and resulting external dorsiflexion and eversion moments when habitually SH runners run BF. The results support using a force plate to measure the COP trajectories relative to the anatomic structures of the foot and suggest this method is capable of detecting subtle differences between experimental conditions. The results also suggest there are differences in the COP trajectory between SH and BF running and, as such, future studies on mechanical and epidemiological differences between SH and BF running may find the COP trajectory to be a useful to examine.

Bridge

Chapter II examined a method for using a force plate to quantify the COP trajectory relative to the anatomic structures of the foot. The data presented in Chapter II demonstrates this method is capable of detecting subtle differences in COP trajectories between experimental conditions such as shod and barefoot running. Based on the nature of these differences, future studies on running mechanics and injuries may find the COP trajectory a useful parameter. This method will be used again in Chapter III to see if the COP trajectory deviations could be a biomechanical parameter indicative of prolonged pronation. The method will also be used in Chapter IV to compare the COP trajectories between healthy individuals and those currently symptomatic with medial tibial stress syndrome or Achilles tendinopathy.

CHAPTER III

BIOMECHANCIAL MARKERS OF PROLONGED PRONATION

This chapter contains co-authored material and was developed by Dr. Li-Shan Chou and James Becker. Dr. Chou contributed to the development and refinement of the methodology as well as provided critiques and editing advice for the manuscript, while Mr. Becker was responsible for conceptual development, development of the protocol, data collection and analysis, as well as all the writing.

Introduction

It is estimated between 27% and 75% of all runners sustain an overuse injury in any one year [2,4,5]. Mechanisms for these injuries can be grouped into three types: training errors such as increasing volume or intensity too quickly; intrinsic factors related to the individual such as their anatomy or alignment; and individual biomechanics [125]. One commonly implicated biomechanical factor is abnormal pronation of the foot.

Pronation is a complex motion consisting of dorsiflexion at the talocrural joint, eversion at the subtalar joint, and abduction of the forefoot across the transverse tarsal joints [129]. Given the anatomy of the foot, movement at one joint rarely occurs in isolation. Rather, eversion of the calcaneus causes the transverse tarsal joints to become parallel, and allows the foot to become soft and flexible. This configuration allows the foot to adapt to uneven terrain as well as providing some shock absorbing capabilities. During the second half of stance, as the calcaneus starts inverting, motion across the transverse tarsal joints reverses. As the axes converge, the foot becomes a rigid lever, allowing for an efficient transmission of force during push off [34].

Excessive amounts or rates of pronation have been suggested as etiologic factors for several common running injuries including medial tibial stress syndrome (MTSS; [59,62]), Achilles tendinopathy [57,58,85], and plantar fasciitis [59–61]. However, the relationship between the excessive pronation and injury is not supported conclusively in the literature. While several studies have reported greater amounts of pronation in individuals with MTSS [24,25,33,59,62–66,95] others have reported no differences in the amount of pronation between injured and uninjured individuals [86–89]. A similar pattern has been observed with Achilles tendinopathy, with some authors reporting injured individuals demonstrate greater amounts or velocities of pronation than healthy controls [46,57,58], while others have reported no difference between injured and healthy subjects [90,91]. Plantar fasciitis also shows this pattern with some authors reporting increased amounts of pronation in injured compared to non-injured individuals [59–61] while others reported rearfoot kinematics are not different between injured and healthy individuals [42]. Further complicating the issue is a study by Hreljac et al. [96] which found that injury free runners demonstrated greater amounts and velocities of pronation than injured runners. The discrepancy in the literature suggests alternative hypothesis regarding the relationship between pronation and running injuries are warranted.

One such hypothesis centers on the view that it is not the amount or velocity of pronation which is necessarily important. Rather, it is the duration of stance phase where the foot remains in a pronated position that is more important for the prediction of injury development. As the calcaneus starts to invert during mid-stance the axes of the transverse tarsal joints converge, turning the foot into a rigid lever and allowing for efficient transmission of force during push off. However, if pronation is delayed beyond mid-stance (prolonged pronation) then extra effort may be required from both the intrinsic and extrinsic musculature of the foot to stabilize the foot during push off [130]. When repeated on every step this extra muscular effort may be enough to eventually lead to injury.

The potential implications of prolonged pronation have been discussed numerous times in the literature [13,41,98–103]. However, parameters reported in these studies usually reflect the amount of pronation, velocity of pronation, or time until peak pronation is reached, rather than the actual duration of pronation. The closest thing to an actual measure of pronation duration is found in two studies which use a measure termed period of pronation, a representation of the amount of time the foot remains in a pronated position through stance phase [104,105]. While these two studies examined how the period of pronation changes with different shoe conditions, they did not provide information on what constitutes excessive values for the period of pronation, nor did they examine whether individuals with longer periods of pronation also demonstrate greater amounts or velocities of pronation. Additionally, there is currently no literature describing any other biomechanical parameters, either those which can be measured in clinical settings or lab settings, which may identify individuals with longer periods of pronation.

Clinicians appear confident they can identify individuals presenting with prolonged pronation in clinical settings [13,131]. Since prolonged pronation is hypothesized to be related to several common running injuries, knowledge of specific biomechanical parameters identifying individuals with prolonged pronation may help in identifying individuals at greater risk for overuse injuries. Therefore, the purpose of this study was to identify specific biomechanical markers which distinguish individuals with prolonged pronation. Specifically examined were the kinematics of the leg and foot and the trajectory of the center of pressure (COP) underneath the foot. It was hypothesized that traditional measures of pronation such as the amount, velocity, or time to peak eversion would not be different between individuals who do and do not demonstrate prolonged pronation while measures such as the period of pronation would differentiate these groups.

Methods

Subjects

An a priori power analysis was conducted based on available data in the literature. Since some degree of kinematic and kinetic asymmetry is common even in healthy runners [118], and since running injuries occur either unilaterally or bilaterally, it was decided a priori that individual limbs would be analyzed rather than subjects. Based on subjective diagnoses in a clinical setting, prolonged pronation may be present in up to thirty percent of runners [131]. Bates et al. (1979a) reported an average period of pronation of 61.8% stance (± 12.3%) in healthy uninjured runners. Assuming mean periods of pronation for prolonged pronators would be at least one and a half standard deviations above this value suggested at least 22 feet (17 non-prolonged pronators, 5 prolonged pronators) were required to adequately power this study (effect size = 1.5, α = 0.05, β = 0.80, allocation ratio = 0.333).

Since these were only estimates based on previous literature, additional subjects were recruited. A total of 20 individuals (sex: 14 men, 6 women; age: 22.7 ± 4.7 years; weekly mileage 59.3 ± 16.2 miles) participated in this study. All subjects were competitive distance runners, were currently healthy at the time of testing, and had not sustained a running related injury in the six months preceding testing. All subjects read and signed an informed consent approved by the University of Oregon Human Subjects Review board.

Experimental Protocol and Instrumentation

Subjects underwent a clinical exam documenting general lower limb alignment, mobility, and flexibility. The exam was performed by one of two collaborating clinicians, both of whom have significant experience treating injured runners. Contents of the exam are shown in Table 3.1 and specific procedures for taking the measurements have been previously detailed in the literature [13,132]. Participants' arch heights were

Table 3.1. Variables measured during the clinical exam.

Variable	Description
Static leg varus angle (STVA)	Varus angle between the long axis of tibia and the floor (°).
Static ankle dorsiflexion (SADF)	Active ankle dorsiflexion range of motion (°).
Static ankle plantarflexion (SAPF)	Active ankle plantar flexion range of motion (°).
Static hip internal rotation ($SHIR_{ROM}$)	Passive prone hip internal rotation range of motion (°).
Static hip external rotation ($SHER_{ROM}$)	Passive prone hip external rotation range of motion (°).
Static hamstring flexibility (HAM)	Popliteal angle measured with the subject lying supine and thigh segment pointing vertically (°).
Static quadriceps flexibility (QUAD)	Angle between the thigh and shank segments with subject lying prone and actively bringing heel as close to buttocks as possible (°).
Static subtalar inversion (STI_{ROM})	Passive inversion range of motion at the subtalar joint (°).
Static subtalar eversion (STE_{ROM})	Passive eversion range of motion at the subtalar joint (°).
Static 1st MPJ flexibility (MPJ)	Active range of motion at the 1st metatarsophalangeal joint (°).

also measured using the arch height index [133].

Following the clinical exam thirty nine retro-reflective markers were attached to specific body landmarks. See Figure 2.1 for details on foot marker placement. A static trial was collected from which anatomic coordinate systems for the pelvis, thigh, shank, and rearfoot segments were established according to ISB recommendations [97]. Subjects then participated in a two part gait analysis consisting of an observational video analysis and a full three-dimensional motion capture session. For the observational gait analysis subjects ran on a treadmill at self-selected pace approximating their easy training run. After a five minute warm up, a 30 second sample of their running gait was recorded with a high definition video camera (JVC Corp., model: GC-PX10) sampling at 60 Hz.

For the three-dimensional motion capture session subjects ran laps around a short track in the laboratory (see Figure 2.1). Their whole body motion was recorded using a 10-camera motion capture system (Motion Analysis Corp., Santa Rosa CA) sampling at 200 Hz. Ground reaction forces were measured with three force plates (AMTI, Watertown, MA) located in series in the capture volume and sampling at 1000 Hz. Subjects ran continuous laps until a minimum of 8 clean trials per foot were recorded. A trial was deemed clean if the foot landed in the middle of the force plate with no visible signs the subject altered their stride to target the force platform.

Data Analysis

Video from the observational gait analysis was slowed down to 1/5th speed using video editing software (VirtualDub, www.virtualdub.org) and then examined by the clinicians. The clinicians classified each foot as demonstrating prolonged pronation or not based on the alignment of the long axis of the tibia and the vertical bisection of the shoe's heel counter at the moment of heel off. When these axes aligned the foot was classified as not showing prolonged pronation and when they diverged the foot was classified as demonstrating prolonged pronation (Figure 3.1). Each clinician analyzed all videos independently and agreement between the two clinicians was examined by calculating a kappa statistic based on their initial classifications. Where the clinicians disagreed on the classification, both clinicians viewed the video simultaneously and discussed the classification until agreement could be reached. Thus, all feet were classified as either demonstrating (PP) or not demonstrating prolonged pronation (NPP), forming two groups from which subsequent statistical analyses could be performed.

Three dimensional marker trajectories and ground reaction forces were filtered with low pass, fourth order, zero lag Butterworth filters using cutoff frequencies of 8 Hz and 50 Hz, respectively. A fifty Newton threshold in the filtered vertical ground reaction

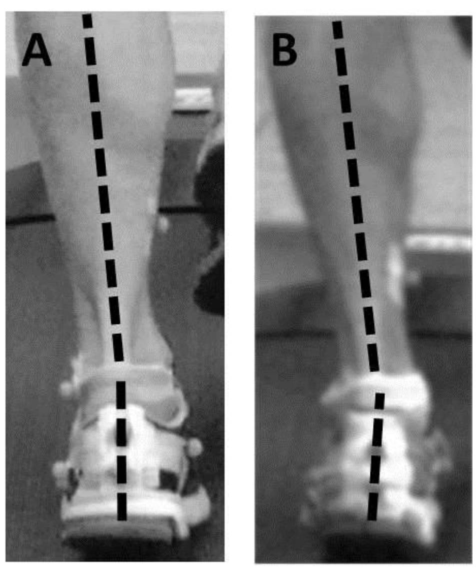

Figure 3.1. Examples of subjects who do not (A) and do (B) demonstrate prolonged pronation based on the video analysis. The snap shots were taken two frames after the first heel rise was visible in the video. Note the different angle formed when comparing the bisection of the tibia and the vertical axis of the shoe heel counter.

force were used to establish the instants of foot contact and toe off [35]. Foot strike pattern was determined using the strike index [18]. Filtered marker trajectories and the anatomic coordinate systems established during the static trial were used to calculate joint angles across stance using a Z-X-Y (flexion-extension, inversion-eversion, internal-external rotation) Cardan angle sequence. From the joint angles eighteen specific variables describing the orientation and movement of the leg and foot segments (Table 3.2) were extracted using custom LabView (National Instrument; Austin, TX) software.

The COP was initially calculated in the global coordinate system (GCS). This coordinate system is referenced to the fixed axes of the force plates or room. However, to be truly informative regarding injury risk or performance factors the COP should be referenced to the anatomic structures of the foot. Therefore, at each instant during stance, a local foot coordinate system (FCS) describing the orientation of the foot relative to the GCS was established and the COP was transformed from the GCS to the FCS. This allowed examination of the COP relative to the anatomical structures of the foot, accounting for any toe out during stance. It also allowed comparison of COP trajectories across multiple foot strikes even if they occur on different force plates (see Figure 2.2).

Once expressed in the FCS, COP trajectories were normalized to 100% stance and the anterior-posterior (COP_{AP}) and mediolateral (COP_{ML}) positions at each 10% stance interval were calculated for comparison between conditions. To examine if or how any observed differences in COP trajectories between PP and NPP groups may contribute to injury risk the external moments in the frontal plane from the ground reaction force (GRF) at the ankle and knee were calculated. Prior to calculating the moments the GRF, ankle joint center (AJC), and knee joint centers (KJC) were all transformed to the FCS.

Table 3.2. The eighteen kinematic variables extracted for comparison between prolonged and non-prolonged pronators.

Foot and Ankle Variables	Description
Eversion excursion (EE)	The total amount of rearfoot eversion from the instant of foot contact until peak eversion measured in degrees (°).
Time to Peak Eversion (TTPE)	Time from foot contact until peak eversion is reached measured in milliseconds (ms).
Average Eversion Velocity (AVE)	Average eversion velocity between time of foot contact and peak eversion measured in degrees per second (°/s).
Maximal Instantaneous Eversion Velocity (MEV)	Maximal instantaneous eversion velocity between time of foot contact and peak eversion measured in degrees per second (°/s).
Time within 25% Maximal Eversion (Tw/in25%)	Time spent within 25% of maximal eversion measured as percentage of stance (%).
Period of Pronation (PerP)	Time during which the alignment between the rearfoot and shank segments indicates the foot is in an everted position measured as a percentage of stance (%).
Eversion at Heel Off (EHO)	The relative orientation between the rearfoot and the shank at the instant of heel off measured in degrees (°).
Time to Heel Off (TTHO)	Time from initial foot contact until heel off measured in milliseconds (ms).
Average Inversion Velocity (AIV)	Average inversion velocity between the time of peak eversion and toe off measured in degrees per second (°/s).
Maximal Instantaneous Inversion Velocity (MIV)	Maximal instantaneous inversion velocity between the time of peak eversion and toe off measured in degrees per second (°/s).
Fore foot Abduction Excursion (FFAE)	The total amount of forefoot abduction from the instant of foot contact until peak abduction. Measured in degrees (°)
Peak Dorsiflexion (PDF)	Peak ankle dorsiflexion during stance phase measured in degreed (°).
Dorsiflexion Excursion (DFE)	The total amount of dorsiflexion from initial foot contact until peak dorsiflexion measured in degrees (°).
Tibia Varus Angle at Contact (TVC)	The varus angle of the tibia relative to a vertical line at the instant of foot contact measured in degrees (°).
Hip Variables	**Description**
Hip Abduction at Contact ($HADD_{contact}$)	Hip adduction angle at the instant of foot contact (°).
Hip Abduction Excursion ($HADD_{excur}$)	Hip adduction range of motion from contact until peak adduction (°).
Hip Internal Rotation at Contact ($HIR_{contact}$)	Hip internal rotation angle at the instant of foot contact (°)
Hip Internal Rotation Excursion (HIR_{excur})	Hip internal rotation range of motion from contact until peak internal rotation (°).

Thus, all external moments were expressed in the FCS. Peak external ankle and knee moments were compared between groups. As with the COP trajectories, the anterior-posterior and mediolateral locations of the AJC and KJC, and the external ankle and knee moments, were examined in increments of 10% stance.

Statistical Analysis

Initial inter-rater agreement between the two clinicians was evaluated with a kappa statistic. Differences between NPP and PP groups on clinical (Table 3.1) and kinematic (Table 3.2) measures were evaluated using a 2x2 analysis of variance (ANOVA), with pronation group and foot strike pattern being the two independent variables. Pronation group was a categorical variable with two levels, prolonged and non-prolonged with assignment based on the clinician's classification of each subject. Foot strike pattern was included as a second independent variable since many of the kinematic variables examined in this study vary with foot strike pattern [134,135]. Foot strike was treated as a categorical variable with two levels, rearfoot (RFS) or mid/forefoot strike (M/FFS), classified based on SI values less than 33% or greater than 33%, respectively. Arch height index and running speed were entered covariates since several of the kinematic variables vary with foot structure [136] and subjects ran at self-selected speeds. Given the number of comparisons performed, $\alpha \leq .01$ was used for determining statistical significance in an attempt to reduce the risk of a type-I error. Effect sizes (Cohen's f) were calculated for all statistically significant differences to aid in the interpretation of results. Effect sizes of 0.1 - 0.25, 0.25 – 0.40 and > 0.40 were used to indicate small, medium, and large effects, respectively [137].

To gain further insight into which clinical and kinematic variables best predicted group membership, all variables where comparisons between NPP and PP groups resulting in a main effect of group at $p \leq .20$ were considered for entry into a forward stepwise binary logistic regression. Prior to entry into the regression model, colinearity between variables was assessed with a bivariate correlation analysis. When variables demonstrated significant correlation ($p < .05$) with another variable, only one was entered into the regression model. Alpha levels of .05 and .01 were used as criteria for entry and removal from the regression model, respectively.

The influence of individual predictor variables on group classification was assessed using the methods described by Pohl et al. [67]. For each predictor variable, the mean value for the NPP group was considered a low risk value. Scores representing a high risk value were calculated as one standard deviation above or below the mean of the NPP group, as appropriate. Odds ratios were then sequentially calculated from the final regression equation, starting with all variables being entered at the low risk values. On the next iteration one predictor variable was entered at the high risk value while the rest remained at the low risk values. Iterations continued until the odds ratios were calculated with all predictor variables at high risk values.

A 2x2 (foot strike x pronation group) ANOVA was also used to examine differences in COP trajectories and external moments between PP and NPP groups. An $\alpha \leq .05$ was used for determining statistically significant differences in the discrete COP related variables (COP_{AP} and COP_{ML} excursions, the most medial location of the COP, and percent stance at which the most medial COP location occurs) and the peak external ankle and knee moments. However, a Bonferroni correction was applied for comparing COP_{AP}, COP_{ML}, AJC, KJC, external ankle, and external knee moments at each 10% stance increment. Therefore, for these six variables an $\alpha \leq .0045$ (.05/11) was used to determine statistical significance. Again, effect sizes (Cohen's f) were calculated to help interpret statistically significant differences.

Except for effect size calculations, all statistical tests were performed using Statistical Packages for the Social Sciences (SPSS; IBM Corp., Armonk NY) version 18. Effect size calculations were performed using the G*Power 3.1 software [138].

Results

Clinical and Kinematic Variables

The kappa statistic for the agreement between the two clinicians on the initial classification of subjects was 0.73, suggesting acceptable agreement between clinicians in classifying subjects. After revision the clinicians classified 21 limbs (12 RFS, 9 M/FFS) in the NPP group and 19 (13 RFS; 6 M/FFS) limbs in the PP group.

Neither the amount of pronation (EE; NPP: 11.8 ± 4.1°, PP: 12.4 ± 4.7°, p = .544) nor the maximal velocity of pronation (MEV; NPP: 315.7 ± 120.4 °/s, PP: 370.8 ± 154.4 °/s, p = .224) were different between groups. From the 10 clinical exam measures and 18 kinematic variables, only four variables (PerP, EHO, STVA, and $SHIR_{ROM}$) demonstrated a significant main effect of group (Figure 3.2). Effect sizes (Cohen's f) for observed differences between PP and NPP groups for these four variables were 0.572, 0.618, 0.548, and 0.449, for Per_P, EHO, STVA, and $SHIR_{ROM}$, respectively, all indicative of large effects [137].

There was a significant foot strike pattern x group interaction for DFE (p = .002). For subjects who used a RFS, individuals in the PP group had greater DFE (M = 19.6° ± 1.4°) than those in the NPP group (M = 14.4° ±1.5°). However, for subjects who used a M/FFS, individuals in the PP group had less DFE (M = 23.5° ±2.0°) than those in the NPP group (M = 29.5 ± 1.6°).

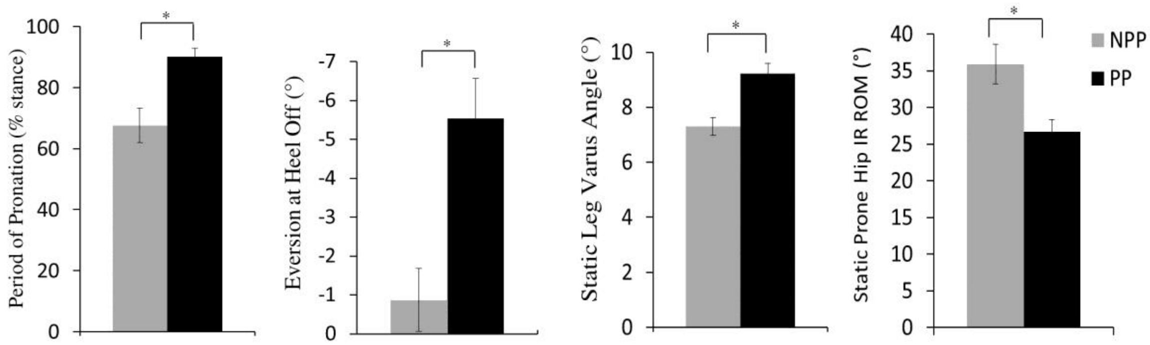

Figure 3.2. The four variables which were significantly different between NPP and PP groups in the 2x2 ANOVA. * indicates $p < .01$

Logistic Regression

In addition to the four variables shown in Figure 3.2, four other comparisons resulted in main effects for group with *p* values < .2 and were therefore considered for entry into the regression model (Table 3.3). Examination of the bivariate correlations revealed TVC was significantly correlated with STVA ($r = .433$, $p = .006$), and therefore only STVA was retained. Additionally, since the goal of the logistic regression was to identify additional clinical or biomechanical markers that could classify individuals with prolonged pronation, both Per_P and EHO were not included in the regression model

Table 3.3. The four variables, in addition to those shown in Figure 2.4, which were not significantly different between groups at the $p < .01$ level, but which did have $p < .2$ so were considered for inclusion in the regression model.

Clinical Exam Variables			
Variables	NPP Group	PP Group	*p*
$SHER_{ROM}$ (°)	25.9 (±6.4)	29.3 (±5.1)	.074
Kinematic Variables			
Variables	NPP Group	PP Group	*p*
MIV (°/s)	138.1 (±57.4)	112.8 (±41.9)	.123
TVC (°)	7.9 (±3.2)	9.7 (±2.2)	.046
HIR_{excur} (°)	5.5 (±2.6)	8.7 (±5.6)	.034

since Per_P essentially directly measures the duration of pronation [104,105] and EHO was the measure used by the clinicians to differentiate groups in the observational gait analysis. Thus, these two variables should differentiate groups and their inclusion in the regression model would not yield any additional insights. Therefore the final variables entered into the forward stepwise binary logistic regression were STVA, $SHIR_{ROM}$, HIR_{excur}, $SHER_{ROM}$, and MIV.

After three steps, a logistic regression model containing the variables of STVA, HIR_{excur}, and $SHIR_{ROM}$, entered in that order, was able to correctly classify 94.9% of the limbs into the NPP or PP groups. The overall model was significant (χ^2 = 29.215, df = 3, p < .001) and the Nagelkerke R^2 value was .80, showing that 80% of the variance between NPP and PP groups was explained by STVA, HIR_{excur}, and $SHIR_{ROM}$. The inclusion of MIV, or $SHER_{ROM}$ did not improve the model. Odds ratios for STVA, HIR_{excur}, and EHO are shown in Table 3.4.

Assuming the mean values for the NPP group represented a "low risk" condition and one standard deviation above or below these values represented "high risk" conditions [67], sequentially evaluating the final regression equation resulted in the following combinations. With all variables at "low risk" values the odds of being

Table 3.4. Output from the forward stepwise binary logistic regression reporting the beta coefficients (β), standard error of the coefficients (S.E.), p-value (*p*), odds ratios, and 95% confidence intervals for the odds ratios (95% CI)

Variable	β	S.E.	*p*	Odds ratio	95% CI	
					Lower	Upper
STVA	1.231	0.467	.008	3.424	1.371	8.554
HIR_{excur}	0.490	0.467	.043	1.632	1.016	2.621
$SHIR_{ROM}$	-0.142	0.790	.040	0.867	0.743	1.012
Constant	-9.386	4.656	-	-	-	-

classified as a PP were 0.01. As STVA, HIR_{excur}, and $SHIR_{ROM}$ were each entered at "high risk" values, the odds of being in the PP group rose to 0.4, 1.4, and 6.7, respectively (Figure 3.3).

Center of Pressure Trajectories

The COP was located significantly more medial at each time point from 10% through 90% of stance in the PP group compared to the NPP group (Figure 3.4B). The effect sizes (Cohen's f) for these differences were all moderate to large, ranging from 0.383 at 20% to 0.529 at 70%, with the average effect size across all points being 0.469 In the A/P direction, there were no main effects of pronation group at any time point. However, for both NPP and PP groups the COP was located significantly more anteriorly at initial contact, 10%, and 20% of stance in subjects who used a M/FFS compared to

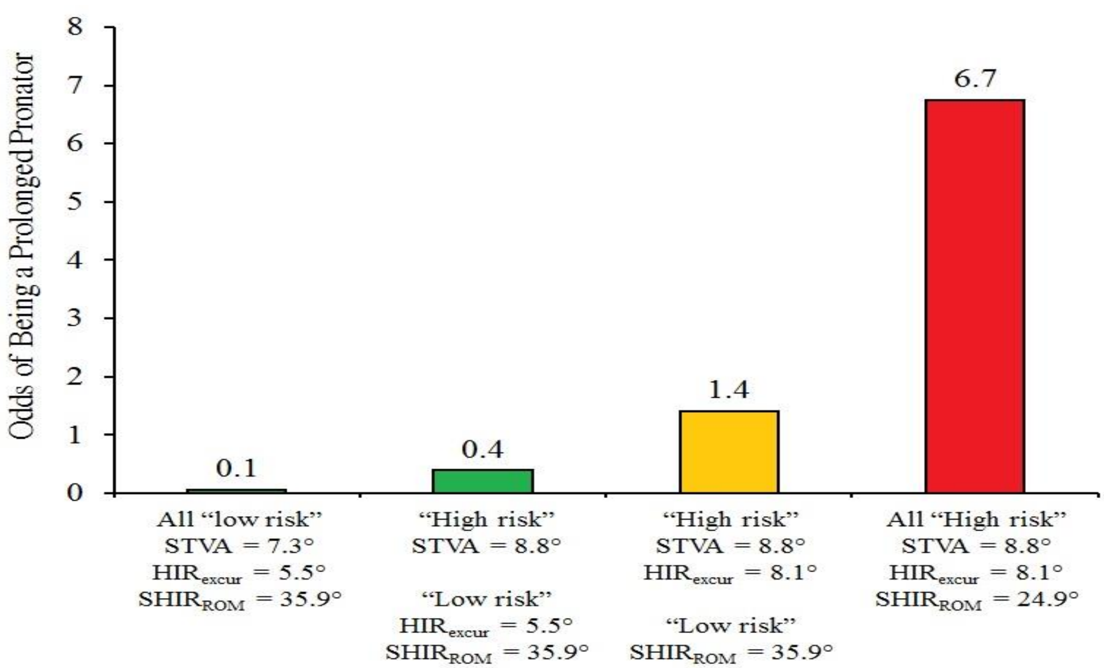

Figure 3.3. Illustration showing the odds ratios for being in the PP group when sequentially evaluating the regression equation.

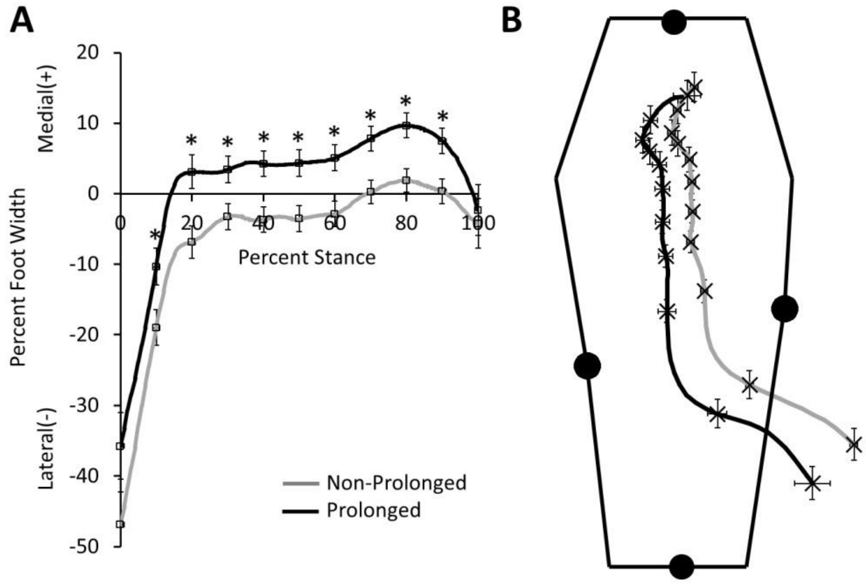

Figure 3.4. M/L location of the COP (A) and the trajectory of the COP plotted in an outline of the foot based on marker locations (B).

those who used a RFS.

There was a significant main effect of foot strike pattern for COP_{AP} excursion with subjects who utilized a RFS demonstrating greater AP excursion than subjects who utilized a M/FFS (Figure 3.5A). However, there was no main effect of pronation group (p = .479). There was also a significant main effect of foot strike pattern for COP_{ML} excursion; however, for this variable, subjects who utilized a RFS demonstrated reduced COP_{ML} excursion than subjects who utilized a M/FFS (Figure 3.5B). Again, there was no main effect of pronation group (p = .565). There was a main effect of pronation group for the most medial COP location, with the PP group having a more medial peak location of the COP than the NPP group (Figure 3.5C). However, the percent stance at which the most medial COP location occurred was not different between groups (Figure 3.5D).

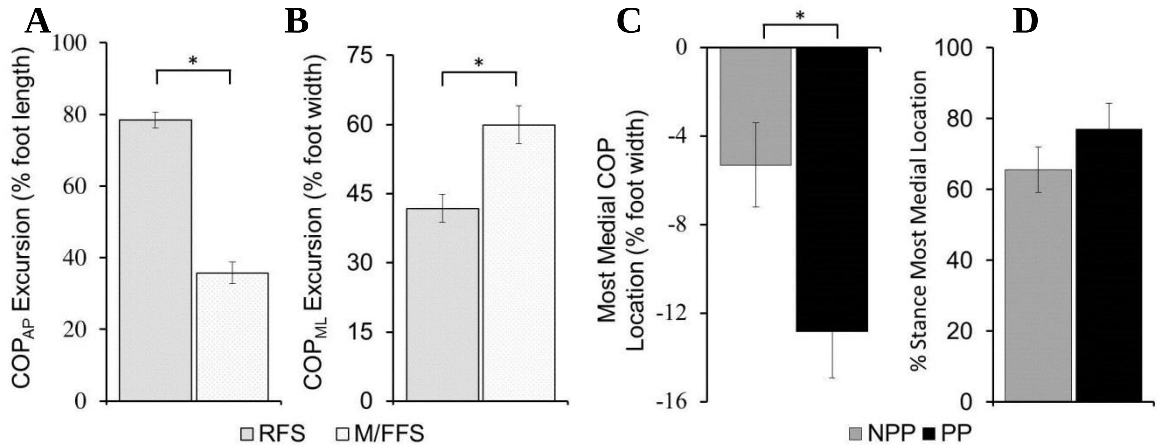

Figure 3.5. Differences in discrete COP trajectory parameters showing COP_{AP} (A) and COP_{ML} (B) excursions, the most medial location of the COP (C), and the percent stance at which the most medial location occurred (D). * indicates either significant main effects of foot strike pattern (A and B) or significant main effects of pronation group (C).

Effect sizes for foot strike differences in COP_{AP} and COP_{ML} excursion and most medial location of the COP were all large at 1.89 and 0.59, and 0.44, respectively.

Mediolateral positioning of the AJC was not different between PP and NPP groups at any time points (Figure 3.6C). For both groups the resulting external ankle moment was an eversion moment for the first 75% of stance, followed by an external inversion moment for the last 25% of stance. For both groups, the moment curve had a bimodal shape, with an initial peak followed by a local minima followed by the maximal peak (Figure 3.6A). Neither the initial nor maximal peaks were different between NPP and PP groups. However, the drop from the initial peak to the local minima was larger for the PP group than the NPP group (Figure 3.6B). There were no differences between groups at any of the 10% stance increments (Figure 3.6A).

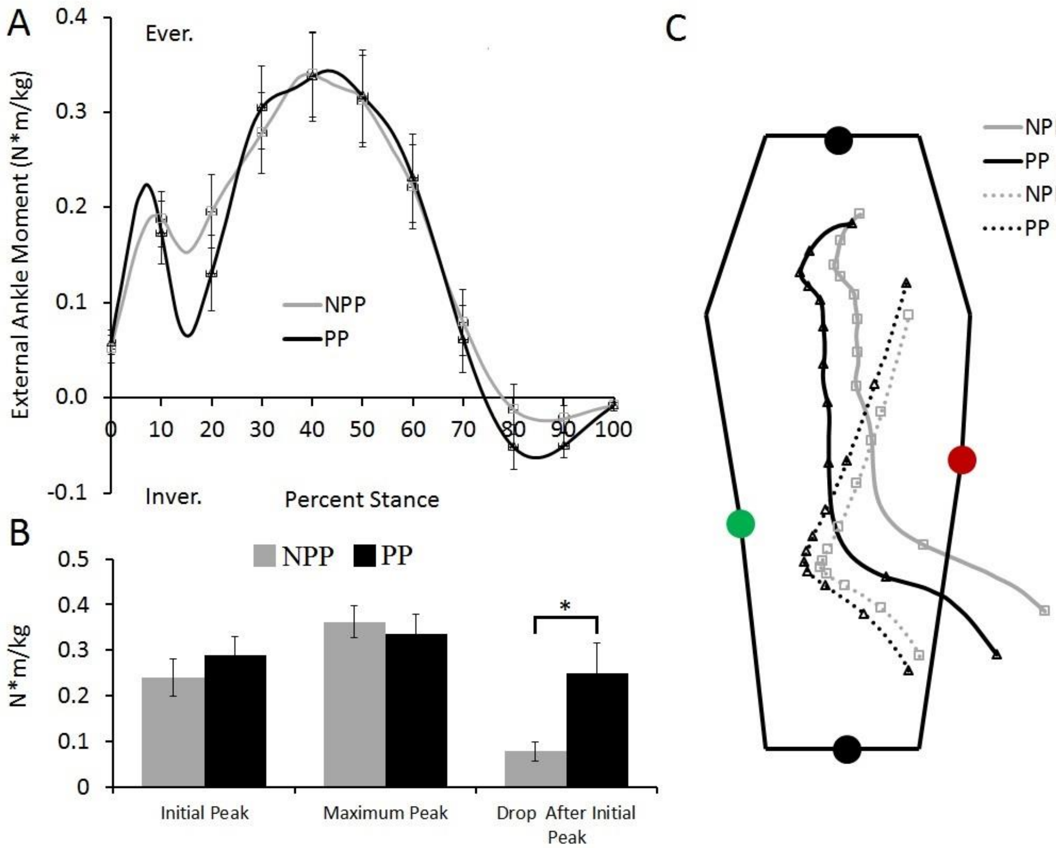

Figure 3.6. External ankle eversion moments (A), the magnitude of the initial peaks, maximal peaks, and drop to the inter-peak minima (B), and the location of both the AJC and COP across stance, plotted in an outline of the foot based on marker locations (C). * indicates significant difference at $p < .05$ level.

As with the AJC, mediolateral positioning of the KJC was not different at any point across stance (Figure 3.7C). The external frontal plant moment at the knee was an adduction moment for the duration of stance for both groups (Figure 3.7A). There were no differences between groups in the external adduction moment at any of the 10% stance increments. The peak external adduction moment and the percent stance at which the peak took place were also not different between groups (Figure 3.7B).

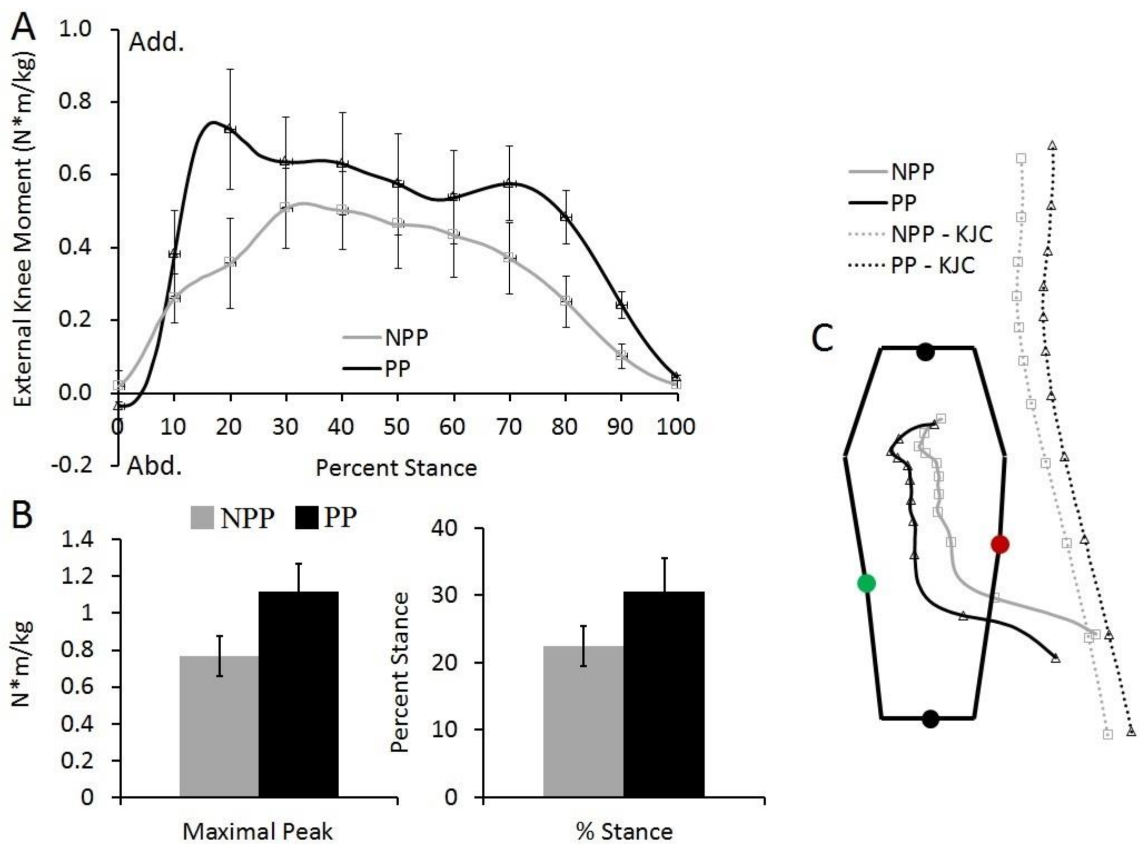

Figure 3.7. External knee adduction moments (A), magnitude of the maximal external adduction moment and the percent stance at which the maxima occur (B), and the location of both the KJC and COP across stance, plotted in an outline of the foot based on marker locations (C).

Discussion

This goal of this study was to identify biomechanical markers of prolonged pronation in distance runners. In this regards, the main findings of this study were that from the 10 clinical exam variables and 17 kinematic variables only four, Per_P, EHO, STVA, and $SHIR_{ROM}$, were significantly different between NPP and PP groups, however, for some variables there were additional foot strike dependent differences between groups. The study also found that a logistic regression using the variables HIR_{exur}, STVA, and $SHIR_{ROM}$ was able to correctly classify 94.9% of the limbs into either NPP or PP groups. Finally, the study found that individuals in the PP group had a more medial location of the COP for most of stance, with small differences observed in the external ankle eversion moment.

Of the four variables that were significantly different between NPP and PP groups (Figure 3.2), EHO and Per_P demonstrated the largest effect sizes, and as such could be considered the most robust biomechanical markers of prolonged pronation. Since the presence or absence of an everted position at heel off in the observational videos was the criteria used by the clinicians to classify individuals into either NPP or PP groups, it makes sense that EHO should be different between groups. However, that the biomechanical results supported the clinician's classifications is important for two reasons. First, it provides support for the validity of the process used to classify the subjects. While both clinicians were experienced in assessing and treating injured runners, the classification of subjects was still based on their professional opinion. Secondly, the agreement between the clinicians and the biomechanical data suggests appropriately trained clinicians are able to visually identify subtle biomechanical differences even in healthy individuals. This could be an important skill for identifying prolonged pronators in clinical settings lacking access to full biomechanical assessments.

While Bates and colleagues [104,105] have previously used Per_P to both examine the effects of orthotics on foot function and as a measure to describe basic foot function, outside these two studies, the measure has seen little to no use in the running literature. Given how the Per_P variable is calculated, one might reasonably assume it should differentiate individuals with prolonged pronation from those without prolonged pronation. However, until the current study, there has been no data in the literature to support this assumption. While the current study's finding that Per_P is larger in the PP group compared to the NPP group support the use of Per_P as an appropriate variable for quantifying prolonged pronation, specific values demarcating normal from prolonged Per_P values still need to be identified.

Since the Per_P values previously reported by Bates and colleages [104,105] for healthy runners (61.8% stance) and the mean Per_P values for the NPP group in this study (66.1 ± 5.6 %) were similar, one can be reasonably confident that a "normal" Per_P should be in in the range of 60 to 70 percent of stance. Exactly which values should be considered "prolonged" still needs to be determined. One possible method to do this could be to estimate a critical value which would generate a z-score greater than or equal to 1.96. Using the mean and standard deviation from the current study shows that a z-score of 1.96 is achieved with a Per_P value of 77.1% stance. Thus, anything over 77.1% of stance could be considered prolonged pronation. However, this only one possible method for identifying a cutoff for prolonged pronation and there are likely others which would work equally as well. Additionally, from a functional consequences perspective, there is likely not one single cutoff score which demarcates prolonged pronation from non-prolonged pronation. Rather it is most likely a continuum with higher Per_P values leading to worse functional outcomes.

While Per_P may be the most straightforward measure for identifying prolonged pronation, the pitfall of relying on it as the main classifier is that this variable would be hard to reproduce in clinical settings lacking access to a full motion capture system. However, the good agreement between the clinician's classifications and the biomechanical data for EHO and the results of the logistic regression analysis suggest that even in such clinical settings individuals with prolonged pronation may be identified with clinically available tools. For instance, the logistic regression suggested every 1° increase in STVA increased the odds of being in the PP group by 3.4, while every 1° decrease in $SHIR_{ROM}$ increased the odds by 0.9. Both of these values are easily measurable in clinical settings with a goniometer [13,132].

The finding that the amounts and velocities of pronation were not different between the NPP and PP groups suggests pronation duration is a unique parameter, and is not necessarily related to the amount or velocity of pronation. This is further supported by the fact that the mean values for the amount and velocities of pronation in the PP group are within "normal" ranges reported by numerous other studies (Table 3.5) and by the fact that a post-hoc analysis of correlations between Per_P, EE, AEV, and MEV, were all found to be non-significant. This finding is especially relevant for future studies on running injuries. While abnormal foot pronation is often suggested as a contributing factor to numerous common overuse injuries [57–60,62,85], the focus has traditionally been directed towards abnormal amounts or velocities or pronation. The results of this study suggest abnormal durations of pronation should be considered independently.

The argument for considering pronation duration in relation to common running injuries is also supported by the fact that PP group displayed several biomechanical characteristics that have been linked to overuse running injuries. For instance, the COP was located more laterally in the PP group than the NPP group at most time

Table 3.5. "Normal" amounts and velocities of pronation based on the current study and several others in the literature. Data from the current study is for the PP group.

Authors	Amount of Pronation (°)	Velocity of Pronation (°/s)
Current study	12.4 (± 4.7)	370.8 (±154.4)
Edington et al. [47]	16.4 (±2.3)	504.3 (±146.7)
McClay and Manal [38]	12.7 (±4.1)	-
Pohl et al. [29]	14.9 (±3.0)	-
Barnes et al. [55]	12.0 (±4.3)	-
Clarke et al. [56]	16.7 (±3.7)	532.0 (±173.0)
Morley et al. [57]	10.6 (±1.5)	-
Messier and Pittala [8]	13.5 (±1.4)	424.3 (±44.2)
Willems et al. [21]	18.1 (±4.5)	447.3 (±131.7)
McCrory et al. [26]	9.9 (±0.5)	374.3 (±23.1)
Hreljac et al. [30]	12.8 (±5.1)	334.0 (±126.0)

points across stance. Similar COP trajectory locations have been reported in both retrospective [95] and prospective studies [47] examining individuals who suffered from exercise related lower leg pain. The standing tibia varus angle relative to the ground (STVA) was larger in the PP group than the NPP group. A larger tibia varus angle relative to the ground suggests a greater genu varus position at the knee. Both retrospective [13,31] and prospective studies [27,139] examining relationships between anatomic alignment and common running injuries have reported higher genu varus in injured compared to non-injured runners. The PP group displayed lower $SHIR_{ROM}$ than the NPP group. Moen at al. [33] reported reduced hip internal rotation range of motion as being a prognostic indicator for the development of medial tibial stress syndrome.

While the above characteristics have already been observed in injured runners, the PP group demonstrated additional characteristics that could be linked to injury through theoretical examinations. For instance, the PP group had a larger STVA than the NPP group. A higher tibia varus angle means the foot will naturally assume a more varus position relative to the group. The ending combination is that an individual would have to pronate more simply to get their foot flat on the ground during midstance [13]. If they lack the requisite range of motion then this additional pronation may be achieve through compensatory strategies. Often these strategies involve additional dorsiflexion at the talocrural joint combined with abduction of the forefoot. This may partially explain the increased DFE observed in the PP subjects who used a RFS. While this motion in and of itself is not necessarily problematic, the increased dorsiflexion increases the strain in the Achilles tendon. High tendon strains are hypothesized to be one of the main contributing factors in the development of Achilles tendon injuries [82–84]. This may be especially problematic if an individual does not have sufficient dorsiflexion range of motion at the ankle. Indeed, reduced static dorsiflexion range of motion has been reported as a prospective predictor for the development of Achilles tendinopathy [23].

Another theoretical connection to injury mechanisms is the rapid change in the external eversion moment early in stance observed in the PP group. This rapid change in the external eversion moment has implications for the demands being placed on the musculature, since the muscles must produce force to create and internal inversion torque to counter the external eversion torque from the GRF. With such rapid forces the muscle must either produce force quickly, relax quickly, then produce force quickly again. While there is no evidence to suggest this is true, it is theoretically possible that this rapid contraction relaxation cycle, when repeated on each step of a run, may lead to earlier fatigue of the muscle and thus reduced ability to control the movement of the foot and leg.

At this point it should be emphasized that just because the PP group demonstrated characteristics that have also been reported in injured individuals or that theoretically could be related to injury development, does not necessarily mean these individuals will get injured. As the discrepancy over the role of excessive pronation clearly illustrates, simply because an individual demonstrates a given characteristic does not mean they will get injured. However, these findings do suggest that the duration of foot pronation should be considered in future studies on running injuries, as it may play an important role.

In summary, this study revealed there are several biomechanical markers which describe prolonged pronation in runners including a longer period of pronation, a more everted heel position at heel off, reduced static hip internal rotation range of motion, increased dynamic hip internal rotation range of motion during running gait, and a higher standing tibia varus angle relative to the floor. This study also demonstrated that the duration of pronation is its own unique variable and is distinctly different from the amounts or velocities of pronation. As such it should be considered an important variable in future studies on running injuries and running mechanics.

Bridge

Chapter III identified several biomechanical markers of prolonged pronation. Individuals demonstrating prolonged pronation also demonstrated characteristics which either have already been observed in injured runners, or can theoretically be linked to the development of running injuries. This chapter also provided evidence that the duration of pronation is distinctly different from either the amount or velocity of pronation.

However, it is still not clear if individuals with running injuries commonly attributed to excessive amounts or velocities of pronation may instead also demonstrate prolonged pronation. Thus, Chapter IV will examine whether individuals currently symptomatic with either Achilles tendinopathy or medial tibial stress syndrome demonstrate prolonged pronation.

CHAPTER IV

PROLONGED PRONATION IN INDIVIDUALS WITH ACHILLES TENDINOPATHY AND MEDIAL TIBIAL STRESS SYNDROME

This chapter contains co-authored material and was developed by Dr. Li-Shan Chou and James Becker. Dr. Chou contributed to the development and refinement of the methodology and provided critiques and editing advice for the manuscript; while Mr. Becker was responsible for conceptual development, development of the protocol, data collection and analysis, as well as all the writing. Additionally, Dr. Stan James and Mr. Robert Wayner both contributed by referring patients treated at their respective clinical practices.

Introduction

Running is an incredibly popular recreational and fitness activity in which an estimated 19 million Americans participate [1]. However, runners report disturbingly high annual injury rates between 25% and 75% [2,3,5]. Medial tibial stress syndrome (MTSS) and Achilles tendinopathy (AT) are two of the most common running related overuse injuries and their incidence has not changed despite over thirty years of research [13–15].

Despite different pathophysiology, the biomechanical movement patterns thought to be responsible for MTSS and AT are similar, with the most commonly cited parameters being excessive amounts or velocities of pronation [46,106,140]. From an anatomical perspective this makes intuitive sense. The structures most commonly implicated in the development of MTSS include the flexor digitorum longus [68,72], tibialis posterior [69,73], and soleus [68,72,74] muscles, as well as the deep crural fascia connecting these muscles to the tibia [68,75]. Greater pronation involves greater eversion of the calcaneus and lowering of the medial longitudinal arch. Based on the insertion sites of these muscles this motion would generate strain within these tissues which could then be transmitted through the fascia and result in higher forces at the bony insertions [68].

Similarly, greater calcaneal eversion would increase strain within the Achilles tendon, especially in the medial aspect of the tendon arising primarily from the soleus muscle [55]. The amount of strain present in the Achilles tendon has been reported as the major factor in determining time to tendon failure when the tendon is subjected to repeated loading cycles [82] and therefore is thought to play a significant role in the development of AT [80,81]. Greater calcaneal eversion will also increase the heterogeneity of strain distribution within the tendon [85], a condition which has been suggested as especially problematic for AT development [83,84].

Despite these anatomical considerations, there is conflicting evidence in the literature regarding whether excessive amounts or velocities of pronation actually lead to the development of these two injuries. While several studies have reported greater amounts of pronation in individuals with MTSS [24,25,33,59,62–66,95] others have reported no differences in the amount or velocity of pronation between injured and uninjured individuals [86–89]. Similarly, some authors have reported individuals with AT demonstrate greater amounts or velocities of pronation than healthy controls [46,57,58], while others have reported no differences between injured and healthy subjects [90,91]. These conflicting findings suggest alternative theories on movement patterns leading to these two common injuries are warranted.

One such hypothesis is that it is not necessarily the amount or velocity of pronation which is important but rather the duration the foot remains in a pronated position throughout stance. During the first half of stance, as the foot pronates, the axes of the transverse tarsal, cuneonavicular, and tarsometatarsal joints align allowing the foot to become soft and flexible [141]. However, during the second half of stance, as the foot supinates, the axes of these joint converge, turning the foot into a rigid lever for use during push off [141]. Therefore, if pronation is prolonged beyond midstance then push off will begin with a soft flexible foot. This configuration may require much greater effort from the intrinsic and extrinsic foot muscles to both stabilize the foot and to generate sufficient torque for push off [13].

While the hypothesis of prolonged pronation was first proposed in 1978 [13], to date studies on MTSS or AT have generally focused on the amounts, velocities, or time to peak pronation, rather than actual measures of the duration of pronation. Therefore the purpose of this study was to examine whether individuals currently symptomatic with MTSS or AT, two injuries commonly attributed to excessive amounts or velocities of pronation, demonstrate differences in the duration of pronation. It was hypothesized that compared to healthy matched controls, injured individuals would not demonstrate differences in the amount or velocities of pronation but rather, would demonstrate more prolonged pronation across stance.

Methods

Subjects

An a priori power analysis was conducted using data previously presented in the literature. Based on differences in rearfoot eversion excursion between individuals injured with MTSS and healthy controls reported by Messier and Pittala [59] it was concluded that a minimum of 10 individuals, 5 with MTSS and 5 healthy controls would be required adequately detect differences between these groups (effect size = 2.35, α = 0.05, β = 0.20). Similarly, based on differences in rearfoot eversion excursion between individuals injured with AT and healthy controls reported by Ryan et al. [58] it was concluded that a minimum of 24 individuals, 12 with AT and 12 healthy controls, would be required to adequately detect differences between these groups (effect size = 1.24, α = 0.05, β = 0.20).

Based on these estimates, a total of 21 injured individuals, 13 currently symptomatic with AT and 8 currently symptomatic with MTSS, were recruited for this study. Injured subjects were specifically diagnosed by and referred from the clinical practices of SJ and RW, the two clinicians participating in this study. In addition to diagnosing and referring patients, SJ and RW also ruled out any other injuries. For each injured subject, a healthy control subject (CON) was also recruited, thus a total of 42 individuals participated in this study. Controls were matched with injured individuals based on sex, weekly mileage, age, and foot strike pattern (Table 4.1). All control subjects ran at least 20 miles per week and had not sustained a running related injury within the previous six months. All subjects read and signed an informed consent approved by the University of Oregon Human Subjects Review board.

Table 4.1. Subject characteristics for individuals with Achilles tendinopathy (AT), medial tibial stress syndrome (MTSS), and matched controls (CON_AT or CON_MTSS).

Achilles Tendinopathy Subjects		
Variable	AT	CON_AT
Sex	9M, 4F	9M, 4F
Weekly Mileage (miles)	50.1 (± 15.1)	52.3 (± 14.7)
Foot Strike Pattern	7 RFS, 6 MFS	7 RFS, 6 MFS
Age (years)	37.6 (± 15.9)	32.6 (± 12.4)

Medial Tibial Stress Syndrome Patients		
Variable	MTSS	CON_MTSS
Sex	7M, 1F	7M, 1F
Weekly Mileage (miles)	27.5 (± 6.0)	28.8 (± 7.4)
Foot Strike Pattern	5 RFS, 3 MFS	5 RFS, 3 MFS
Age (years)	35.3 (± 11.8)	36.4 (± 9.7)

Experimental Protocol and Instrumentation

Subjects underwent a clinical exam documenting general lower limb alignment, mobility, and flexibility. The exam was performed by one of two collaborating clinicians (SJ or RW), both of whom have significant experience treating injured runners. Contents of the exam have already been discussed (see Table 3.1) and specific procedures for taking the measurements have been detailed in the literature [13,132]. Two variables measured in the clinical exam, standing tibia varus angle (STVA) and the prone static hip internal rotation range of motion ($SHIR_{ROM}$), have been previously identified as biomechanical markers of prolonged pronation while the other variables were used to assess whether there were any innate structural differences between injured and healthy runners, as such differences have been previously suggested to play a role in the development of both AT and MTSS [63,91].

Following the clinical exam 39 retro-reflective markers were attached to specific body landmarks. See Figure 2.1 for details on how foot markers were placed. A static trial was collected from which anatomic coordinate systems for the pelvis, thigh, shank, and rearfoot segments were established according to ISB recommendations [97]. Subjects then participated in a running gait analysis where their whole body motion was recorded using a 10-camera motion capture system (Motion Analysis Corp., Santa Rosa CA) while they ran continuous laps around a short track in the laboratory (see Figure 3.1). Ground reaction forces were measured with three force plates (AMTI, Watertown MA) located in series in the capture volume. Motion and ground reaction force data were sampled at 200 Hz and 1000 Hz, respectively. Subjects ran continuous laps until a minimum of 8 clean trials were recorded. A trial was deemed clean if the foot landed in the middle of a force plate with no visible signs the subject altered their stride pattern to target the force platform. For the AT and MTSS patients, their involved limb was used while the matching limb was used for control subjects. In cases where an individual was symptomatic bilaterally the right limb was used.

Data Analysis

Three dimensional marker trajectories and ground reaction forces were filtered with low pass, fourth order, zero lag Butterworth filters using cutoff frequencies of 8 Hz and 50 Hz, respectively. A fifty Newton threshold in the filtered vertical ground reaction force were used to establish the instants of foot contact and toe off [18]. Foot strike pattern was determined using the strike index [18]. Filtered marker trajectories and the anatomic coordinate systems established during the static trial were used to calculate joint angles across stance according to the ISB recommendations [97].

From the joint angles eight specific variables describing the orientation and movement of the leg segments (Table 4.2) were extracted using custom LabView (National Instruments, Austin TX) software. Three of these variables (Per_P, EHO, and HIR_{excur}) were chosen since they have been previously identified as biomechanical markers of prolonged pronation (see Chapter III). PSHO was calculated since it provides insight into whether the timing of heel off is similar between groups. The last four

Table 4.2. The eight kinematic variables extracted for analysis and definitions for how they were calculated or measured.

Variable (Abbreviation)	Definition
Period of pronation (Per_P)	The time the foot is in a pronated position across stance. Calculated based on when the rearfoot crosses 0° of eversion early in stance as the foot pronates to when it re-crosses 0° of eversion late in stance as the foot supinates.
Eversion at heel off	Eversion of the rearfoot at the instant the heel starts lifting off the ground.
Dynamic hip internal rotation range of motion (HIR_{excur})	Hip internal rotation range of motion during the stance phase of running measured from the position at touchdown until peak internal rotation.
Percent stance of Heel Off (PSPO)	The percent stance when the heel first rises off the ground. Calculated based on the vertical velocity of the heel marker.
Eversion excursion (EE)	The amount of rearfoot eversion from touchdown until peak eversion.
Time to peak eversion (TTPE)	Time until peak eversion is reached.
Average eversion velocity (AEV)	Average eversion velocity between touchdown and peak eversion.
Maximal eversion velocity (MEV)	Maximal instantaneous eversion velocity between touch down and maximal eversion.

variables (EE, TTPE, AEV, and MEV) were chosen as these are variables traditionally reported in studies examining biomechanical factors contributing to AT or MTSS.

In addition to the kinematic variables, the following kinetic variables were calculated: peak propulsive forces (PPF) and propulsive impulses (PI) from the anterior-posterior ground reaction force, peak vertical force (PVF) from the vertical ground reaction forces, and the center of pressure (COP) trajectory. Forces and impulses were normalized based on subject's body mass. The force and impulse variables were chosen as they provide insight into the dynamics of a subject's push off while the COP trajectories were chosen as differences in these trajectories have also been previously identified as biomechanical markers of prolonged pronation (see Chapter III).

The COP was initially calculated in the global coordinate system (GCS). This coordinate system is referenced to the fixed axes of the force plates or room. However, to be truly informative regarding injury risk or performance factors the COP should be referenced to the anatomic structures of the foot. Therefore, at each instant during stance, a local foot coordinate system (FCS) describing the orientation of the foot relative to the GCS was established and the COP was transformed from the GCS to the FCS. This allowed examination of the COP relative to the anatomical structures of the foot, accounting for any toe out during stance. It also allowed comparison of COP trajectories across multiple foot strikes even if they occur on different force plates (see Figure 2.2). Once expressed in the FCS, COP trajectories were normalized to 100% stance and the anterior-posterior (COP_{AP}) and mediolateral (COP_{ML}) positions at each 10% stance interval were calculated for comparison between conditions. Additionally the following discrete COP trajectory parameters were calculated: COP_{AP} and COP_{ML} excursions, the most medial location of the COP, and percent stance at which the most medial COP location occurs.

Statistical Analysis

Three sets of statistical analyses were performed. First the AT subjects were compared with their respective controls (CON_AT). Second the MTSS subjects were compared with their respective controls (CON_MTSS). Finally, since these two injuries are thought to share common biomechanical causes, statistical comparisons were performed using a combined pool of both AT and MTSS subjects (INJ) with a combined pool of all control subjects (CON).

Differences between injured and control groups on the clinical measures were evaluated using independent observations *t*-tests. Group differences in the kinematic measures (Table 4.2) were evaluated using a 2x2 analysis of variance (ANOVA), with injury group and foot strike pattern being the two independent variables. Injury group was a categorical variable with two levels, injured and control. Foot strike pattern was included as a second independent variable since several of the kinematic variables examined in this study vary with foot strike pattern [134,135]. Foot strike was treated as a categorical variable with two levels, rearfoot (RFS) or mid/forefoot strike (M/FFS), classified based on SI values less than 33% or greater than 33%, respectively [18]. Running speed was entered as a covariate since subjects ran at self-selected speeds. Statistical significance was indicated when $\alpha < .05$. Effect sizes (Cohen's d or f, as appropriate) were calculated for all statistically significant differences to aid in the interpretation of results. Effect sizes of 0.1 - 0.25, 0.25 – 0.40, and > 0.40 were used to indicate small, medium, and large effects, respectively [137].

To gain further insight into which clinical and kinematic variables best predicted group membership, all variables where comparisons between injured and control groups resulting in a main effect of group at $p \leq .20$ were considered for entry into a forward stepwise binary logistic regression. Prior to entry into the regression model, colinearity between variables was assessed with a bivariate correlation analysis. When variables demonstrated significant correlation ($p < .05$) with another variable, only one was entered into the regression model. Alpha levels of .05 and .01 were used as criteria for entry and removal from the regression model, respectively.

A 2x2 (foot strike x pronation group) ANOVA was also used to examine differences in COP trajectories between injured and control groups. As with the clinical and kinematic variables, an $\alpha \leq .05$ was used for determining statistically significant differences in the discrete COP related variables (COP_{AP} and COP_{ML} excursions, most medial location of the COP, and percent stance the most medial COP location occurred). However, for comparing COP_{AP} and COP_{ML}, positions at each 10% stance increment a Bonferroni correction was applied. Therefore, for these two variables an $\alpha \leq .0045$ (.05/11) was used to determine statistical significance. Again, effect sizes (Cohen's f) were calculated to help interpret any statistically significant differences.

Except for effect size calculations, all statistical tests were performed using Statistical Packages for the Social Sciences (SPSS, IBM Corp., Armonk NY) version 18. Effect size calculations were performed using the G*Power 3.1 software [138].

Results

Clinical Exam Measures

The comparison between the AT and CON_AT groups resulted in statistically significant differences for standing tibia varus angle (STVA) and static ankle dorsiflexion range of motion (SADF), with the AT group demonstrating higher values of STVA and reduced SADF than the CON_AT group (Table 4.3). Effect sizes for these differences

Table 4.3. Results from the comparison of the clinical exam variables between AT and CON_AT groups. [a] indicates a significant difference between groups at the $p < .05$ level. [b] indicates variables which were considered for inclusion in the logistic regression.

Variable	AT	CON_AT	p	ES
STVA (°)	8.6 (± 1.9)	6.9 (± 2.0)	.025 [a,b]	0.94
SADF (°)	7.6 (± 4.1)	11.9 (± 5.4)	.034 [a,b]	0.88
SAPF (°)	55.9 (± 6.5)	53.5 (± 9.9)	.461	-
$SHIR_{ROM}$ (°)	34.5 (± 6.1)	33.2 (± 1.9)	.668	-
$SHER_{ROM}$ (°)	25.6 (± 8.1)	25.7 (± 8.5)	.962	-
Popliteal (°)	-27.0 (± 14.7)	-20.7 (± 9.9)	.219	-
Quad (°)	122.6 (± 5.9)	120.1 (± 7.6)	.356	-
Ober's Test	12 pos., 1 neg.	12 pos., 1 neg.	-	-
SSTI (°)	19.6 (± 4.4)	16.2 (± 5.1)	.075 [b]	-
SSTE (°)	6.7 (± 2.3)	6.5 (± 3.4)	.841	-
1st MPJ (°)	48.5 (± 9.4)	49.0 (± 18.1)	.924	-

Note: STVA: standing tibia varus angle; SADF: static active ankle dorsiflexion ROM with knee flexed; SAPF: static active ankle plantar flexion ROM with knee flexed; $SHIR_{ROM}$: prone hip internal rotation ROM; $SHER_{ROM}$: prone hip external rotation ROM; Popliteal: popliteal angle measured as distance from full extension; Quad: prone passive knee flexion angle; Ober's test: Iliotibial band tightness; SSTI: static passive subtalar inversion ROM; SSTE: static passive subtalar eversion ROM; 1st MPJ, static active 1st metatarsophalangeal joint extension ROM.

were both large (STVA = 0.94; SADF = 0.88).

Similar results were seen for the comparison of the MTSS and CON_MTSS groups, with the MTSS group demonstrating higher values of STVA and reduced SADF than the CON_MTSS group (Table 4.4). However, in this case while the effect size for STVA was still large (1.29), the effect size for the difference in SADF was much smaller (0.21). Finally, the comparison between INJ and CON groups also resulted in higher values of STVA and reduced SADF in the INJ group compared to the CON groups (Table 4.5). Effect sizes were again large for the STVA comparison (1.05) but small for the SADF comparison (0.13).

Table 4.4. Results from the comparison of the clinical exam variables between MTSS and CON_MTSS groups. [a] indicates a significant difference between groups at the $p < .05$ level. [b] indicates variables which were considered for inclusion in the logistic regression.

Variable	MTSS	CON_MTSS	*p*	ES
STVA (°)	8.6 (± 1.8)	6.6 (± 1.2)	.020 [a,b]	1.29
SADF (°)	3.8 (± 5.4)	10.0 (± 4.9)	.030 [a,b]	0.21
SAPF (°)	53.0 (± 7.3)	50.0 (± 11.0)	.530	-
$SHIR_{ROM}$ (°)	26.6 (± 6.1)	32.0 (± 11.7)	.270	-
$SHER_{ROM}$ (°)	29.4 (± 7.8)	30.5 (± 7.8)	.780	-
Popliteal (°)	-18.5 (± 8.1)	-19.4 (± 6.9)	.820	-
Quad (°)	121.0 (± 1.9)	119.4 (± 5.6)	.450	-
Ober's Test	7 pos., 1 neg.	5 pos., 3 neg.	-	-
SSTI (°)	16.5 (± 4.5)	19.3 (± 3.9)	.208	-
SSTE (°)	6.6 (± 3.0)	6.3 (± 2.9)	.803	-
1^{st} MPJ (°)	48.8 (± 7.4)	55.0 (± 12.8)	.252	-

Note: See note in Table 4.3 for variable abbreviations and definitions.

Table 4.5. Results from the comparison of the clinical exam variables between INJ and CON groups. [a] indicates a significant difference between groups at the $p < .05$ level. [b] indicates variables which were considered for inclusion in the logistic regression.

Variable	INJ	CON	*p*	ES
STVA (°)	8.7 (± 1.8)	6.8 (± 1.8)	.001 [a,b]	1.05
SADF (°)	6.1 (± 5.0)	11.2 (± 5.1)	.003 [a,b]	0.13
SAPF (°)	54.8 (± 6.8)	52.1 (± 10.2)	.325	-
$SHIR_{ROM}$ (°)	31.5 (± 7.1)	32.8 (± 9.4)	.619	-
$SHER_{ROM}$ (°)	27.0 (± 8.0)	27.6 (± 8.4)	.837	-
Popliteal (°)	-23.8 (± 13.1)	-20.2 (± 8.7)	.311	-
Quad (°)	122.0 (± 4.8)	119.8 (± 6.8)	.237	-
Ober's Test	19 pos., 2 neg.	17 pos., 4 neg.	-	-
SSTI (°)	18.4 (± 4.6)	17.3 (± 4.8)	.454	-
SSTE (°)	6.7 (± 2.5)	6.4 (± 3.2)	.746	-
1^{st} MPJ (°)	48.6 (± 8.5)	51.3 (± 16.2)	.500	-

Note: See note in Table 4.3 for variable abbreviations and definitions.

Push Off Characteristics

Results were consistent regardless of which groups were compared. There were no differences in PSHO between the AT and CON_AT groups ($p = .323$), MTSS and CON_MTSS groups ($p = 0.33$), or the INJ and CON groups ($p = .624$; Figure 4.1A). Similarly, there were no differences in PI between AT and CON_AT groups ($p = .622$), between MTSS and CON_MTSS groups ($p = .443$), or between INJ and CON groups ($p = .679$; Figure 4.1B). There were also no differences in PPF between AT and CON_AT groups ($p = .551$), between MTSS and CON_MTSS groups ($p = .219$), or between INJ and CON groups ($p = .489$; Figure 4.1C). Finally, there were no differences in PVF

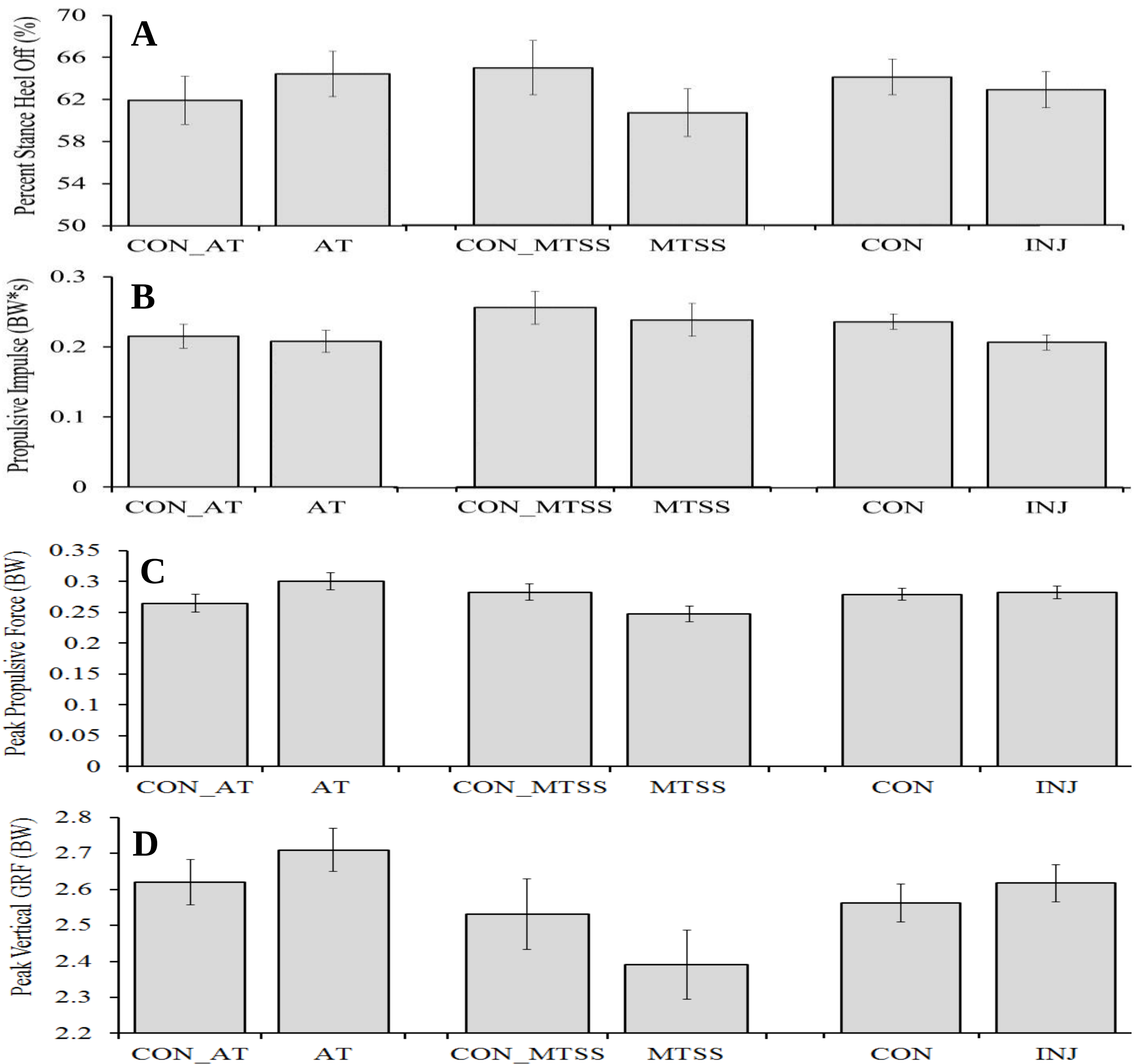

Figure 4.1. Comparisons of the percent stance heel off occurs (PSHO; A), propulsive impulses (PI; B), peak propulsive forces (PPF: C), and peak vertical forces (PVF; D) between the AT and CON_AT groups, between MTSS and CON_MTSS groups, and between INJ and CON groups. No statistically significant differences were observed between groups for any comparison.

between AT and CON_AT groups (p = .311), between MTSS and CON_MTSS groups (p = .482) or between INJ and CON groups (p = .489; Figure 4.1D).

Kinematic Variables

Results from the kinematic variables are shown in Tables 4.6, 4.7, and 4.8. As with the push off characteristics, results were consistent whether comparing AT to CON_AT, MTSS to CON_MTSS, or INJ to CON. Injured subjects demonstrated greater

Table 4.6. Results from the comparison of the kinematic variables between AT and CON_AT groups. [a] indicates a significant difference between groups at the $p < .05$ level. [b] indicates variables which were considered for inclusion in the logistic regression.

Variable	AT	CON_AT	*p*	ES
Per_P (% stance)	85.2 (± 16.3)	61.6 (± 14.2)	.002 [a,b]	0.82
EHO (°)	-6.0 (± 5.8)	-0.5 (± 2.2)	.002 [a,b]	0.82
HIR_{excur} (°)	11.3 (± 6.5)	13.6 (± 4.8)	.485	-
EE (°)	10.6 (± 3.9)	11.9 (± 2.5)	.398	-
TTPE (ms)	63.0 (± 16.3)	52.4 (± 15.0)	.131 [b]	-
AEV (°/s)	192.5 (± 49.7)	247.4 (± 77.2)	.061 [b]	-
MEV (°/s)	295.6 (± 83.1)	366.9 (± 99.6)	.089 [b]	-

Table 4.7. Results from the comparison of the kinematic variables between MTSS and CON_MTSS groups. [a] indicates a significant difference between groups at the $p < .05$ level. [b] indicates variables which were considered for inclusion in the logistic regression.

Variable	MTSS	CON_MTSS	*p*	ES
Per_P (% stance)	83.2 (± 15.1)	55.7 (± 21.2)	.002 [a,b]	0.823
EHO (°)	-5.1 (± 6.5)	1.9 (± 2.2)	.001 [a,b]	0.934
HIR_{excur} (°)	7.5 (± 5.8)	7.6 (± 3.7)	.748	-
EE (°)	10.1 (± 3.7)	11.8 (± 4.5)	.834	-
TTPE (ms)	55.6 (± 11.9)	53.5 (± 11.2)	.770	-
AEV (°/s)	192.0 (± 80.7)	226.4 (± 84.7)	.451	-
MEV (°/s)	299.4 (± 109.9)	360.8 (± 127.1)	.669	-

Table 4.8. Results from the comparison of the kinematic variables between INJ and CON groups. [a] indicates a significant difference between groups at the $p < .05$ level. [b] indicates variables which were considered for inclusion in the logistic regression.

Variable	INJ	CON	*p*	ES
Per_P (% stance)	85.2 (± 15.6)	58.9 (± 16.5)	< .001 [a,b]	0.854
EHO (°)	-6.1 (± 5.9)	1.0 (± 2.3)	< .001 [a,b]	0.912
HIR_{excur} (°)	9.81 (± 6.4)	11.0 (± 5.2)	.514	-
EE (°)	10.4 (± 3.8)	11.9 (± 3.4)	.107 [b]	-
TTPE (ms)	59.4 (± 14.8)	53.3 (± 12.9)	.182 [b]	-
AEV (°/s)	196.2 (± 61.4)	234.9 (± 78.5)	.069 [b]	-
MEV (°/s)	299.9 (± 104.5)	361.5 (± 109.8)	.054 [b]	-

values of STVA, longer Per_P values, and greater EHO than their respective controls. Effect sized for these comparisons all indicated large effects, ranging from 0.82 to 1.074. However, no differences in HIR_{excur} were observed between any groups. Similarly, no differences in the traditional variables of EE, TTPE, AEV, or MEV were observed between any groups.

Logistic Regression

For the comparison between AT and CON_AT groups the following variables resulted in main effects of injury group with $p < .2$ and were therefore considered for inclusion in the regression model: STVA, SADF, SSTI, Per_P, EHO, TTPE, AEV, and M EV. A bivariate correlation analysis revealed significant correlations between Per_P and STVA ($r = .43$, $p = .032$), PerP and SADF (r = --.603, $p = .001$), Per_P and EHO ($r = -.734$, $p < .001$), TTPE and AEV ($r = -.489$, $p = .013$), and between AEV and MEV ($r = .880$, $p < .001$). Therefore, only SSTI, Per_P, TTPE, and MEV were retained for entry into the regression model.

After two steps, a logistic regression model containing the variables of Per_P, and TTPE, entered in that order, was able to correctly classify 92% of the limbs into the AT or CON_AT groups. The overall model was significant (χ^2 = 18.992, df = 2, p < .001) and the Nagelkerke R^2 value was .71, indicating that 71% of the variance between AT and CON_AT groups was explained by Per_P and TTPE. The inclusion of SSTI or MEV did not improve the model. Odds ratios for Per_P and TTPE, as well as the complete regression results, are shown in Table 4.9.

For the comparisons between MTSS and CON_MTSS, four variables resulted in main effects of injury group with p < .2: STVA, SADF, Per_P, and EHO. Bivariate correlation analysis revealed significant correlations between Per_P and STVA (r = .531, p = .034) and between Per_P and EHO (r = -.675, p = .004). Therefore, only SADF and Per_P were entered into the logistic regression.

A logistic regression model containing Per_P was able to correctly classify 87.5% of the limbs into the AT or CON_AT groups. The overall model was significant (χ^2 = 7.971, df = 1, p < .005) and the Nagelkerke R^2 value was .523, suggesting that 52.3% of the variance between MTSS and CON_MTSS groups was explained by Per_P. The addition of SADF did not improve the model. Odds ratios for Per_P, as well as the

Table 4.9. Results of the logistic regression model for predicting membership in the AT group.

Variable	β	S.E.	Sig.	Odds Ratio	95% CI for Odds Ratio	
					Lower	Upper
Per_P	0.136	0.050	.006	1.145	1.039	1.262
TTPE	0.094	0.044	.033	1.098	1.008	1.197
Constant	-15.460	5.622	-	-	-	-

Table 4.10. Results of the logistic regression model for predicting membership in the MTSS group.

Variable	β	S.E.	Sig.	Odds Ratio	95% CI for Odds Ratio	
					Lower	Upper
Per_P	0.092	0.045	.039	1.097	1.005	1.197
Constant	-6.785	3.454	-	-	-	-

complete regression results, are shown in Table 4.10.

Finally, the comparison between the INJ and CON groups resulted in eight variables with a main effect of group at $p \leq .2$ or below: STVA, SADF, Per_P, EHO, EE, TTPE, AEV, and MEV. Bivariate correlation analysis revealed STVA was significantly correlated with Per_P (r = .482, p = .001) and EHO (r = -.493, p = .001), SADF was also significantly correlated with Per_P (r = -.510, p = .001), and EE was significantly correlated with AEV (r = .739, p < .001) and with MEV (r = .856, p <.001). Therefore, only Per_P, TTPE, and MEV were retained for inclusion in the logistic regression.

After two steps, a logistic regression model containing the variables of Per_P, and TTPE, entered in that order, was able to correctly classify 85.7% of the limbs into the INJ or CON groups. The overall model was significant (χ^2 = 26.302, df = 2, p < .001) and the Nagelkerke R^2 value was .621, suggesting that 62.1% of the variance between INJ and CON groups was explained by Per_P and TTPE. The inclusion of MEV did not improve the model. Odds ratios for Per_P and TTPE, as well as the complete regression results, are shown in Table 4.11.

Table 4.11. Results of the logistic regression model for predicting membership in the combines INJ group.

Variable	β	S.E.	Sig.	Odds Ratio	95% CI for Odds Ratio	
					Lower	Upper
Per_P	1.09	0.031	< .001	1.115	1.050	1.184
TTPE	0.62	0.030	.039	1.064	1.003	1.129
Constant	-11.558	3.392	-	-	-	-

COP Trajectories

Similar patterns were observed for comparisons of the discrete COP trajectory parameters between AT and CON_AT groups, MTSS and CON_MTSS groups, and INJ and CON groups. For all three comparisons, there was no main effect of pronation group for COP_{AP} excursion. However, there were significant main effects of foot strike pattern (Figures 4.2A, 4.3A, and 4.4A), with individuals who utilized a RFS demonstrating

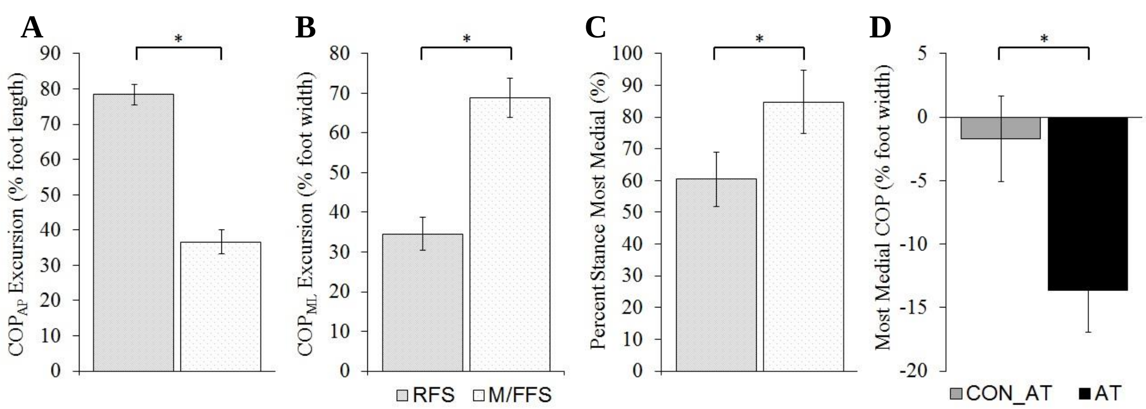

Figure 4.2. Comparisons between the AT and CON_AT groups for COP_{AP} excursions (A), COP_{ML} excursions (B), percent stance at which the most medial location of the COP occurred (C), and the most medial location of the COP (D). * indicates significance at the $p < .05$ level. A, B, and C show a main effect of foot strike pattern while D shows a main effect of injury group.

greater COP_{AP} excursion than individuals who used a M/FSS. Similarly, for COP_{ML} excursion, there was no main effect of injury group for any of the three comparisons but there was a main effect of foot strike pattern, with individuals who utilized a RFS demonstrating less COP_{ML} excursion than those who utilized a M/FFS (Figures 4.2B, 4.3B, and 4.4B).

No significant main effects of injury group were observed for any of the three comparisons for the percent stance at which the most medial location of the COP was located. There was, however, a significant main effect of foot strike pattern, but only for the comparison between AT and CON_AT where individuals who utilized a RFS reached their most medial COP location sooner in stance than individuals who utilized a M/FFS (Figure 4.2C). Similar main effects of foot strike pattern were not observed for comparisons between MTSS and CON_MTSS or INJ and CON groups (Figures 4.3C and 4.4C. Finally, for the most medial location of the COP, there was a significant main

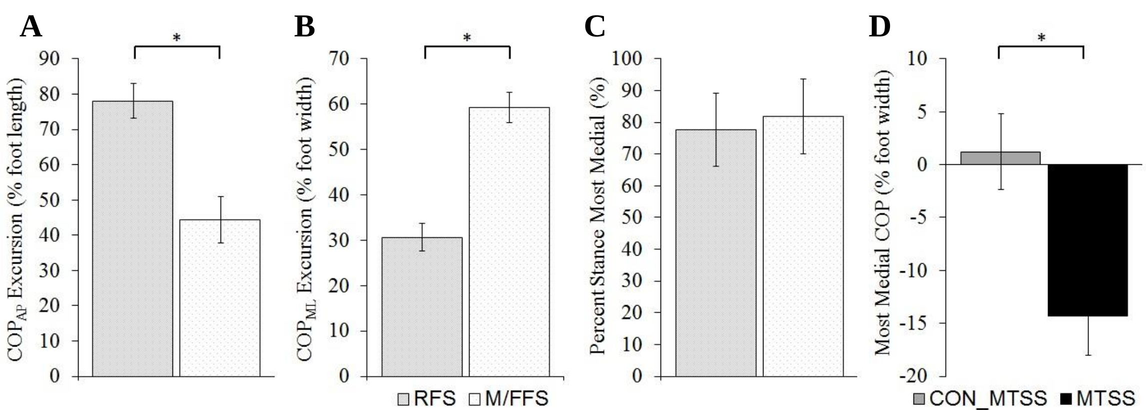

Figure 4.3. Comparisons between the MTSS and CON_MTSS groups for COP_{AP} excursions (A), COP_{ML} excursions (B), percent stance at which the most medial location of the COP occurred (C), and the most medial location of the COP (D). * indicates significance at the $p < .05$ level. A and B show a main effect of foot strike pattern while D shows a main effect of injury group.

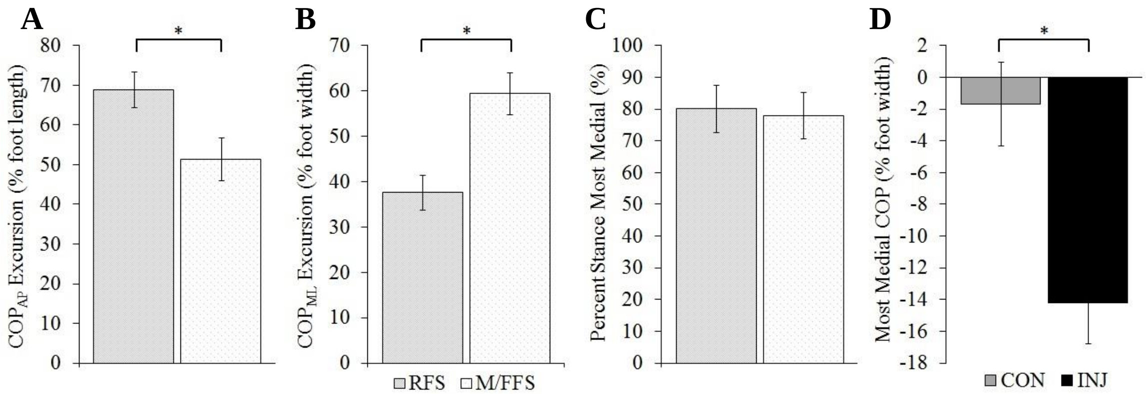

Figure 4.4. Comparisons between the combined INJ and CON groups for COP_{AP} excursions (A), COP_{ML} excursions (B), percent stance at which the most medial location of the COP occurred (C), and the most medial location of the COP (D). * indicates significance at the $p < .05$ level. A and B show a main effect of foot strike pattern while D shows a main effect of injury group.

effect of injury group for all three comparisons, with the AT, MTSS, and INJ groups having a more medial peak COP_{ML} location than the CON_AT, CON_MTSS, or CON groups, respectively (Figures 4.2D, 4.3D, and 4.4D).

For the overall COP trajectories, in the AT compared to the CON_AT groups, the COP was located significantly more medially from 20% of stance through 90% of stance (Figure 4.5). There were no significant main effects of injury group for COP_{AP} positioning at any time point during stance. However, there was a significant main effect of foot strike, with individuals who utilized a M/FFS having a more anteriorly located COP_{AP} at initial contact, 10%, and 20% stance compared to individuals who utilized a RFS. Comparisons between MTSS and CON_MTSS groups and the INJ and CON groups yielded similar results as the AT and CON_AT groups, with the COP being located significantly more medially from 20% through 90% of stance (Figures 4.6 and 4.7). Additionally, for both the MTSS compared to CON_MTSS and INJ compared to

CON comparisons, there were no significant main effects of injury group for the COP_{AP} positioning at any percent stance. However, both comparisons resulted in significant main effects of foot strike, with individuals who utilized a M/FFS having a more anteriorly located COP_{AP} at initial contact, 10%, and 20% stance compared to individuals who utilized a RFS.

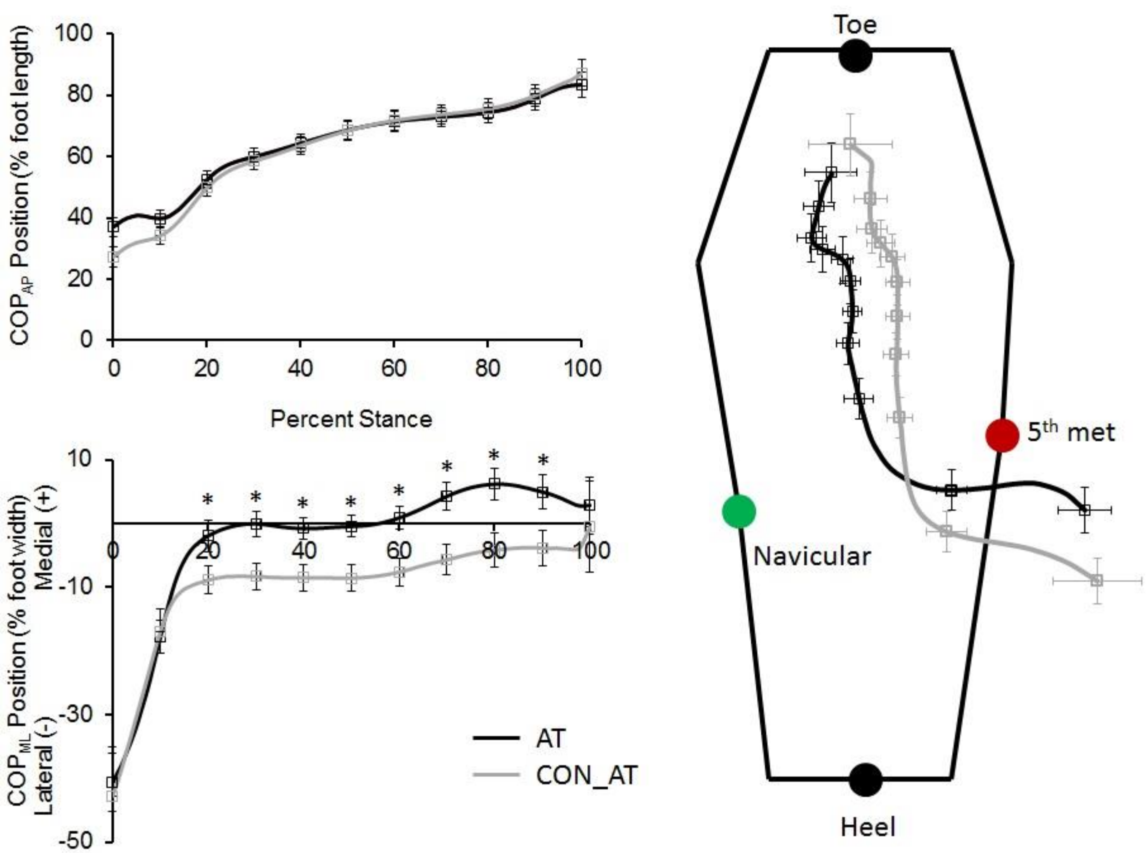

Figure 4.5. Results of the comparisons of the COP_{AP} (A) and COP_{ML} (B) positions between AT and CON_AT groups. Also shown are the COP trajectories plotted in an outline of the foot drawn based on marker locations (C). * indicated significant injury group difference at the $p < .0045$ level.

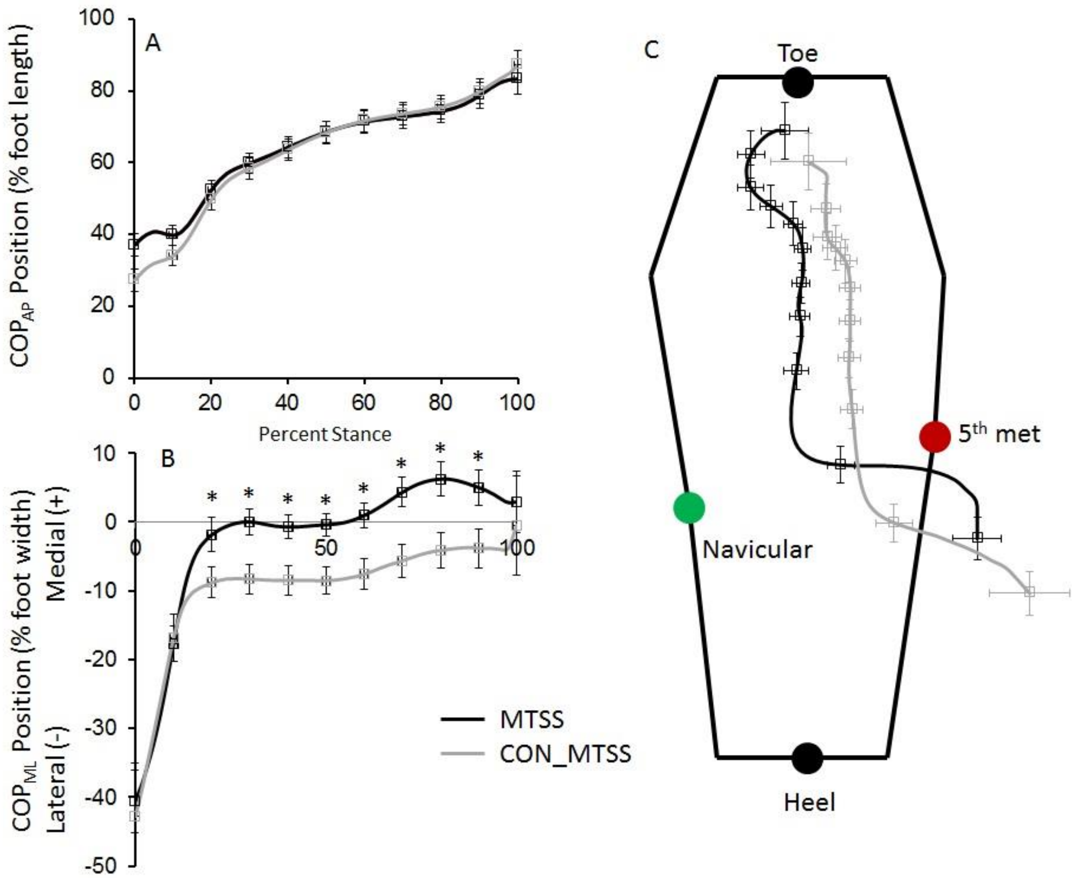

Figure 4.6. Results of the comparisons of the COP_{AP} (A) and COP_{ML} (B) positions between MTSS and CON_MTSS groups. Also shown are the COP trajectories plotted in an outline of the foot drawn based on marker locations (C). * indicated significant injury group difference at the $p < .0045$ level.

Discussion

The purpose of this study was to compare periods of pronation in runners currently symptomatic with two injuries often attributed to excessive amounts or velocities of pronation, Achilles tendinopathy (AT) and medial tibial stress syndrome (MTSS). Currently symptomatic individuals were compared to matched controls separately based on injury (AT or MTSS), and as a joint injury group. The main findings were similar regardless of which comparison group was examined. In support of our

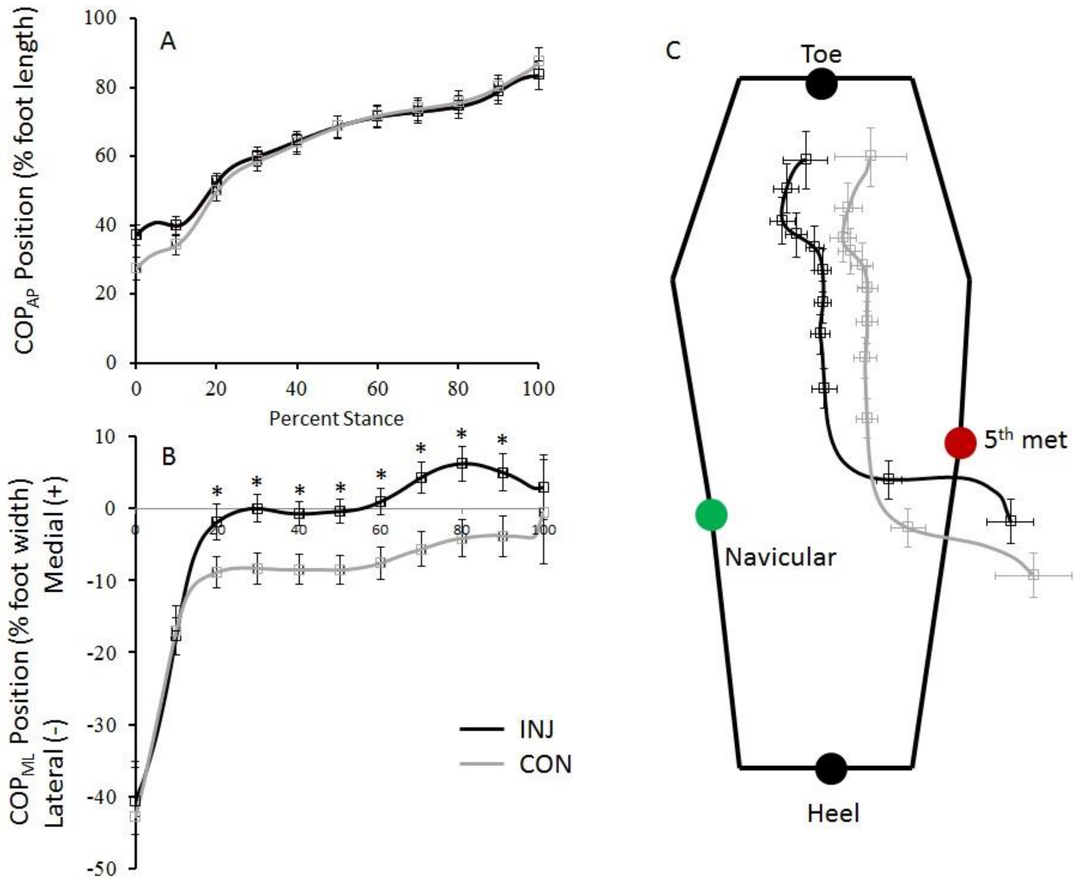

Figure 4.7. Results of the comparisons of the COP$_{AP}$ (A) and COP$_{ML}$ (B) positions between INJ and CON groups. Also shown are the COP trajectories plotted in an outline of the foot drawn based on marker locations (C). * indicated significant injury group difference at the $p < .0045$ level.

hypothesis, injured individuals did not demonstrate greater amounts or velocities of rearfoot eversion than healthy controls. However, compared to healthy controls, injured individuals did demonstrate longer periods of pronation, reduced static dorsiflexion range of motion, a more everted heel orientation at heel off, and higher levels of standing tibia varus. Additionally, the period of pronation was a significant predictor for membership in the AT or MTSS groups. Finally, injured individuals demonstrated more medial locations of their COP from 20% through 90% of stance. Since these variables are all markers of prolonged pronation, when viewed as a whole these findings support the hypothesis that the amount or velocity of pronation is not necessarily as important as for these two injuries as the duration of pronation.

The literature is not in agreement on whether anatomic alignment is associated with the development of these two injuries. While some authors have reported a more pronated foot during quiet standing in individuals with MTSS compared to healthy controls [62–65], other authors have reported no differences in foot structure between injured and non-injured subjects [33,87]. Similarly, some authors have reported a more pronated type in individuals with AT [46] while others have found foot type does not appear related to the development of AT [14,91]. The results from the current study would support the idea that foot structure may not play an important role in the development of AT or MTSS since there were no differences in arch height index, subtalar joint range of motion in inversion or eversion, or 1st metatarsophalangeal joint range of motion between injured and control subjects.

There were two clinical exam measures however, which were consistently different between injured and healthy subjects, the standing tibia varus angle and the static dorsiflexion range of motion. Previous authors have suggested that higher tibia varus angles may be lead to compensatory pronation simply to get the foot flat [13]. In support of this we have previously identified tibia varus angles as one biomechanical marker of prolonged pronation. This specific variable has been examined in previous work comparing individuals with MTSS to healthy controls [62–64], and while these results tend to support this association, to date they have been only trends, and lacked statistical significance.

A lack of static ankle dorsiflexion has previously been reported to be predictive of developing both MTSS [59] and AT [91]. Lower static ankle dorsiflexion was also observed in the injured groups compared to the controls in the current study. However, given significant large correlations with the period of pronation, this variable was not entered into any of the logistic regression models. Previous authors have suggested a lack of dorsiflexion may be indicative of a functional equinus and, as a result these runners may demonstrate compensatory pronation simply to get their forefoot flat on the ground [13]. Since compensatory pronation would include additional dorsiflexion and forefoot abduction beyond what is observed in "normal" pronation, it may well increase the forces being applied to the Achilles tendon. This may be one possible reason why these studies have reported reduced dorsiflexion as a predictor of AT development.

Currently the literature lacks prospective studies on AT or MTSS. However, the few that do exist report more medial pressure under the metatarsals during forefoot push off in individuals who developed AT [90] and more pressure under the medial aspect of the foot during midstance in those who developed exercise related lower leg pain [94,95]. Interestingly, while not mentioned by the authors of the studies, a recent review on lower limb biomechanics during running in individuals with AT by Munteanu et al. [142] suggested these more medial pressures could be interpreted as an "unlocking" of the midtarsal joints thereby increased forefoot mobility and reducing the foot's ability to act as a rigid lever during propulsion. This then leads to higher tensile forces being transmitted through the Achilles tendon during push off. The more medial COP locations from 20% through 90% of stance and more medial peak COP_{ML} positioning found in the current study both agree with previously reported COP trajectory differences in individuals with AT or MTSS and, especially when considered in combination with the kinematic findings of the current study, support the interpretation of Munteanu et al. [142]. These results also reinforce the idea that COP trajectories and plantar pressure distributions are important variables for future studies on these two injuries to consider.

One might reasonably suspect, given the nature of the injuries, that individuals with AT or MTSS would apply lower forces with the involved limb during push off. However, the present study found there were no differences in peak propulsive forces, propulsive impulses, or peak vertical ground reaction force between injured and healthy individuals. While data on ground reaction forces in individuals with MTSS is absent from the literature, these results are consistent with previous investigations that reported no differences in peak propulsive forces or peak vertical ground reaction forces between individuals with AT and healthy individuals [46,143]. The lack of differences in push off parameters suggests the mechanics of push off are similar between injured and healthy individuals. This is further supported by the finding of no differences in the time to heel off between injured and healthy controls. Thus, overall, it appears injured that individuals push off at similar times during stance phase and with similar amounts of force as non-injured individuals.

The injured individuals did, however, greater periods of pronation throughout stance and a more everted heel position at heel off. This suggests that while the timing and amount of force being applied are similar, the actual configuration of the foot during this time differs between injured and healthy individuals. This can be made especially clear with an examination of the inversion-eversion curves for the INJ and CON subjects (Figure 4.8). Both groups pass from an inverted rear foot to an everted rear foot at the

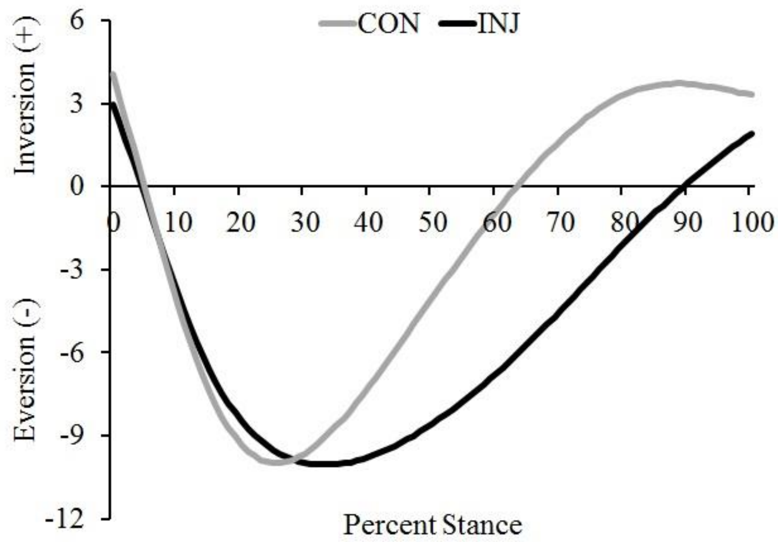

Figure 4.8. Mean inversion-eversion curves for the INJ and CON groups.

same time early during stance. Figure 4.1 shows that there were no differences in when push off began between INJ and CON groups, with both groups showing heel off occurring at approximately 63% stance. From Table 4.5 and Figure 4.8 we see that the CON group has a slightly inverted foot at this time while the INJ group is still everted approximately 6°. In fact, the INJ group does not invert to a neutral position until almost 90% of stance, well after the main work of push off is completed.

As the foot supinates the axes of the transverse tarsal and tarsometatarsal joints converge "locking" the midfoot and turning the foot into a rigid lever [141]. As can be observed in Figure 4.8, this is what happens with the CON group. However, since the INJ group is still pronated for much of the push off phase (Figure 4.8), then the axes of these joints are still somewhat parallel, and as a result, the foot is still a soft, flexible structure. Yet, individuals in the INJ group are still able to generate propulsive forces similar to those in the CON group (Figure 4.1). Since their bony anatomy is not configured to achieve such a feat, this suggests they must have significant help in stabilizing the foot from both the intrinsic and extrinsic foot musculature.

Whether this extra muscular effort is directly related to the development of these two common injuries cannot be concluded from the current study. However, since the extra muscular effort would be required on each step over the course of a run, one may reasonably hypothesize that it could be involved. This could occur through several mechanisms including inefficiencies within the foot's lever system and changes in the operating region of the muscles on their respective force length operating curves.

During the push off phase of the gait cycle the COP is generally located under the metatarsal heads and the forward progression is achieved with the foot pivoting around a point under the metatarsals, a condition commonly referred to as the forefoot rocker [144]. At this point in time the foot can be models as a simple 2nd class lever system with the pivot point under the metatarsals, the resistive force being the body weight and the effort force being the force through the Achilles tendon (Figure 4.9). In an ideal (i.e. rigid) 2nd class lever system, the lever arm for the effort force is longer the lever arm for the resistive force. Since the torques produced by each lever will be equivalent, a significant mechanical advantage will be achieved and less force will be required at the effort end of the lever.

However, the main limitation to the approach detailed here, and especially relevant for prolonged pronation and AT, is that it relies on the assumption that the lever is a rigid body. While applying this assumption to the human foot is dubious in the best of scenarios [145,146], it may be especially problematic when considering prolonged pronation since the very movement patterns associated with

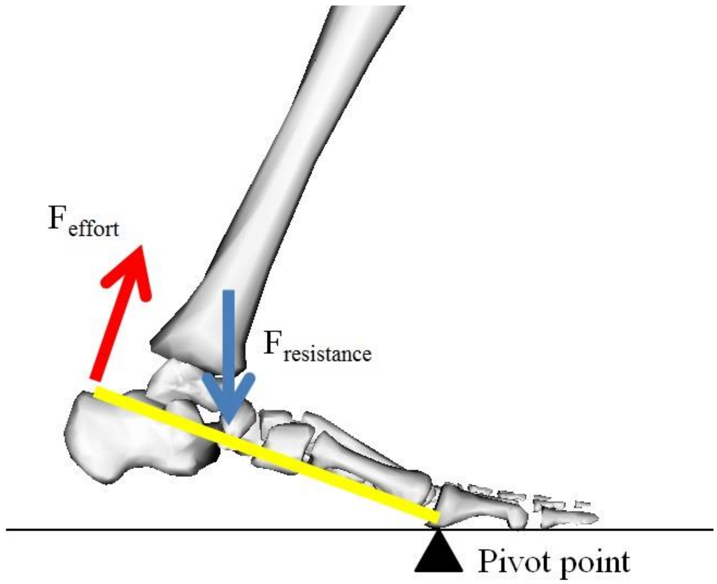

Figure 4.9. Illustration of the foot as a simple second class lever during the push off phase.

prolonged pronation suggest a soft, flexible bony alignment of the foot. Thus, the lever is no longer rigid and, instead of generating torque, some of the force from the Achilles tendon will be used to deform the lever.

Maintaining the torque equilibrium in the presence of such deformation will require greater amounts of the effort force. In anatomic terms this would be equivalent to more force passing through the tendons of the plantar flexors. While this is primarily the Achilles, there are also small contributions from the flexors hallucis and digitorum longus as well as the tibialis posterior. Depending on the tendon stiffness, greater forces applied to these tendons may induce higher strains within the tendons. In vitro studies have suggested the level of strain present is a major factor in determining time to tendon failure when tendons are subjected to repeated loading [82] while epidemiology discussions have implicated high strain levels as a major factor in the development of AT

[83,84]. Thus, the flexible foot during push off associated with prolonged pronation may be linked to the development of the actual injury. However, prospective studies are needed to further investigate this hypothesis.

A second possible mechanism through which prolonged pronation my lead to the development of MTSS or AT involves the lengths of the muscles and tendons. The anatomic structures most often implicated for the development of MTSS include the flexor digitorum longus [68,72], tibialis posterior [69,73], and soleus [68,72,74] muscles while for AT it is the actual Achilles tendon itself, and thus by extension, the soleus, and medial and lateral gastrocnemius muscles. Since pronation involves eversion of the calcaneus and a lowering of the medial longitudinal arch, then, given the bony attachment sites of these muscles, pronation should increase the length of their respective musculotendinous units. Since the musculotendinous unit consists of both the active and passive components, it could respond in several ways.

First, if the muscle belly itself does not actually lengthen, then the increased length will have to come from the tendon component. As already mentioned, higher strains within tendons is thought to be a major factor in their time to failure under repeated loading conditions [82–84]. Another possibility is that the actual muscle belly itself lengthens. EMG data suggests these muscles are very active during the push off period from 50% to 80% of stance [51–53]. If the muscle belly is simply longer but not actually lengthening during the contraction then the muscles are likely working concentrically. However, with a longer length the muscle may be operating in a different location on the force length curve. For instance, the soleus generally operates on the ascending limb of the force length curve up until about 30° of dorsiflexion [147].

Dorsiflexion during the push off phase is well under this value, thus a longer soleus would generally lead to higher force production. Similar effects may be observed in the flexor digitorum longus and tibialis posterior.

If the muscle bellies are actually lengthening while they are actively generating force then the muscles are likely working eccentrically. It is well established that eccentric contractions lead to higher forces. Thus, whether from differences in force-length curve operating conditions, or from having to operate eccentrically, prolonged pronation may result in higher force generation from the muscles. When repeated on each step, higher muscular forces could increase the force going through the Achilles tendon and increase the traction forces being applied at the bony origins of the muscles into the tibia, the two mechanisms currently hypothesized to be responsible for causing these injuries.

While the above discussion suggests some theoretical ways prolonged pronation may play a role in the development of AT or MTSS, whether or not this is actually happening cannot be concluded from the current study. Future work, both in prospective studies following individuals with prolonged pronation to see if they develop AT or MTSS, and in terms of connecting the movement patterns associated with prolonged pronation to actual injury mechanisms are clearly required.

In conclusion, this study examined whether individuals currently symptomatic with either AT or MTSS, two common running injuries typically attributed to excessive amounts or velocities of pronation, instead exhibit prolonged pronation. Compared to healthy controls, injured individuals in the current study demonstrated several biomechanical markers of prolonged pronation including longer periods of pronation, higher standing tibia varus angles, and more medial locations of the COP. The lack of differences in either the amount or velocity of pronation between injured and control subjects, and the finding that the strongest predictor of AT or MTSS group membership was the period of pronation, suggests the problematic mechanics associated with these two injuries occur later in stance phase, during push off, not during the initial loading phase early in stance. These results call into question the traditional theories of rapid pronation creating a “whip like” effect within the Achilles [46,140], or excessive amounts and velocities of pronation leading to the development of MTSS [47,62,64,65,95]. Though additional work is required to confirm the findings of the current study, if supported, these results have significant implications for future studies on prevention and rehabilitation of these two common running injuries.

Bridge

Chapter II examined a method for describing COP trajectories in relation to the anatomic structures of the foot using a force platform. Chapter III identified higher standing tibia varus angles, longer periods of pronation, a more everted heel position at heel off, and a more medial location of the COP as biomechanical markers of prolonged pronation. The current chapter examined whether individuals with AT and MTSS demonstrated prolonged pronation as opposed to the commonly cited excessive amounts or velocities of pronation. In this regard, compared to health controls, injured individuals did in fact demonstrate the biomechanical markers of prolonged pronation while not demonstrating any differences in the amounts or velocities of pronation.

Several hypotheses linking prolonged pronation to the development of AT and MTSS were presented in Chapter IV, including the idea that prolonged pronation results in higher strains within the musculotendinous structures. While this theory makes sense based on the combination of anatomy and the kinematics observed in the injured individuals, at this point is not known whether this is actually happening. Thus, Chapter V will apply musculoskeletal modeling tools to examine what is happening at the musculotendinous level in individuals with AT, MTSS, and matched healthy controls. Currently healthy individuals who demonstrate prolonged pronation will also examined as this will provide insights into whether prolonged pronation may predispose individuals for one of these two common overuse injuries.

CHAPTER V

PLANTAR FLEXOR MUSCULOTENDON KINEMATICS IN INJURED AND HEALTHY RUNNERS WITH AND WITHOUT PROLONGED PRONATION

This chapter contains co-authored material and was developed by Dr. Li-Shan Chou and James Becker. Dr. Chou contributed to the development and refinement of the methodology as well as providing critiques and editing advice for the manuscript while Mr. Becker was responsible for conceptual development, development of the protocol, data collection and analysis, as well as all the writing. Additionally, Dr. Stan James and Mr. Robert Wayner both contributed by referring patients treated at their respective clinical practices.

Introduction

The injury rate among runners is alarmingly high, with anywhere from 25% to 75% of all runners experiencing an overuse injury in any given year [2,3,5]. Over the last 40 years, medial tibial stress syndrome (MTSS) and Achilles tendinopathy (AT) have consistently ranked among the five most common overuse injuries sustained by runners [13–15]. Despite the prevalence of these two injuries, there is little agreement in the literature regarding what movement patterns may lead to these injuries.

While excessive amounts or velocities of pronation are often suggested as movement patterns leading to the development of these injuries, there is conflicting evidence in the literature supporting this hypothesis. Some authors have reported injured individuals demonstrate higher amounts or velocities of pronation than non-injured individuals [46,106,140] while others have reported no differences in the amount or velocity of pronation between injured and non-injured individuals [86–89]. An alternative hypothesis suggests it is not the amount or velocity of pronation that is necessarily important, rather it is the duration the foot remains in a pronated position during stance [13]. This is supported by recent work in our laboratory (see Chapter IV) which suggests individuals currently symptomatic with either Achilles tendinopathy or medial tibial stress syndrome do not demonstrate greater amounts or velocities of pronation compared to healthy control subjects, but that they do demonstrate more prolonged pronation.

While there are many possible explanations for the discrepancy in the literature, two in particular stand out. First, the literature does not present a clear picture of when during the stance phase the mechanisms thought to cause these injuries reach their most damaging potential. For instance, while both the excessive amounts or velocities and the prolonged pronation hypotheses suggest foot pronation as the main movement pattern involved in these injuries, they implicate different phases of stance as the primary times of concern. The excessive amounts or velocities hypothesis emphasizes the early portion of stance where the eccentric actions of soleus, tibialis posterior, and flexor digitorum longus may lead to increased load in the tendons and muscles [47,142]. If true, then the highest musculotendinous loads should be observed early in stance phase. However, the prolonged pronation hypothesis suggests the problematic period occurs later in stance, during the push off phase, as the intrinsic and extrinsic foot muscles try to stabilize a non-rigid foot and generate a propulsive impulse [13]. If this hypothesis is true, the highest musculotendinous loads should be observed later in stance. While there is some modeling based evidence reporting the highest forces in the Achilles tendon occurs around 60% of stance [20,21,124], to date these studies have all been in healthy, uninjured runners.

A second possible explanation for the discrepancy in the literature regarding biomechanical factors involved in Achilles tendinopathy and medial tibial stress syndrome is that connections between movement patterns and injury are based on inferences drawn from anatomical descriptions of muscle action. For instance, since the soleus comprises the medial portion of the Achilles tendon [54,55] it is usually assumed that increased calcaneal eversion will lead to higher strains within the tendon. Similarly, the flexor digitorum longus [68,72], tibialis posterior [69,73], and soleus [68,72,74] muscles are the anatomic structures most often implicated in the development of MTSS. Since pronation involves eversion of the calcaneus and a lowering of the medial longitudinal arch, it is assumed that based on the bony attachment sites of these muscles, increased pronation will lead to higher strains within their respective musculotendinous units. While recent *in vitro* cadaver work supports these assumptions [68,85], to date there have not been *in vivo* investigations reporting these relationships in healthy or injured runners.

The assumptions based on anatomic descriptions of muscle action are also problematic when one considers the redundancy naturally built into the human musculoskeletal system. For instance, there are at least six extrinsic foot muscles which can resist calcaneal eversion [50]. Given this redundancy one cannot be sure whether two individuals who demonstrate similar amounts of calcaneal eversion have similar dynamics within all these muscles without quantitative assessment of each individual muscle's musculotendon kinematics. While there are several *in vivo* methods for doing so such as ultrasound [148] or intra-tendon transducers [149], they are limited in that they can either only examine a single muscle or tendon of interest, or involve highly invasive procedures. An alternative method for quantitative analysis of musculotendon dynamics involves the use of musculoskeletal modeling. This approach has been widely used for investigating hamstring muscle dynamics in relation to hamstring strains [150–153], investigating iliotibial band strains in runners with IT band syndrome [154], as well as for examining general muscle function during running [155–158] and walking [156,159,160].

Therefore, the purpose of this study was to use three dimensional motion analysis in conjunction with appropriately scaled musculoskeletal models to characterize and compare plantar flexor musculotendon kinematics during running in individuals currently symptomatic with Achilles tendinopathy or medial tibial stress syndrome, and healthy individuals demonstrating prolonged pronation. It was hypothesized that musculotendon percent elongation would be higher in injured groups than in their respective control groups. Additionally, to provide further characterization of plantar flexor musculotendon kinematics, musculotendon lengths, lengthening and shortening velocities, and percent elongation rates were examined and compared between injured subjects, healthy subjects, and prolonged pronators.

<u>Methods</u>

Subjects

A total of 21 injured individuals, 13 currently symptomatic with AT and 8 currently symptomatic with MTSS, were recruited for this study. Injured subjects were specifically diagnosed by and referred from the clinical practices of SJ and RW, the two clinicians participating in this study. In addition to diagnosing and referring patients, SJ and RW also ruled out any other injuries. For each injured subject, a healthy control subject (CON_AT or CON_MTSS) was also recruited. Controls were matched with injured individuals based on sex, weekly mileage, age, and foot strike pattern (see Table 4.1 in Chapter IV). All control subjects ran at least 20 miles per week and had not sustained a running related injury within the previous six months.

In addition to the AT, CON_AT, MTSS, and CON_MTSS subjects, 13 individuals with clinically determined prolonged pronation (PP) also participated in this study. These individuals were identified by one of the two clinicians (SJ or RW) participating in this study. See Chapter III for a detailed description of how these individuals were classified. Therefore, a total of 55 subjects participated in this study. All subjects read and signed an informed consent approved by the University of Oregon Human Subjects Review board.

Experimental Protocol and Instrumentation

Thirty nine retro-reflective markers were attached to specific body landmarks. A static trial was collected from which anatomic coordinate systems for the pelvis, thigh, shank, and rearfoot segments were established according to ISB recommendations [97]. Subjects then participated in a running gait analysis where their whole body motion was recorded using a 10-camera motion capture system (Motion Analysis Corp., Santa Rosa CA) while they ran continuous laps around a short track in the laboratory (see Figure 3.1). Ground reaction forces were measured with three force plates (AMTI, Watertown MA) located in series in the capture volume. Motion and ground reaction force data were sampled at 200 Hz and 1000 Hz, respectively. Subjects ran continuous laps until a minimum of 8 clean trials were recorded. A trial was deemed clean if the foot landed in the middle of a force plate with no visible signs the subject altered their stride pattern to target the force platform. For the AT and MTSS patients, their involved limb was used while the matching limb was used for control subjects. For the PP subjects, the limb which had been identified by the clinicians as demonstrating prolonged pronation was used.

Data Analysis

Three dimensional marker trajectories and ground reaction forces were filtered with low pass, fourth order, zero lag Butterworth filters using cutoff frequencies of 8 Hz and 50 Hz, respectively. A fifty Newton threshold in the filtered vertical ground reaction force was used to establish the instants of foot contact and toe off [18]. Foot strike pattern was determined using the strike index [18]. Filtered marker trajectories and the anatomic coordinate systems established during the static trial were used to calculate joint angles across stance using a Cardan angle rotation sequence as according to the ISB recommendations [97]. All calculations were performed using custom LabView (National Instruments, Austin TX) software. For each subject, the joint angles from three representative trials were exported and used to drive the musculoskeletal model to gain insight into the kinematics of the musculotendinous units.

Musculoskeletal Model

A three dimensional musculoskeletal model was created in OpenSim [107] to examine plantar flexor musculotendon kinematics during stance phase. The model was modified from the lower limb model described by Arnold et al. [156] in the following ways. First, only the involved limb and the extrinsic foot muscles were retained. Secondly, two additional degrees of freedom were added at the knee joint to allow for varus-valgus movements and internal-external rotation. An additional degree of freedom was also added at the ankle joint to allow for internal-external rotation. Finally, a midfoot joint was added to the model. The joint center was positioned midway between the 5th metatarsal and navicular markers and the joint was allowed three degrees of freedom to rotate about the X, Y, and Z axes. Therefore, the final model contained 18 degrees of freedom and consisted of the following segments: pelvis, femur (with patella), tibia and fibula, rearfoot (talus and calcaneus), and forefoot (tarsal and metatarsals). Muscles included in the final model and subsequent analyses were the medial and lateral gastrocnemius (MG and LG), soleus (SOL), tibialis posterior (TibP), flexor digitorum longus (FDL), flexor hallucis longus (FHL), and peroneus longus (PL). Additionally, the tibialis anterior (TibA), extensor digitorum longus (EDL), extensor hallucis longus (EHL), peroneus brevis (PB), and peroneus tertius (PT) were included in the model, but the kinematics of these muscles were not analyzed in the present study. An illustration of the final model used is shown in Figure 5.1

Marker coordinates from the static trial were used to scale the model to match each individual subject's anthropometrics. Specifically, the ratio of the distance between two actual markers on the subject and the same two virtual markers on the

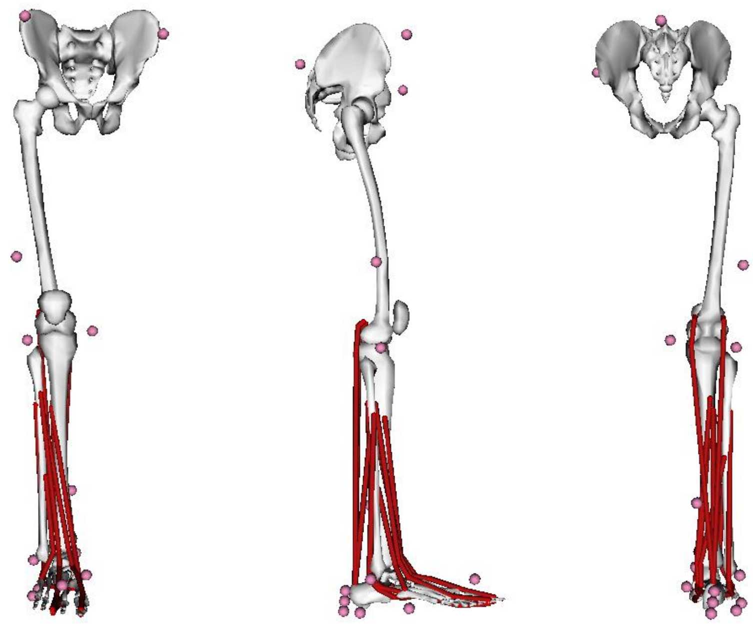

Figure 5.1. Anterior, lateral, and posterior views of the musculoskeletal model used in this study. A mirror image was used for left limbs, when appropriate.

musculoskeletal model was used to compute scale factors for each segment. These scale factors were then used to scale segment lengths and muscle attachment points based on each individual subject's anthropometry [107]. Rearfoot and forefoot segments were scaled using the same scale factors. Since the quality of the scaling has direct implications for the quality of the simulation, it was run iteratively until the root mean square error and absolute maximum error between the actual and virtual markers reached acceptable levels [161].

After scaling each model, the previously calculated joint angles were used as inputs to drive the model through the stance phase of the gait cycle. Therefore, each model exhibited the same kinematics as the subject, as determined based on the motion capture data. The model output consisted of the simulated lengths of the MG, LG, SOL, TibP, FDL, FHL, and PL muscles at each instant during stance phase. For each muscle the maximum (L_{max}) and minimum (L_{min}) musculotendinous lengths and the percent stance at which the maximum ($\%L_{max}$) and minimum ($\%L_{min}$) lengths occurred were identified. Musculotendon lengthening or shortening velocities were calculated by differentiating the musculotendon lengths with respect to time and then maximum ($\dot{L}_{max}$) and minimum ($\dot{L}_{min}$) lengthening or shortening velocities and the percent stance at which the maximum ($\%\dot{L}_{max}$) and minimum ($\%\dot{L}_{min}$) velocities occurred were identified.

From the length data, musculotendinous strain was estimated by calculating the percent elongation (increase or decrease) based on length in the static trial by

$$\%\ \text{Elongation}\ (\varepsilon^{\text{MT}}) = \frac{\text{L}_i - \text{L}}{\text{L}}$$

where L_i is the length at the *i*th percent of stance and L is the resting length in the static trial [151,154]. Maximum values of musculotendinous percent elongation (ε^{MT}_{max}) and the percent stance at which the maximum percent elongation values occurred ($\%\varepsilon^{MT}_{max}$) were identified. The rate of change in musculotendinous percent elongation was calculated by differentiating ε^{MT} with respect to time and the maximum ($\dot{\varepsilon}^{MT}_{max}$) percent elongation rate and percent stance at which the maximum percent elongation rate occurred ($\%\dot{\varepsilon}^{MT}_{max}$) where then identified. Finally, to examine the evolution of musculotendinous percent elongation over time, the average musculotendinous percent elongation between 0% - 20% ($\varepsilon^{MT}_{0\text{-}20}$), 20 - 60% ($\varepsilon^{MT}_{20\text{-}60}$), and 60 – 100% ($\varepsilon^{MT}_{60\text{-}100}$) of stance were calculated. These ranges were specifically chosen as they correspond to the initial contact, midstance, and propulsive periods within the stance phase of running [39].

Statistical Analysis

Linear regression was used to evaluate the quality of the scaling of the musculoskeletal model. The model's segment and muscle lengths should scale with each individual subject's anthropometrics. Therefore, each subject's shank length was regressed against the muscle length from the static calibration trial. Additionally, one sample *t*-tests were used to examine whether the RMS and maximum marker errors were below the suggested cutoffs of 1 cm and 2 cm, respectively [161]. Significant *t*-tests and significant regressions with R^2 values greater than 0.30 were interpreted as indicative of good scaling.

Between groups comparisons of musculotendinous kinematics were performed twice, once to compare differences between individuals with Achilles tendinopathy (AT), their respective control subjects (CON_AT), and non-injured individuals who demonstrate prolonged pronation (PP) and once to compare individuals with medial tibial stress syndrome (MTSS), their respective controls (CON_MTSS), and the same group of prolonged pronators (PP). Since the anatomic structures implicated in the development of AT and MTSS are different a combined pooled injury group comparison was not performed.

For each of the seven muscles analyzed, group differences in the musculotendon kinematic measures (L_{max}, L_{min}, $\%L_{max}$, $\%L_{min}$, $\dot{L}_{max}$, $\dot{L}_{min}$, $\%\dot{L}_{max}$, $\%\dot{L}_{min}$, $\varepsilon^{MT}{}_{max}$, $\%\varepsilon^{MT}{}_{max}$, $\dot{\varepsilon}^{MT}{}_{max}$, $\%\dot{\varepsilon}^{MT}{}_{max}$, $\varepsilon^{MT}{}_{0\text{-}20}$, $\varepsilon^{MT}{}_{20\text{-}60}$, and $\varepsilon^{MT}{}_{60\text{-}100}$) were evaluated using a 3x2 analysis of variance (ANOVA), with group and foot strike pattern being the two independent variables. Group was a categorical variable with three levels, injured (AT or MTSS), control (CON_AT or CON_MTSS), and non-injured prolonged pronators (PP). Foot strike pattern was included as a second independent variable since several of the kinematic variables examined in this study could vary with foot strike pattern [124]. Foot strike was treated as a categorical variable with two levels, rearfoot (RFS) or mid/forefoot strike (M/FFS), classified based on SI values less than 33% or greater than 33%, respectively. Running speed was entered as a covariate since subjects ran at self-selected speeds. Statistical significance was indicated when $\alpha < .05$. Where main effects of group were observed, post-hoc comparisons were conducted using a Bonferroni correction. Effect sizes (Cohen's f) were calculated for all comparisons to aid in the interpretation of results. Effect sizes of < 0.25, 0.25 – 0.40, and > 0.40 were used to indicate small, medium, and large effects, respectively [137].

Results

Musculoskeletal Model Scaling

Root mean squared errors (RMS) and the maximum marker error for each subject are shown in Table 5.1. One sample *t*-tests revealed the grand mean RMS errors were less than 1.0 cm ($p < .001$) and the grand mean maximum marker errors were less than 2.0 cm ($p < .001$). For all seven muscles the regression analyses between each subject's shank length and the muscle static length yielded significant relationships with $p < .001$ (Figure 5.2). R^2 values ranged from 0.30 for the FHL to 0.52 for the FDL.

Musculotendon Kinematics: AT vs. CON_AT vs. PP

Musculotendon lengths of the seven muscles throughout stance for the AT, CON_AT, and PP subjects are shown in Figure 5.3. For all three groups, the SOL demonstrated the largest increase from resting length of the seven muscles. There were no significant main effects of group for maximum (L_{max}) or minimum (L_{min})

Table 5.1. RMS and maximum marker error resulting from scaling the musculoskeletal model. Results are presented both by group and as a grand mean.

Subject	RMS (cm)	Max Error (cm)	Subject	RMS (cm)	Max Error (cm)
AT1	1.34	3.33	CON_AT1	1.09	2.61
AT2	0.82	1.44	CON_AT2	0.94	1.68
AT3	0.71	1.28	CON_AT3	0.76	1.30
AT4	0.82	1.61	CON_AT4	0.78	1.17
AT5	0.63	1.18	CON_AT5	0.89	1.75
AT6	0.41	0.62	CON_AT6	0.66	0.99
AT7	0.96	1.69	CON_AT7	0.51	1.07
AT8	0.91	1.59	CON_AT8	0.21	0.48
AT9	0.98	1.35	CON_AT9	0.94	1.56
AT10	0.45	1.07	CON_AT10	0.96	1.48
AT11	0.68	0.94	CON_AT11	0.89	1.63
AT12	1.00	1.91	CON_AT12	0.67	1.52
AT13	0.27	0.49	CON_AT13	1.49	3.13
Mean	0.78	1.42	Mean	0.83	1.57
St. Dev.	± 0.29	± 0.71	St. Dev.	± 0.30	± 0.68
Subject	RMS (cm)	Max Error (cm)	Subject	RMS (cm)	Max Error (cm)
MTSS1	0.95	1.97	CON_MTSS1	0.25	0.43
MTSS2	1.26	1.26	CON_MTSS2	0.92	1.82
MTSS3	0.99	1.80	CON_MTSS3	0.87	1.54
MTSS4	0.85	1.30	CON_MTSS4	0.66	1.38
MTSS5	0.87	1.26	CON_MTSS5	0.68	1.18
MTSS6	0.71	1.06	CON_MTSS6	0.95	1.83
MTSS7	0.21	2.27	CON_MTSS7	0.51	1.51
MTSS8	0.95	1.94	CON_MTSS8	0.93	1.70
Mean	0.85	1.61	Mean	0.72	1.42
St. Dev.	± 0.30	± 0.44	St. Dev.	± 0.25	± 0.46
Subject	RMS (cm)	Max Error (cm)	Subject	RMS (cm)	Max Error (cm)
PP1	0.89	1.51	Grand Mean	0.79	1.49
PP2	1.11	1.89	Grand St. Dev.	± 0.29	± 0.62
PP3	0.95	1.98			
PP4	0.84	1.35			
PP5	1.65	3.31			
PP6	0.39	0.72			
PP7	0.67	1.29			
PP8	0.96	1.97			
PP9	0.29	0.51			
PP10	0.66	1.53			
PP11	0.69	1.12			
PP12	0.40	0.82			
PP13	0.62	1.08			
Mean	0.78	1.47			
St. Dev.	± 0.36	± 0.72			

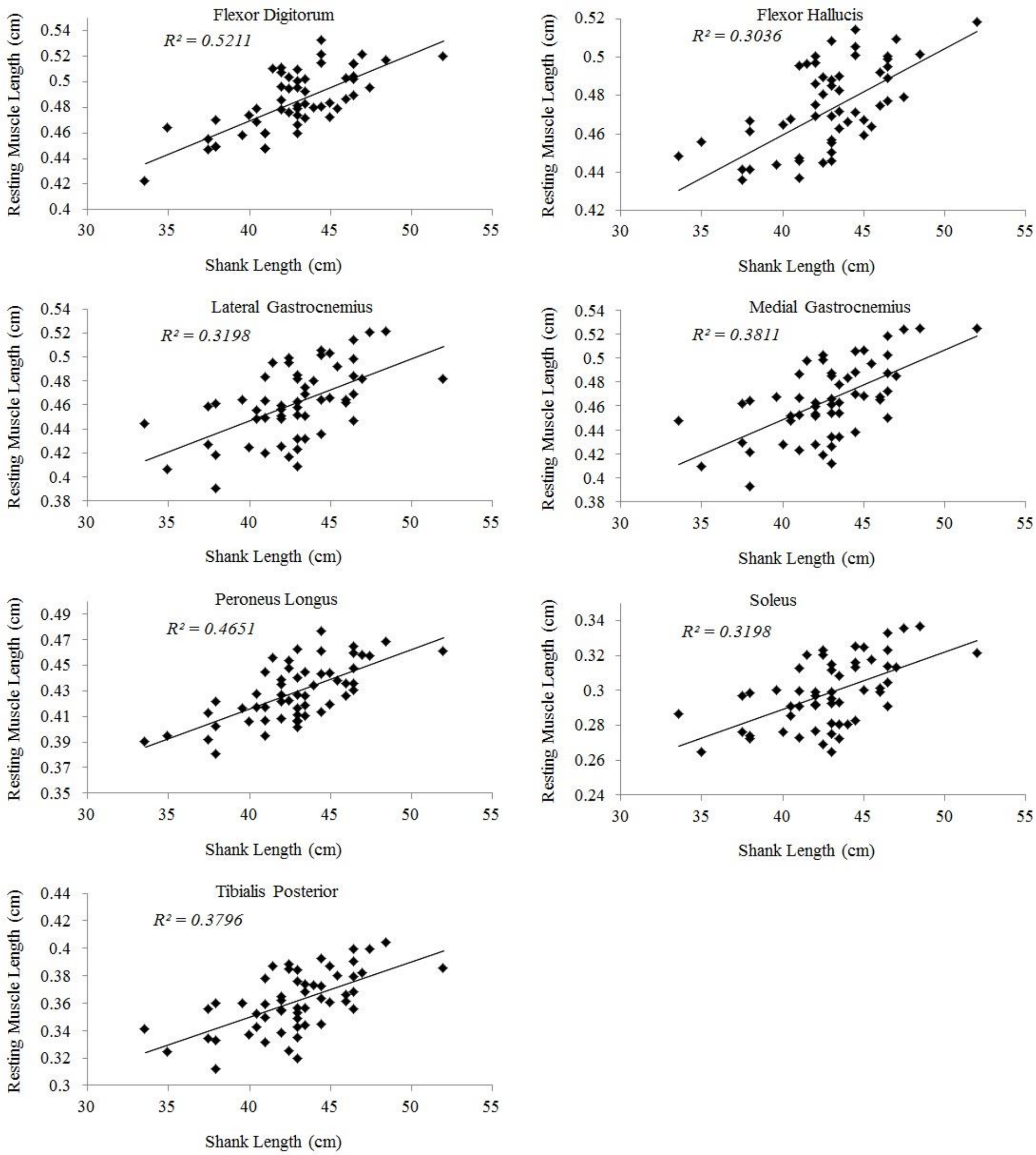

Figure 5.2. Regression plots showing each individual subject's shank length regressed against the resting muscle length from the static pose.

musculotendon length or the percent stance at which maximum (%$L_{\max}$) and minimum (%$L_{\min}$) lengths occurred. However, for both the FHL and LG muscles, there were main effects of foot strike, with %$L_{\max}$ occurring later in stance for subjects who used a RFS

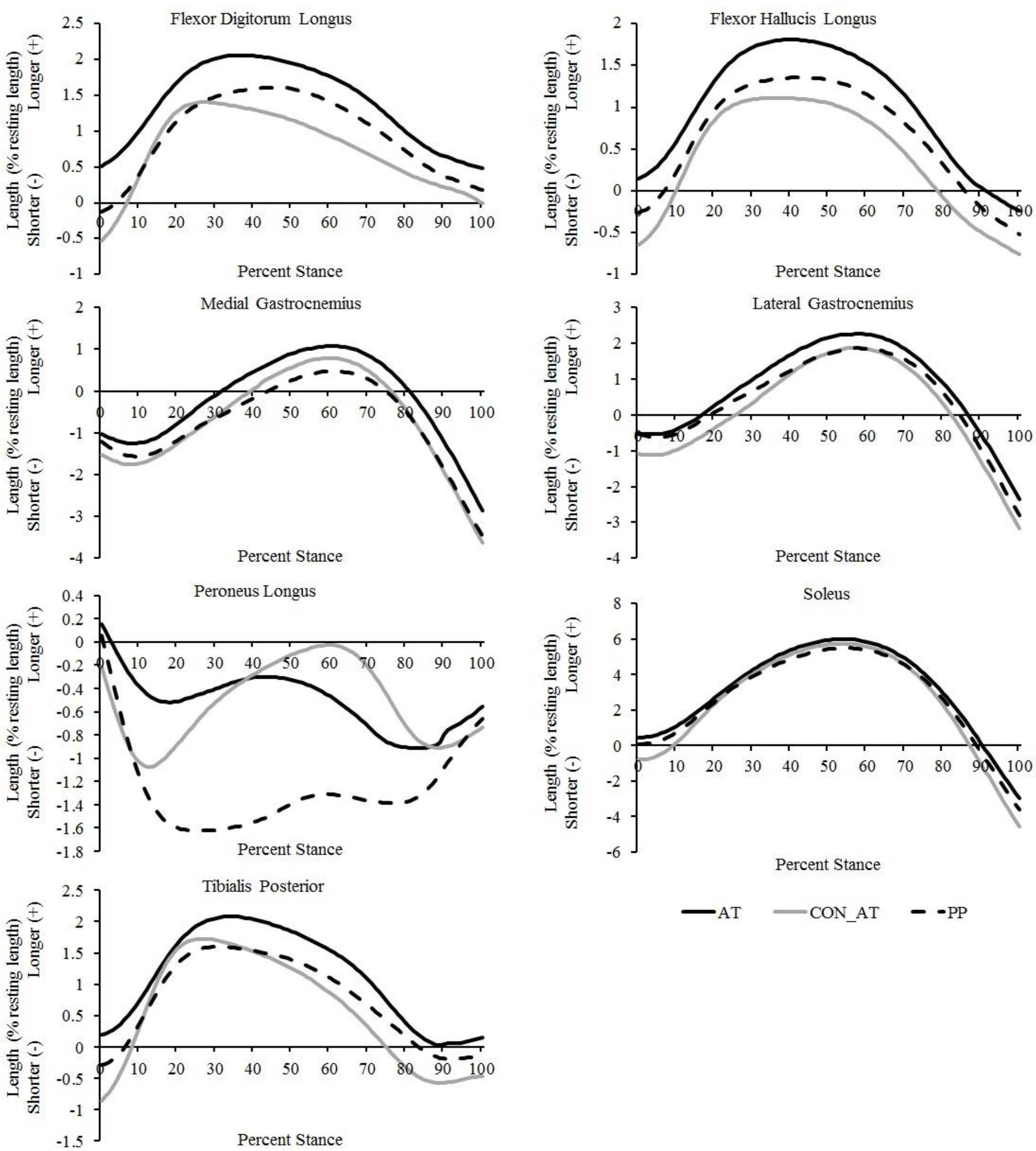

Figure 5.3. Average musculotendon lengths across stance for the AT, CON_AT, and PP groups. Lengths are expressed as percent increase or percent decrease from resting length.

than for subjects who used a M/FFS. Statistical comparisons between AT, CON_AT, and PP groups which resulted in a main effect of foot strike are summarized in Table 5.2.

Table 5.2. Statistical comparisons resulting in a main effect of foot strike for comparisons between AT, CON_AT, and PP groups.

Variable	RFS	M/FFS	*p*	ES
Flexor Hallucis Longus				
$\%L_{max}$ (% stance)	46.04 (± 15.18)	34.60 (± 15.87)	.045	0.369
$\%\dot{L}_{max}$ (%stance)	16.04 (± 9.41)	10.07 (± 4.59)	.020	0.433
$\%\varepsilon^{MT}{}_{max}$ (%)	46.67 (± 15.02)	33.60 (± 15.31)	.008	0.495
Lateral Gastrocnemius				
$\%L_{max}$ (% stance)	63.46 (± 9.73)	50.60 (± 15.44)	.003	0.562
$\dot{L}_{max}$ (m/s)	0.16 (± 0.05)	0.26 (± 0.11)	.001	0.636
$\%\dot{L}_{max}$ (% stance)	36.83 (± 10.62)	21.93 (± 11.93)	.001	0.667
$\%\varepsilon^{MT}{}_{max}$ (% stance)	63.46 (± 9.79)	50.60 (± 15.45)	.003	0.562
$\dot{\varepsilon}^{MT}{}_{max}$ (%/s)	41.29 (± 15.14)	75.72 (± 27.37)	< .001	0.851
$\%\dot{\varepsilon}^{MT}{}_{max}$ (%)	36.13 (± 14.09)	20.40 (± 11.71)	< .001	0.731
Medial Gastrocnemius				
$\dot{L}_{max}$ (m/s)	0.16 (± 0.05)	0.26 (± 0.09)	.001	0.636
$\%\dot{L}_{max}$ (% stance)	36.83 (± 10.62)	21.93 (± 11.93)	.001	0.667
$\dot{\varepsilon}^{MT}{}_{max}$ (%/s)	39.01 (± 18.29)	60.95 (± 24.94)	.007	0.516
$\%\dot{\varepsilon}^{MT}{}_{max}$ (%)	35.04 (± 10.87)	23.20 (± 12.72)	.009	0.500
Soleus				
$\dot{L}_{max}$ (m/s)	0.24 (± 0.08)	0.36 (± 0.12)	.001	0.633
$\%L_{max}$ (% stance)	57.33 (± 5.26)	51.47 (± 4.91)	.001	0.617
$\%\varepsilon^{MT}{}_{max}$ (%)	57.33 (± 5.26)	51.47 (± 4.92)	.001	0.617
$\dot{\varepsilon}^{MT}{}_{max}$ (%/s)	85.31 (± 29.88)	137.95 (± 48.22)	< .001	0.713
$\%\dot{\varepsilon}^{MT}{}_{max}$ (%)	22.71 (± 5.88)	13.13 (± 2.59)	< .001	1.175

While not statistically significant, a large between group effect size (> 0.40) was observed for $\%L_{min}$ in the TibP. All other effect sizes indicated small to moderate effects.

Musculotendon lengthening and shortening velocities for the CON_AT, AT, and PP groups are shown in Figure 5.4. There were no significant main effects of group for maximum musculotendon lengthening ($\dot{L}_{max}$) or shortening ($\dot{L}_{min}$) velocities or for the percent stance at which maximum lengthening ($\%\dot{L}_{max}$) or shortening ($\%\dot{L}_{min}$) velocities occurred. However, for the FHL there was a significant main effect of foot strike for

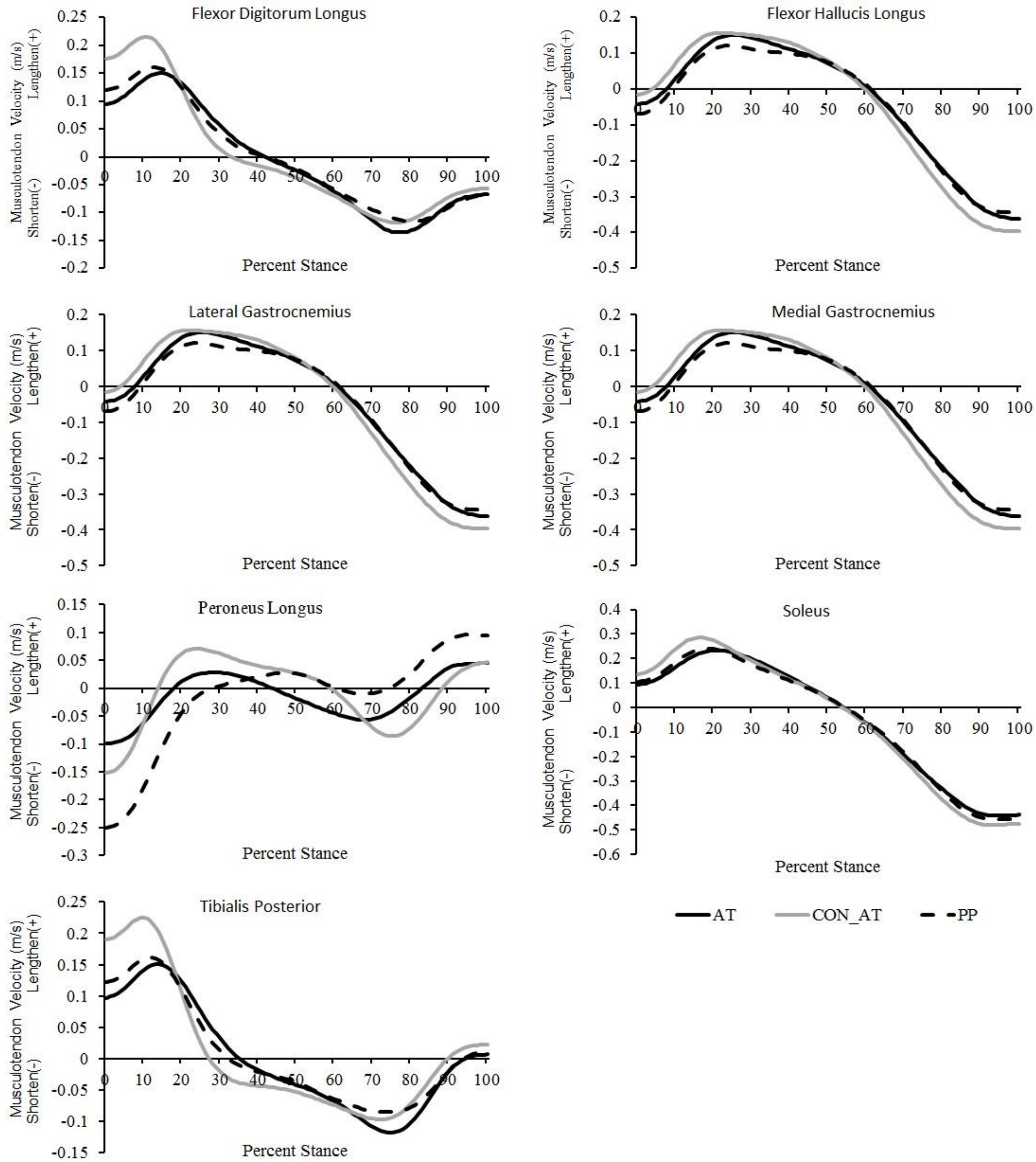

Figure 5.4. Mean musculotendon lengthening and shortening velocities, in meters per second, for the AT, CON_AT, and PP groups.

%$\dot{L}_{\max}$, with %$\dot{L}_{\max}$ occurring later for subjects who used a RFS compared to subjects who used a M/FFS (Table 5.2). Significant main effects of foot strike were also observed for the LG, MG, and SOL for both $\dot{L}_{\max}$ and %$\dot{L}_{\max}$. For all three muscles $\dot{L}_{\max}$ was lower and

$\%\dot{L}_{max}$ occurred later in stance in subjects who used a RFS than in those who used a M/FFS (Table 5.2). While not statistically significant, a large between group effect size (> 0.40) was observed for $\dot{L}_{min}$ in the PL. All other effect sizes indicated small to moderate effects.

Peak musculotendinous percent elongation ($\varepsilon^{MT}{}_{max}$), the percent stance at which peak percent elongations occurred ($\%\varepsilon^{MT}{}_{max}$), peak percent elongation rates ($\dot{\varepsilon}^{MT}{}_{max}$), and the percent stance at which peak percent elongation rates ($\%\dot{\varepsilon}^{MT}{}_{max}$) occurred are shown in Table 5.3. As with the length and velocity parameters, no significant main effects of group were observed. However, there were numerous main effects of foot strike. For the FHL, LG, and SOL peak musculotendinous percent elongation occurred later during stance for subjects who used a RFS than for those who used a M/FFS (Table 5.2). For the LG, MG, and SOL, peak musculotendinous percent elongation rates were lower and peak percent elongation rates were reached later in stance in subjects who used a RFS than in those who used a M/FFS (Table 5.2). All effect sizes were small to moderate.

Mean values for $\varepsilon^{MT}{}_{0\text{-}20}$, $\varepsilon^{MT}{}_{20\text{-}60}$, and $\varepsilon^{MT}{}_{60\text{-}100}$ are shown in Figure 5.5 (FDL, FHL, LG and MG) and Figure 5.6 (PL, SOL, TibP). While there were no consistent patterns between muscles or time periods within stance, all seven muscles demonstrate significant differences between groups for at least one of the three time points (Figures 5.5 and 5.6).

Table 5.3. Mean values (± standard deviation) for ε^{MT}_{max}, %ε^{MT}_{max}, $\dot{\varepsilon}^{MT}_{max}$, and %$\dot{\varepsilon}^{MT}_{max}$ for the CON_AT, AT, and PP groups. ^{F} indicates a significant main effect of foot strike. See Table 5.2 for detailed comparison of foot strikes.

Flexor Digitorum Longus	CON	AT	PP	ES
ε^{MT}_{max} (%)	1.68 (± 1.15)	2.23 (± 1.36)	1.92 (± 0.85)	0.229
%ε^{MT}_{max} (%)	36.00 (± 18.48)	44.38 (± 18.42)	47.69 (± 26.43)	0.287
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	56.14 (± 26.79)	41.00 (± 23.97)	45.53 (± 18.68)	0.339
%$\dot{\varepsilon}^{MT}_{max}$ (%)	14.46 (± 13.91)	15.00 (± 4.72)	25.62 (± 8.79)	0.348
Flexor Hallucis Longus				
ε^{MT}_{max} (%)	1.95 (± 1.72)	1.37 (± 0.97)	1.51 (± 0.90)	0.263
%ε^{MT}_{max} (%)	**43.92 (± 12.05)**	**36.23 (± 18.94)**	**44.77 (± 16.99)** ^{F}	**0.366**
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	40.75 (± 22.11)	54.97 (± 25.49)	45.51 (± 22.86)	0.225
%$\dot{\varepsilon}^{MT}_{max}$ (%)	16.08 (± 4.07)	14.62 (± 10.39)	13.92 (± 4.821)	0.095
Lateral Gastrocnemius				
ε^{MT}_{max} (%)	1.97 (± 2.04)	2.61 (± 2.06)	2.01 (± 0.97)	0.163
%ε^{MT}_{max} (%)	**54.00 (± 17.02)**	**60.77 (± 13.10)**	**60.38 (± 9.42)** ^{F}	**0.360**
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	**61.23 (± 30.79)**	**58.43 (± 27.69)**	**43.94 (± 19.97)** ^{F}	**0.383**
%$\dot{\varepsilon}^{MT}_{max}$ (%)	**29.69 (± 12.24)**	**31.38 (± 13.17)**	**33.85 (± 14.50)** ^{F}	**0.212**
Medial Gastrocnemius				
ε^{MT}_{max} (%)	0.84 (± 1.45)	1.36 (± 1.22)	0.59 (± 0.62)	0.316
%ε^{MT}_{max} (%)	50.77 (± 22.39)	60.53 (± 14.19)	56.46 (± 17.38)	0.318
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	**51.28 (± 19.82)**	**52.07 (± 30.35)**	**38.99 (± 17.44)** ^{F}	**0.318**
%$\dot{\varepsilon}^{MT}_{max}$ (%)	**31.38 (± 13.72)**	**29.23 (± 12.56)**	**30.50 (± 13.25)** ^{F}	**0.112**
Peroneus Longus				
ε^{MT}_{max} (%)	0.65 (± 2.62)	0.76 (± 1.75)	0.35 (± 1.66)	0.084
%ε^{MT}_{max} (%)	58.46 (± 32.48)	36.54 (± 39.35)	50.69 (± 44.22)	0.207
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	37.55 (± 29.03)	39.08 (± 33.70)	41.20 (± 26.83)	0.032
%$\dot{\varepsilon}^{MT}_{max}$ (%)	57.77 (± 34.51)	63.77 (± 36.30)	69.23 (± 30.44)	0.143
Soleus				
ε^{MT}_{max} (%)	5.79 (± 1.23)	6.17 (± 1.85)	5.52 (± 1.29)	0.175
%ε^{MT}_{max} (%)	**53.77 (± 4.29)**	**56.00 (± 8.31)**	**55.46 (± 4.18)** ^{F}	**0.259**
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	**117.98 (± 56.45)**	**104.55 (± 41.73)**	**94.16 (± 36.11)** ^{F}	**0.303**
%$\dot{\varepsilon}^{MT}_{max}$ (%)	**21.31 (± 8.54)**	**15.92 (± 5.58)**	**19.85 (± 5.49)** ^{F}	**0.505**
Tibialis Posterior				
ε^{MT}_{max} (%)	2.18 (± 1.29)	2.42 (± 2.32)	2.01 (± 1.13)	0.110
%ε^{MT}_{max} (%)	41.54 (± 24.42)	48.38 (± 25.95)	50.00 (± 31.08)	0.146
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	78.09 (± 34.39)	59.22 (± 31.70)	61.85 (± 27.75)	0.331
%$\dot{\varepsilon}^{MT}_{max}$ (%/s)	21.31 (± 25.82)	30.92 (± 31.37)	36.85 (± 40.41)	0.234

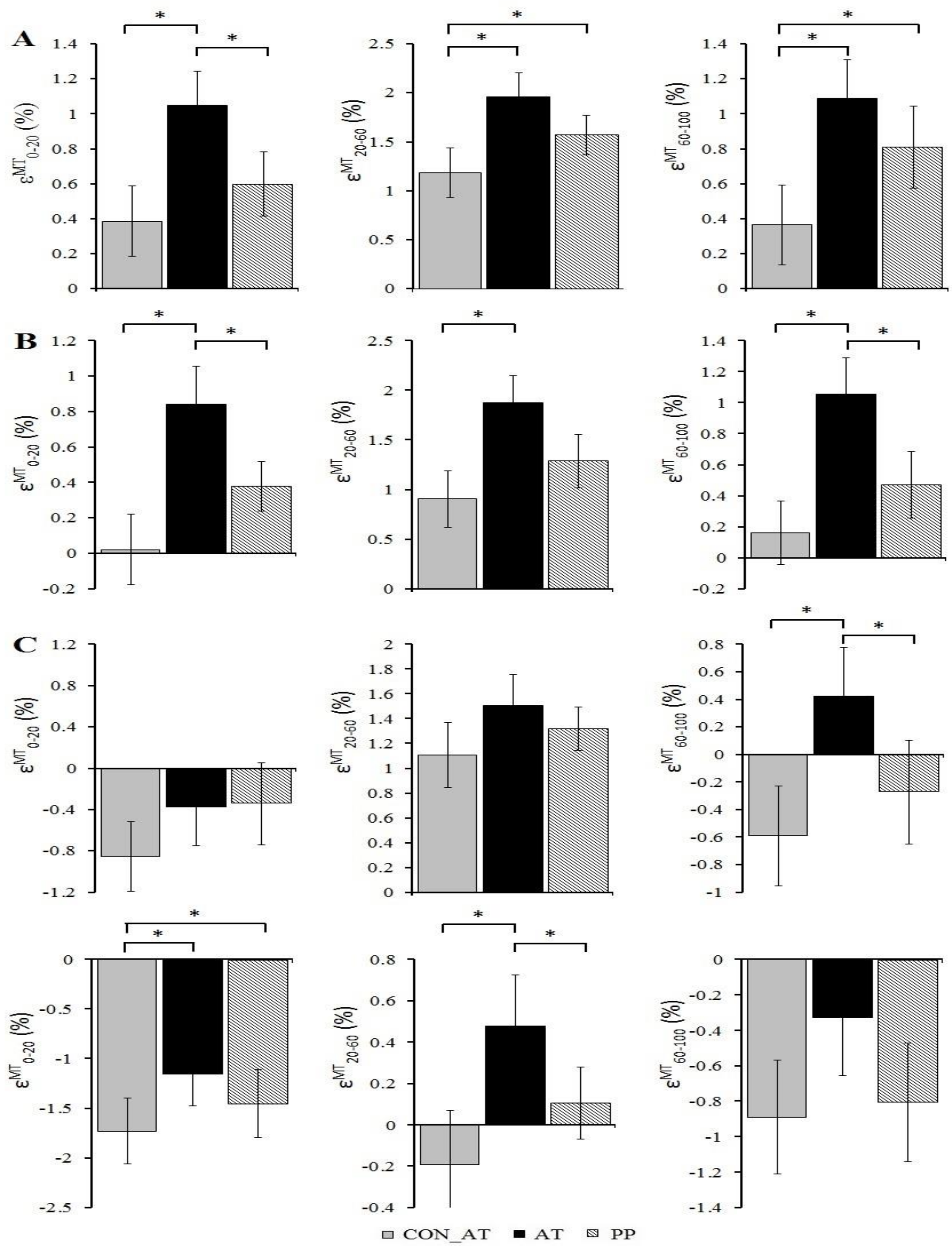

Figure 5.5. Average musculotendon percent elongation for 0-20% stance (left column), 20-60% stance (middle column) and 60-100% stance (right column) for the FDL (A), FHL (B), LG (C), and MG (D) for the CON_AT, AT, and PP groups. * indicates a main effect of group with a significant post-hoc comparisons at $p < .0166$.

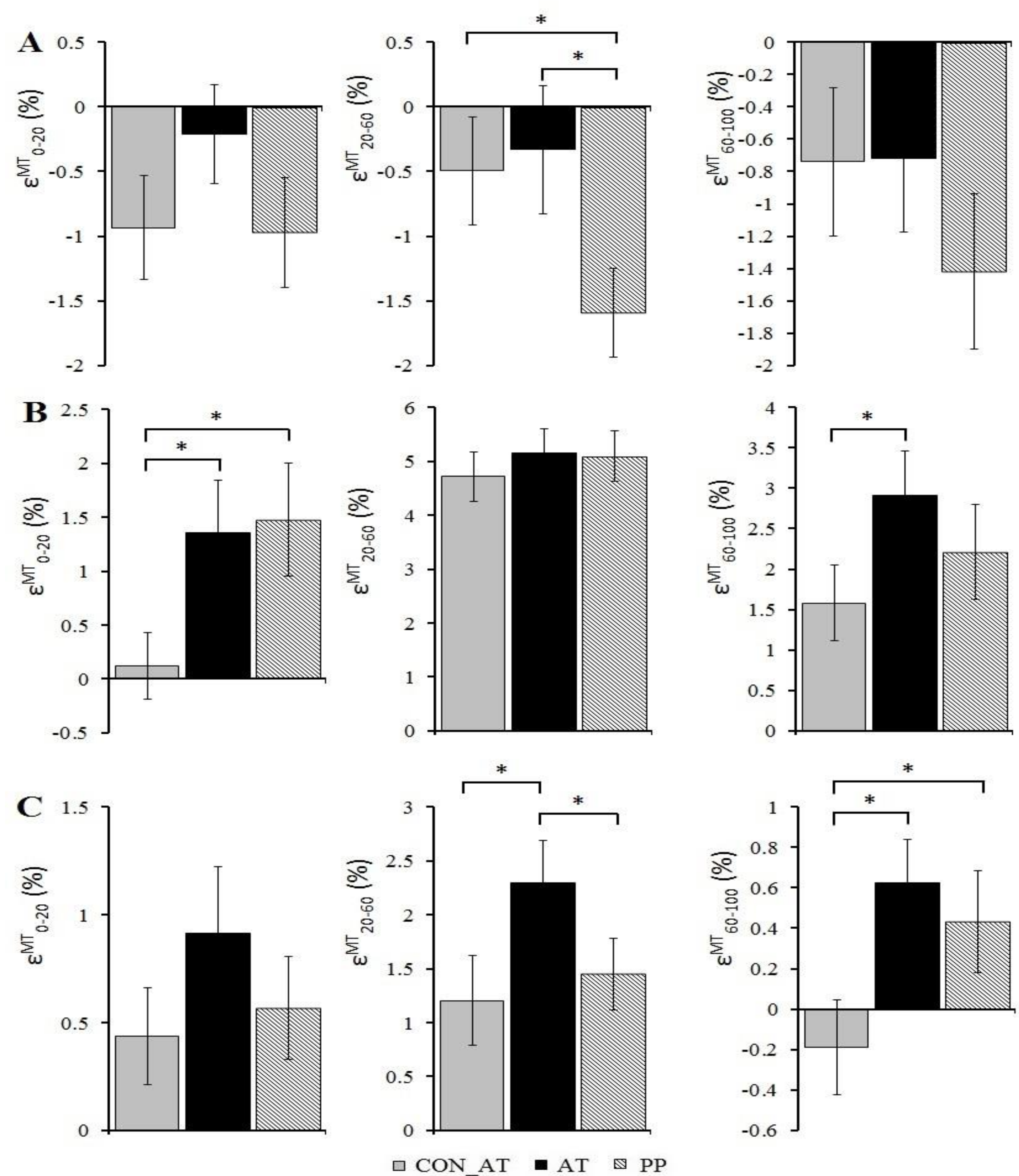

Figure 5.6. Average musculotendon percent elongation for 0-20% stance (left column), 20-60% stance (middle column) and 60-100% stance (right column) for the PL (A), SOL (B), and TibP (C) for the CON_AT, AT, and PP groups. * indicates a main effect of group with a significant post-hoc comparisons at $p < .0166$.

Musculotendon Kinematics: MTSS vs. CON_MTSS vs. PP

Musculotendon lengths of the seven muscles throughout stance for the MTSS, CON_MTSS, and PP subjects are shown in Figure 5.7. The only statistically significant main effect of group was observed in the SOL muscle, where peak musculotendinous length was longer in the MTSS group (106.60 ± 1.85 % resting length) than in the CON_AT group (104.12 ± 1.22 % resting length; p = .008). Peak length in the PP group (105.53 ± 1.29 % resting length) was also longer than the CON_AT group (p = .007). However, the MTSS and PP groups were not different (p = .251). No significant main effects of group or foot strike were observed for any of the four musculotendon length variables (L_{max}, L_{min}, $\%L_{max}$, and $\%L_{min}$) in the other six muscles. While not statistically significant, large between group effect sizes (> 0.40) were observed for L_{max} and $\%L_{max}$ in the FDL, $\%L_{min}$ in the FHL, $\%L_{max}$ in the LG, L_{max} in the MG, and L_{min} and $\%L_{min}$ in the TibP. All other effect sizes indicated small to moderate effects.

Musculotendon lengthening and shortening velocities are shown in Figure 5.8. The only comparison resulting in a significant main effect of group was for the percent stance at which maximum shortening velocity ($\%\dot{L}_{min}$) occurred in the TibP muscle. $\%\dot{L}_{min}$ occurred earlier in stance in the MTSS group than in the CON_MTSS group (p = .002) or in the PP group (p = .004). The timing of $\%\dot{L}_{min}$ was not different between the CON_MTSS and PP groups (p = .934). While not statistically significant, large between group effect sizes (ES > 0.40) were observed for $\%\dot{L}_{max}$ in the FHL and $\%\dot{L}_{min}$ in the PL. All other effect sizes indicated small to moderate effects.

However, there were numerous main effects of foot strike, which are summarized in Table 5.4. In the PL and TibP muscles, maximum lengthening velocity ($\%\dot{L}_{max}$)

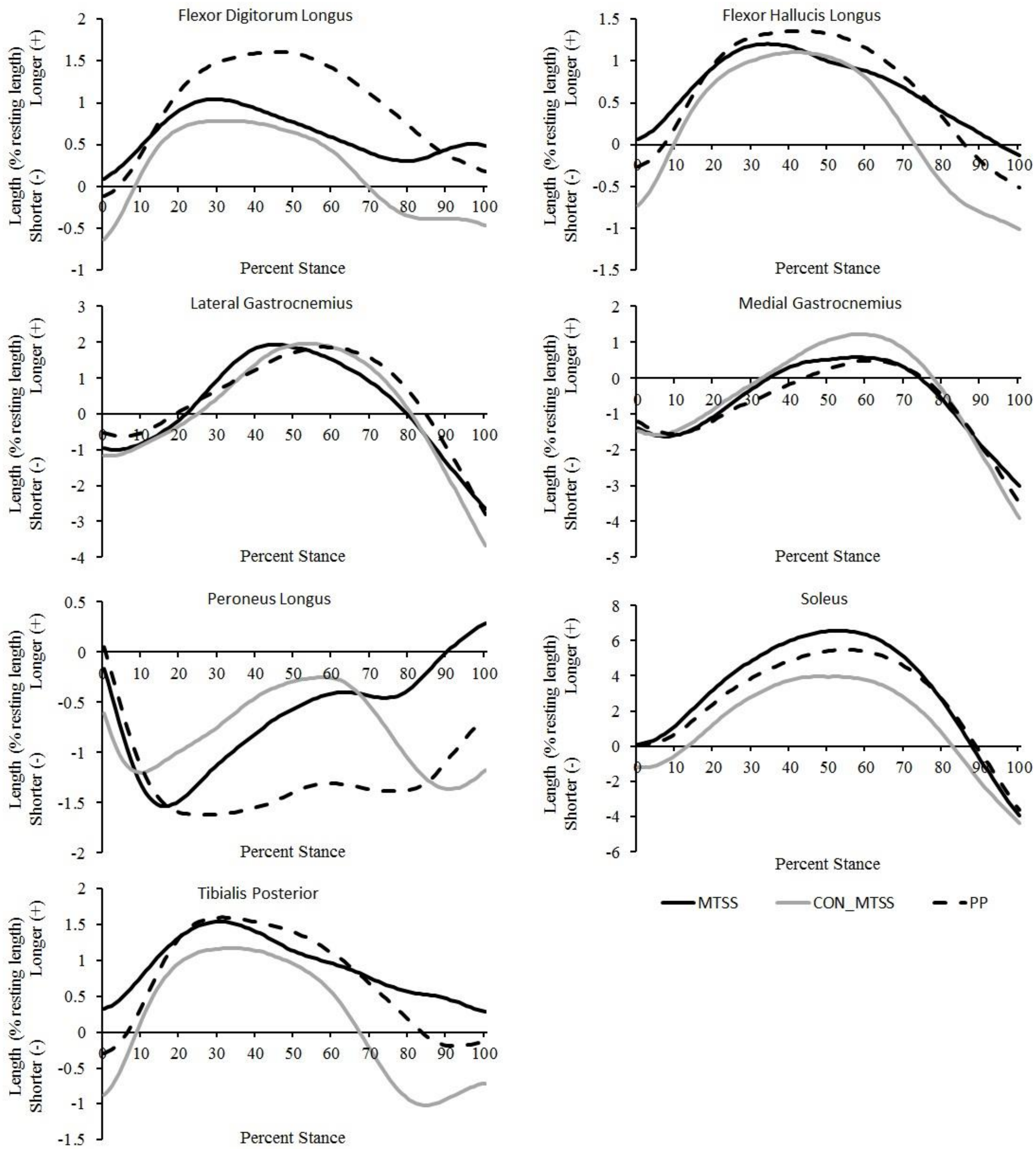

Figure 5.7. Average musculotendon lengths across stance for the MTSS, CON_MTSS, and PP groups. Lengths are expressed as percent increase or percent decrease from resting length.

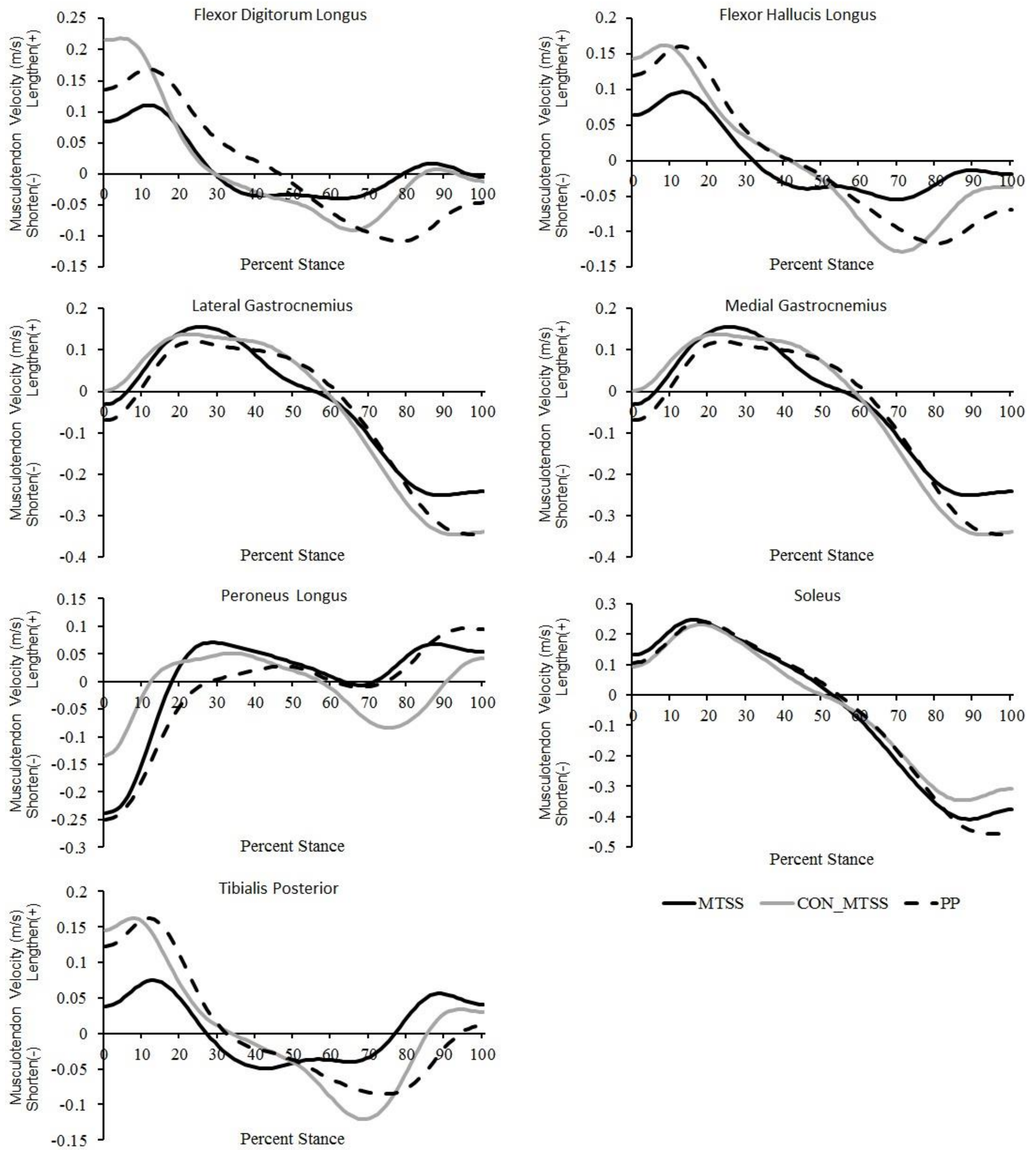

Figure 5.8. Average musculotendon lengthening and shortening velocities across stance for the MTSS, CON_MTSS, and PP groups. Velocities are in meters per second.

occurred earlier in stance for subjects who used a RFS than for subjects who used a M/FFS (Table 5.4). The opposite pattern was observed for the SOL muscle, where $\%\dot{L}_{max}$ occurred later in stance for subjects who used a RFS than for those who used a M/FFS

Table 5.4. Statistical comparisons resulting in a main effect of foot strike for comparisons between MTSS, CON_MTSS, and PP groups.

Variable	RFS	M/FFS	*p*	ES
Peroneus Longus				
$\%L_{max}$ (% stance)	51.61 (± 6.37)	80.08 (± 8.09)	.013	0.575
$\%\dot{\varepsilon}^{MT}{}_{max}$ (%)	50.13 (± 6.96)	80.13 (± 8.84)	.016	0.554
Soleus				
$\dot{L}_{max}$ (m/s)	0.23 (± 0.07)	0.34 (± 0.07)	.004	0.685
$\%\dot{L}_{max}$ (% stance)	21.16 (± 5.60)	12.70 (± 3.23)	.001	0.794
$\dot{\varepsilon}^{MT}{}_{max}$ (%/s)	81.37 (± 26.4)	116.67 (± 25.25)	.009	0.611
$\%\dot{\varepsilon}^{MT}{}_{max}$ (%)	20.37 (± 5.43)	12.70 (± 2.71)	.002	0.752
Tibialis Posterior				
$\%\dot{L}_{max}$ (% stance)	20.26 (± 25.25)	54.40 (± 45.55)	.043	0.457
$\%\dot{\varepsilon}^{MT}{}_{max}$ (%)	20.11 (± 25.26)	56.20 (± 44.45)	.028	0.502

(Table 5.4). Also in the SOL muscle, maximum lengthening velocity ($\dot{L}_{max}$) was lower for subjects who used a RFS than for subjects who used a M/FFS (Table 5.4).

Mean values for peak musculotendinous percent elongation ($\varepsilon^{MT}{}_{max}$), the percent stance at which peak percent elongation occurred ($\%\varepsilon^{MT}{}_{max}$), peak percent elongation rates ($\dot{\varepsilon}^{MT}{}_{max}$), and the percent stance at which peak percent elongation rates ($\%\dot{\varepsilon}^{MT}{}_{max}$) occurred are shown in Table 5.5. As with the length and velocity parameters, only one statistically significant main effect of group was observed. In the SOL muscle, peak musculotendinous percent elongations were higher in MTSS group (6.60 ± 1.84 %) than in the CON_MTSS group (4.12 ± 1.21 %; p = .007). Peak percent elongations were also higher in the PP group (5.52 ± 1.29 % stance) than the CON group (p = .008). The MTSS and PP groups were not different (p = .251). No other main effects of group were observed. While not statistically significant, large between group effect sizes

Table 5.5. Mean values (± standard deviation) for ε^{MT}_{max}, $\%\varepsilon^{MT}_{max}$, $\dot{\varepsilon}^{MT}_{max}$, and $\%\dot{\varepsilon}^{MT}_{max}$ for the MTSS, CON_MTSS, and PP groups. [G] indicates a main effect of group with significant post-hoc comparisons at $p < .0166$. [F] indicates a significant main effect of foot strike. See Table 5.4 for detailed comparison of foot strike effects.

Flexor Digitorum Longus	CON	MTSS	PP	ES
ε^{MT}_{max} (%)	1.11(± 0.58)	1.45 (± 0.98)	1.92 (± 0.85)	0.408
$\%\varepsilon^{MT}_{max}$ (%)	24.50 (± 13.67)	51.88 (± 37.40)	47.69 (± 26.43)	0.429
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	56.14 (± 26.79)	41.00 (± 23.97)	45.53 (± 18.68)	0.339
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	11.46 (± 13.91)	15.00 (± 8.72)	17.62 (± 8.79)	0.348
Flexor Hallucis Longus				
ε^{MT}_{max} (%)	1.26 (± 0.61)	1.52 (± 0.72)	1.51 (± 0.90)	0.156
$\%\varepsilon^{MT}_{max}$ (%)	40.50 (± 10.64)	40.75 (± 21.97)	44.77 (± 16.99)	0.084
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	40.75 (± 22.11)	54.97 (± 25.49)	45.51 (± 22.86)	0.225
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	11.08 (± 8.07)	17.62 (± 10.39)	12.92 (± 7.82)	0.446
Lateral Gastrocnemius				
ε^{MT}_{max} (%)	2.04 (± 0.56)	2.19 (± 1.44)	2.01 (± 0.97)	0.135
$\%\varepsilon^{MT}_{max}$ (%)	53.88 (± 4.67)	43.88 (± 18.96)	60.77 (± 9.42)	0.259
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	61.23 (± 30.79)	58.43 (± 27.69)	43.94 (± 19.97)	0.383
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	24.69 (± 12.24)	23.38 (± 9.17)	33.85 (± 15.50)	0.212
Medial Gastrocnemius				
ε^{MT}_{max} (%)	1.23 (± 0.93)	0.80 (± 0.60)	0.59 (± 0.61)	0.455
$\%\varepsilon^{MT}_{max}$ (%)	59.63 (± 2.50)	49.50 (± 21.34)	56.46 (± 17.84)	0.295
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	51.28 (± 19.82)	52.07 (± 30.35)	38.99 (± 17.44)	0.318
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	24.38 (± 12.72)	23.23 (± 9.56)	33.50 (± 14.25)	0.212
Peroneus Longus				
ε^{MT}_{max} (%)	0.34 (± 0.99)	1.01 (± 1.64)	0.35 (± 1.66)	0.160
$\%\varepsilon^{MT}_{max}$ (%)	40.75 (± 25.75)	53.37 (± 44.41)	50.69 (± 44.29)	0.259
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	37.55 (± 29.03)	39.08 (± 33.70)	41.20 (± 26.83)	0.032
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	**52.63 (± 33.52)**	**53.75 (± 34.84)**	**70.38 (± 33.19) [F]**	**0.143**
Soleus				
ε^{MT}_{max} (%)	**4.12 (± 1.21)**	**6.60 (± 1.84)**	**5.52 (± 1.29) [G]**	**0.659**
$\%\varepsilon^{MT}_{max}$ (%)	51.00 (± 6.09)	52.88 (± 2.35)	55.46 (± 4.18)	0.190
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	**89.53 (± 19.58)**	**96.56 (± 33.61)**	**94.16 (± 36.11) [F]**	**0.303**
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	**17.25 (± 5.62)**	**16.38 (± 7.23)**	**18.85 (± 5.49) [F]**	**0.169**
Tibialis Posterior				
ε^{MT}_{max} (%)	1.56 (± 0.82)	2.05 (± 1.34)	2.01 (± 1.13)	0.182
$\%\varepsilon^{MT}_{max}$ (%)	41.25 (± 27.46)	39.13 (± 23.63)	50.00 (± 31.41)	0.100
$\dot{\varepsilon}^{MT}_{max}$ (%/s)	78.09 (± 34.39)	59.22 (± 31.70)	61.85 (± 27.75)	0.331
$\%\dot{\varepsilon}^{MT}_{max}$ (%/s)	**20.50 (± 30.37)**	**48.25 (± 38.74)**	**30.31 (± 38.34) [F]**	**0.410**

(ES > .40) were observed for ε^{MT}_{max} in the FDL, MG, and the SOL muscles, as well as in $\%\dot{\varepsilon}^{MT}_{max}$ for the FHL muscle. All other effect sizes indicated small to moderate effects.

However, there were numerous main effects of foot strike observed in the FL, Sol, and TibP muscles (Table 5.4). For the PL and TibP muscles, $\%\dot{\varepsilon}^{MT}_{max}$ occurred earlier in stance for subjects who used a RFS than for subjects who used a M/FFS (Table 5.4). In contrast, for the SOL muscle, $\%\dot{\varepsilon}^{MT}_{max}$ occurred later in stance for subjects who used a RFS than for those who used a M/FFS (Table 5.4). $\dot{\varepsilon}^{MT}_{max}$ values were also lower in the SOL for subjects who used a RFS compared to those who used a M/FFS.

Mean values for $\varepsilon^{MT}_{0\text{-}20}$, $\varepsilon^{MT}_{20\text{-}60}$, and $\varepsilon^{MT}_{60\text{-}100}$ for the CON_MTSS, MTSS, and PP are shown in Figure 5.9 (FDL, FHL, LG and MG) and Figure 5.10 (PL, SOL, TibP). While there were no consistent patterns between muscles or time periods within stance, all muscles except MG demonstrated significant differences between groups for at least one of the three time points (Figures 5.9 and 5.10).

Discussion

The goal of this study was to utilize a three dimensional musculoskeletal model to characterize and compare plantar flexor musculotendon kinematics during the stance phase of running in individuals currently symptomatic with AT or mMTSS, healthy matched control subjects, and currently healthy individuals who demonstrate prolonged pronation. Only three statistically significant main effect of group were observed, two in the SOL muscle and one in the TibP muscle. In the SOL muscle, peak musculotendinous lengths were longer and peak musculotendinous percent elongations were higher in the MTSS group than the CON_MTSS group.

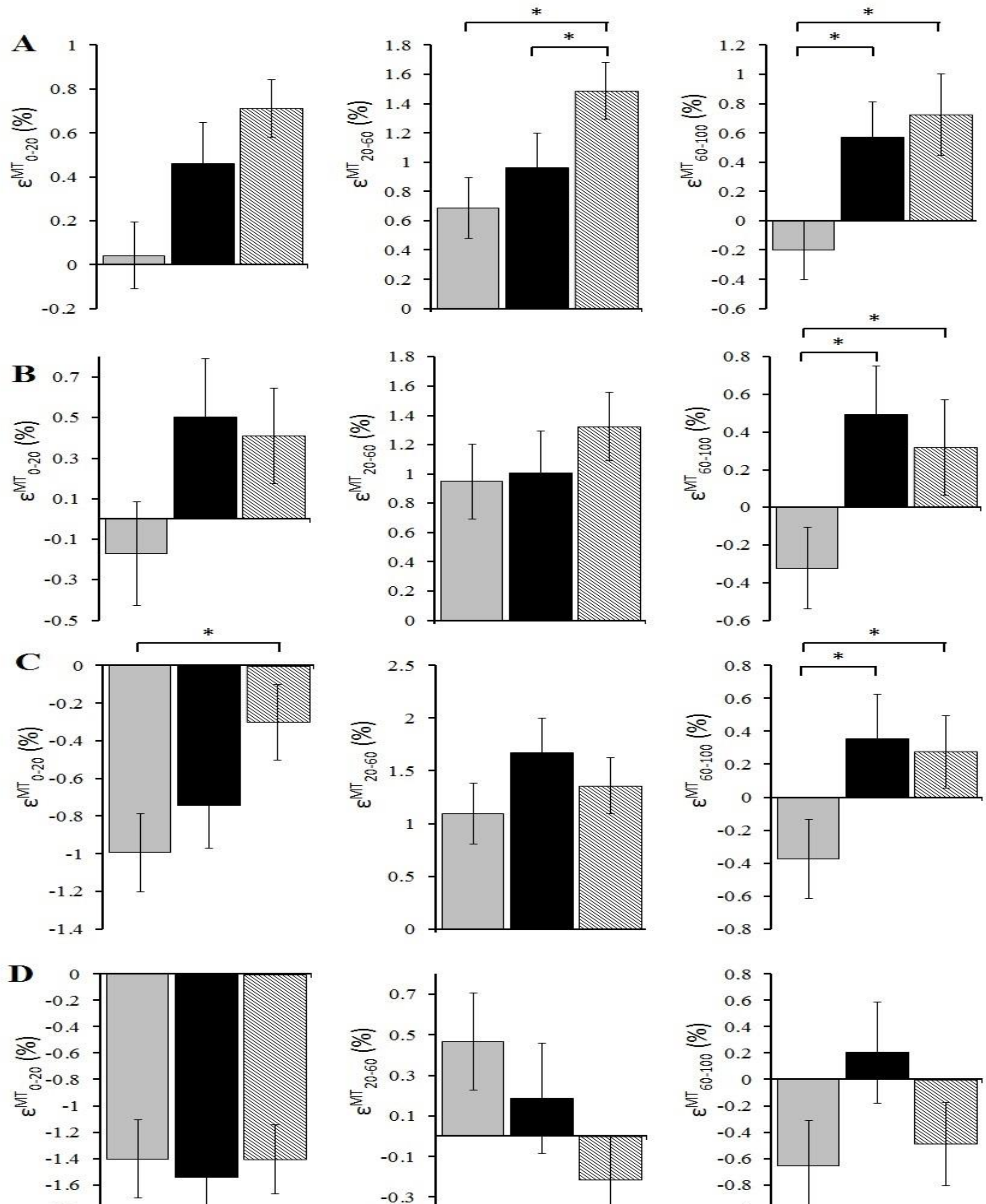

Figure 5.9. Average musculotendon percent elongations for 0-20% stance (left column), 20-60% stance (middle column) and 60-100% stance (right column) for the FDL (A), FHL (B), LG (C), and MG (D) for the CON_MTSS, MTSS, and PP groups. * indicates a main effect of group with a significant post-hoc comparisons at $p < .0166$.

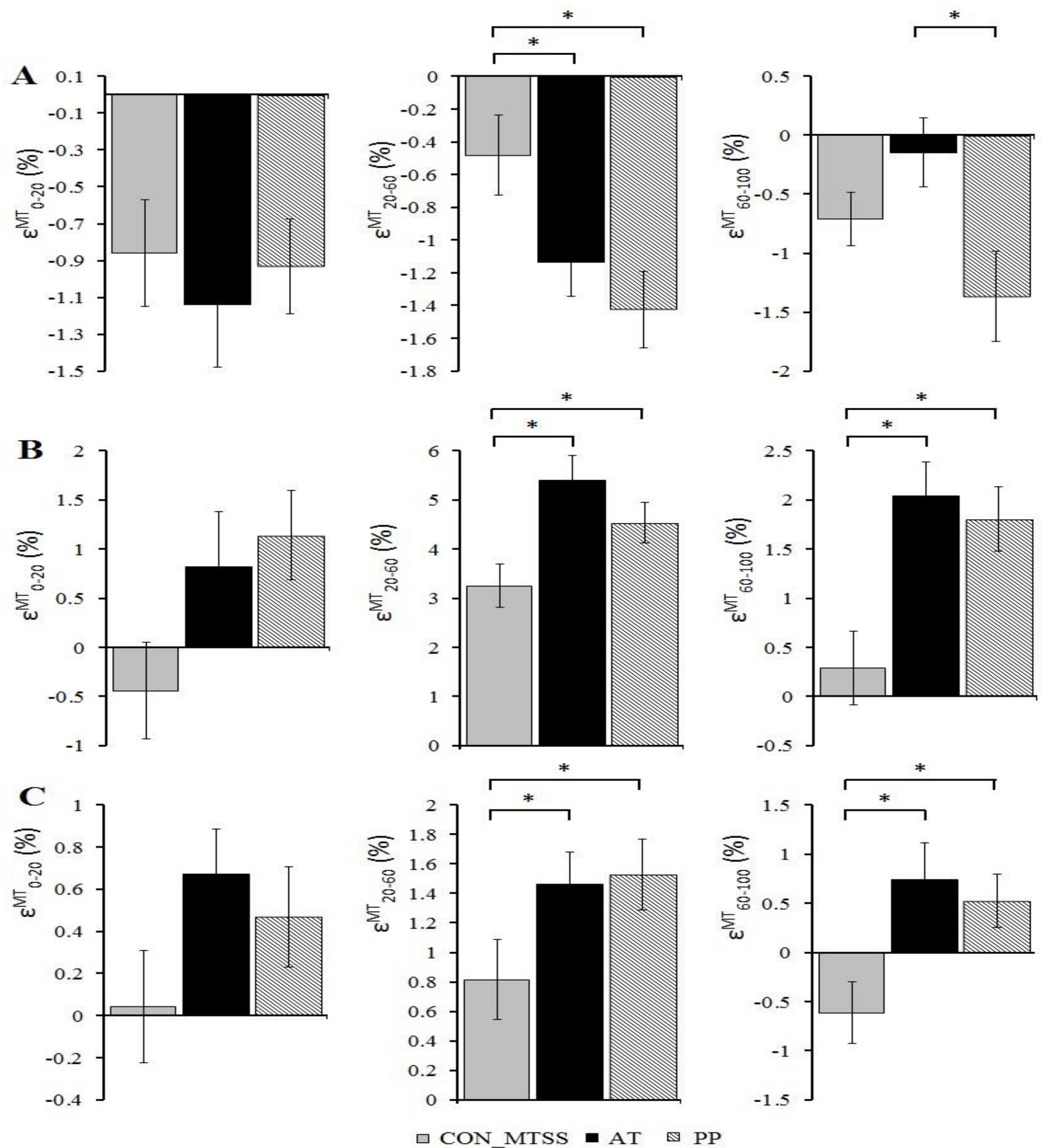

Figure 5.10. Average musculotendon percent elongations for 0-20% stance (left column), 20-60% stance (middle column) and 60-100% stance (right column) for the PL (A), SOL (B), and TibP (C) for the CON_MTSS, MTSS, and PP groups. * indicates a main effect of group with a significant post-hoc comparisons at $p < .0166$.

Values in the PP group were also higher than the CON_MTSS group, but were not different from the MTSS group. In the TibP, the percent stance at which maximum shortening velocity occurred was earlier in stance for subjects in the MTSS group compared to those in the CON_MTSS or PP groups. The CON_MTSS and PP groups were not different. No other statistically significant differences were observed for any of the musculotendinous length (L_{max}, L_{min}, $\%L_{max}$, $\%L_{min}$), lengthening or shortening velocities ($\dot{L}_{max}$, $\dot{L}_{min}$, $\%\dot{L}_{max}$, $\%\dot{L}_{min}$), percent elongation $\varepsilon^{MT}{}_{max}$, $\%\varepsilon^{MT}{}_{max}$) or percent elongation rate ($\dot{\varepsilon}^{MT}{}_{max}$, $\%\dot{\varepsilon}^{MT}{}_{max}$) variables. However, when musculotendinous percent elongations were averaged across different periods of stance phase ($\varepsilon^{MT}{}_{0\text{-}20}$, $\varepsilon^{MT}{}_{20\text{-}60}$, and $\varepsilon^{MT}{}_{60\text{-}100}$) numerous between group differences were observed. Additionally, numerous significant main effects foot strike pattern were observed for the musculotendon length, velocity, percent elongation, and percent elongation rate variables.

Most muscles displayed a unimodal length profile across stance. Compared to the resting musculotendon length in the static upright, musculotendon units were slightly shorter at initial contact, subsequently lengthened during midstance, length between 25 % stance (FDL in the CON_MTSS group) and 60% stance (LG and MG in the AT group), and then shortened until toe off (Figures 5.3 and 5.7). This pattern was observed for all muscles except the PL, which was shorter than it's resting length throughout most of stance and demonstrated a biomodal length profile (Figures 5.3 and 5.7).

The findings regarding musculotendon length and percent elongation in the SOL, FDL, and TibP have implications for the role of the extrinsic muscles in the development of MTSS. While numerous authors agree with a traction induced periostitis [68,71,72,74,162] or traction induced periosteal remodeling [76] being the underlying cause of the painful symptoms associated with MTSS, there is disagreement in the literature regarding exactly which muscles are involved. Some authors have suggested the TibP is involved [163,164] while others, based on anatomic dissections, have argued the FDL or SOL muscles are the main culprits [71,72,74]. In the current study, peak musculotendinous lengths and percent elongations in the SOL muscle were higher in the MTSS and PP groups than the CON_MTSS group. While not statistically significant, in the FDL these same comparisons both resulted in large effect sizes (ES > 0.40). However, no significant between group differences, and only small effect sizes, were observed for these comparisons in the TibP. While additional work is required to clarify the exact mechanisms behind MTSS, these results support the idea of the SOL and FDL being involved while discounting the involvement of the TibP.

Given proposed relationships between musculotendon percent elongation and the development of AT [82–84] and MTSS [68,75], it was hypothesized that the injured individuals would demonstrate higher peak musculotendinous percent elongations and percent elongation rates than the non-injured individuals. The results did not fully support this hypothesis as the only statistically significant between group differences in peak musculotendinous percent elongation was observed in the SOL, and only when muscle when comparing the MTSS, CON_MTSS, and PP groups. However, when peak musculoskeletal percent elongations were averaged between 0 – 20% stance, between 20 – 60% stance, and between 60 – 100% of stance, numerous between group differences were observed. In general, for all subjects, the average musculotendinous percent elongations were highest between 20 and 60% of stance. More specifically, average musculotendinous percent elongations between 0 and 20% of stance were higher in the

AT or MTSS group then their respective controls for 4 of the 14 comparisons. Between 20 and 60% of stance average musculotendinous percent elongation were higher in the AT or MTSS group then their respective controls for 8 of the 14 comparisons. Finally, between 60 and 100% of stance, average musculotendinous percent elongations were higher in the AT or MTSS group then in their respective controls for 10 of the 14 comparisons. When viewed as a whole this suggests that, on average, musculotendinous percent elongations in the injured subjects were higher than their matched controls late in stance phase with fewer differences occurring early in stance phase.

This finding also has implications for when during stance phase problematic mechanics related to the development of AT or MTSS are likely to occur. The hypothesis that excessive amounts or velocities of pronation are the dominant biomechanical factors involved in the development of these injuries implicates early stance phase as being the primary time of concern while the hypothesis that prolonged pronation is the more important factor implicates late stance phase. Assuming musculotendinous strain plays an important role in the development of AT [82–84] and MTSS [68,75], the finding that average musculotendinous percent elongations were highest during midstance and that the greatest number of differences between AT or MTSS and their respective controls were observed during propulsion late in stance, suggests mid to late stance may be the more important period of focus. This, in turn, supports the hypothesis that prolonged pronation, not excessive amounts or velocities of pronation, may be the more important factor in the development of AT or MTSS.

Further support for the prolonged pronation hypothesis can be observed through comparison of the timing of peak musculotendinous loading and peak pronation. Figure

4.8 in Chapter IV shows that peak eversion occurs around 30% stance for the AT, MTSS, CON_AT, and CON_MTSS subjects. Results from Chapter III show there was not a difference in time to peak eversion between the prolonged pronators and non-prolonged pronators, thus one could reasonably assume peak pronation occurs around 30% for this group as well. Tables 5.3 and 5.5 show peak musculotendinous lengths and percent elongations occurred, on average, around 50% of stance, well after this time peak eversion. Previous modeling studies have reported force produced by the LG, MG, and SOL muscles reaches a peak slightly after 50% of stance [21,155] while EMG data indicates these muscles are very active during the time period from 50% to 80% of stance phase [51–53]. Thus, when viewed as a whole, these data suggest peak musculotendinous load does not occur until midstance. This calls into question whether excessive amounts or velocities of pronation, both movements that occur early during stance, are really responsible for these two injuries.

Additional support for the explanation that prolonged pronation plays a role in the development or AT or MTSS can be drawn from the Figures showing average percent elongations across the 0-20%, 20-60%, and 60-100% time periods of stance (Figures 5.5, 5.5, 5.9, and 5.10). Focusing on the 20 – 60% stance and 60 - 100% stance periods, as previously mentioned there were 18 comparisons where the AT or MTSS group demonstrated greater average musculotendinous percent elongations than their respective controls. However, in 12 of those 18 comparisons the PP group also demonstrated significantly greater average musculotendinous percent elongations than the control groups while not being statistically different from the AT or MTSS groups. In many of the other comparisons, while not statistically significant, the mean values for the PP group tended to fall between those of either the AT or MTSS group, and their respective controls. This is especially interesting when considered along with the fact that the prolonged pronators were all healthy at the time of testing. In this light, perhaps runners who demonstrate prolonged pronation, and have the accompanying higher levels of average musculotendinous percent elongation during mid and late stance, are more susceptible to developing soft tissue overuse injuries such as AT or MTSS. While much additional work is required to validate this hypothesis, it is an intriguing hypothesis as it might allow for *a* priori screening and identification of individuals especially susceptible to these two common running overuse injuries.

While few significant main effects of group were observed, there were numerous significant main effects of foot strike (Tables 5.2 and 5.4). Running barefoot [116,128,165,166] or switching from a RFS to a M/FFS [167–169] is currently being promoted as one way to reduce running related injuries, primarily due to the reduced first peak and loading rates in the vertical ground reaction force typically observed when running with a MFS [18,128]. While not the primary focus, the results of the current study provide additional insights into the mechanics involved with different foot strike patterns. While the exact muscles involved varied depending on whether the AT or MTSS groups were analyzed, in general, subjects who utilized a M/FFS demonstrated greater musculotendinous percent elongation rates, and reached peak percent elongation rate earlier in stance than subjects who utilized a RFS. Overall, this suggests higher eccentric loading of the plantar flexor musculature early in stance when using a M/FFS then when using a RFS. Thus, individuals considering switching foot strike patterns as a way to possible prevent injury should be aware that they may simply be trading higher loading rates in the ground reaction force for higher loading within the soft tissue structures.

In summary, this study revealed there are distinct differences in musculotendinous kinematics between individuals currently symptomatic with AT or MTSS, healthy matched control subjects, and runners who are currently healthy but demonstrate prolonged pronation. For all subjects musculotendinous loads were highest during mid-stance but the greatest number of differences between injured and control subjects occurred when examining average musculotendinous percent elongation between 60 – 100% of stance. Thus, future studies, both on biomechanical factors causing AT or MTSS, and on intervention and rehabilitation strategies, should consider kinematics and kinetic factors during late stance as well as the traditionally emphasized early stance period.

CHAPTER VI

DISCUSSION AND CONCLUSION

Main Findings

Differences Based on Shoe Condition or Foot Strike Patterns

Chapter II focused on a method for using force plates to examine the center of pressure (COP) trajectories relative to the anatomic structures of the foot, and applying this method for comparing COP trajectories between barefoot and shod running. The results from this chapter suggest that the use of the force plate for examining anatomically relevant COP trajectories both results in measures similar to those obtained using pressure sensor techniques and is capable of detecting subtle differences between experimental conditions. The main differences between shod and barefoot running observed in this study included a more medially located COP throughout much of stance phase when running barefoot, a reduced COP anterior-posterior excursion when running barefoot, a higher peak external dorsiflexion moment when running barefoot, and no differences in the external eversion moment during the early portion of stance phase between shod and barefoot conditions.

While not directly related to the concept of prolonged pronation, these findings have implications for the idea that one type of shoe condition (i.e. shod or barefoot) or foot strike pattern (i.e. RFS or a M/FFS) is "better" than another for preventing common running injuries. The more medial location of the COP observed while running barefoot raises questions about soft tissue injury risk, as similar medial COP positions and associated higher pressures under the medial aspects of the foot, have been previously reported in several prospective studies as identifying individuals who subsequently developed exercise related lower leg pain [94,95]. Similarly, the reduced COP anterior-posterior translation has been reported as a risk factor for individuals who subsequently developed Achilles tendinopathy [90]. The higher peak external dorsiflexion moment observed when running barefoot would require an individual to generate a greater internal plantarflexor moment to compensate, a requirement which my increase load on the Achilles tendon.

All the subjects in the Chapter II study switched from a RFS while running shod to a M/FFS while running barefoot. Thus, we cannot be sure whether the observed results were due to the change in shoe condition or to the change in foot strike pattern. However, the results from Chapter II mesh nicely with those reported in Chapter V, where subjects ran in shoes using their normal foot strike pattern. While few between group differences in musculotendinous kinematics were observed, there were numerous differences between subjects who used naturally used a RFS and those who naturally used a M/FFS. While the exact muscles involved varied depending on whether the AT or MTSS groups were analyzed, in general, subjects who utilized a M/FFS demonstrated greater musculotendinous percent elongation rates, and reached peak percent elongation rate earlier in stance than subjects who utilized a RFS, suggesting higher eccentric loading of the plantar flexor musculature early in stance when using a M/FFS.

Thus, when considered as a whole, the results from Chapter II and Chapter V do not support the idea that one shoe type (shod or barefoot) or foot strike pattern (RFS or M/FFS) is necessarily “better” than another. The only definitive conclusion possible from these studies is that the different shoe conditions or foot strike patterns results in different mechanical stimulus to the body. What that means for injury prevention remains to be seen.

Biomechanical Markers of Prolonged Pronation

Based on the results of Chapter II, the force plate derived COP trajectories were used in Chapter III, in combination with full three dimensional motion analysis, as a tool for identifying biomechanical markers of prolonged pronation in healthy runners. In this chapter two clinicians identified two groups of individuals, one who, based on their expert clinical opinions, demonstrated prolonged pronation, and another who did not demonstrate prolonged pronation. Biomechanical differences between the two groups were then examined. There were no differences in either the amounts or velocities of pronation between the groups. However, four variables were significantly different between the groups. Compared to "normal" pronators, the prolonged pronators demonstrated longer periods of pronation, more eversion of the heel at heel off, higher standing tibia varus angles relative to the floor, and reduced static hip internal rotation range of motion.

Since the goal for this study was to identify biomechanical markers of prolonged pronation, it makes sense that the period of pronation was significantly different between the groups. Additionally, since the two clinicians classified individuals based on video showing the orientation of their calcaneus at heel off, it makes sense that the more everted position at heel off would be different between groups as well. However, to examine what other measures might help identify individuals with prolonged pronation any variable where differences between groups results in a p value $< .2$ was entered into a binary stepwise forward logistic regression. The results revealed that standing tibia varus angle, hip internal rotation range of motion, and static hip internal rotation range of motion were the variables which were capable of predicting group membership. That two of these variable are easily measureable in clinical settings suggests it may, in the future, be possible to develop a fairly simply screen to identify individuals who may demonstrate prolonged pronation while running without requiring a full three dimensional biomechanical analysis.

The results from the COP trajectories revealed the prolonged pronators had a more medial positioning of the COP trajectory during most of stance phase. This study also revealed the prolonged pronators did not display differences in the peak external ankle eversion moment early in stance, but did display differences in the rate at which the external ankle eversion moment rose, then fell, then rose again. Finally, the prolonged pronators also demonstrated greater external knee adduction moments throughout stance phase than the non-prolonged pronation groups.

As a whole, these findings have implications for the relationship between prolonged pronation and common overuse injuries, as many of the biomechanical markers of prolonged pronation have been previously associated with common running injuries. For instance, a more medial location of the COP trajectory during stance has been reported in both retrospective [95] and prospective studies [47] examining individuals who suffered from exercise related lower leg pain. The higher standing tibia varus angle suggests a more genu varus knee alignment. Both retrospective [13,31] and prospective studies [27,139] examining relationships between anatomic alignment and common running injuries have reported higher genu varus in injured compared to non-injured runners. Finally, the reduced static hip internal rotation has also been identified as a prognostic indicator for individuals at risk for developing medial tibial stress syndrome [33]. While these findings suggest prolonged pronation may be an important factor to consider in the development of common running injuries, the prolonged pronators in this study were actually healthy at the time. Thus, it is necessary to examine whether currently injured individuals also demonstrate these biomechanical markers of prolonged pronation.

Prolonged Pronation in Runners with Achilles Tendinopathy and Medial Tibial Stress Syndrome

Chapter IV examined individuals currently symptomatic with either Achilles tendinopathy or medial tibial stress syndrome to determine whether they demonstrated the biomechanical markers of prolonged pronation identified in Chapter III. Three setes of analyses were performed: once comparing the Achilles tendinopathy subjects to their respective controls; once comparing the medial tibial stress syndrome subjects to their respective controls; and once comparing combined injured and control groups. Results were the same regardless of which groups were compared.

There were no differences in the percent stance at which heel off occurred, the propulsive impulse, peak propulsive forces, or peak vertical ground reaction forces between any of the groups, suggesting the mechanics of their push off are similar. There were no differences in the amounts or velocities of pronation between injured and control subjects. However, the injured subjects demonstrated many of the biomechanical markers of prolonged pronation identified in Chapter III. Compared to controls, injured subjects had longer periods of pronation, more everted orientation of the heel at heel off, and higher standing tibia varus angles. A forward stepwise binary logistic regression suggested the period of pronation was able to predict group membership with a high degree of accuracy. Additionally, injured subjects displayed a more medially located position of their COP during stance.

The results of this study suggest that individuals currently symptomatic with Achilles tendinopathy or medial tibial stress syndrome do not display greater amounts or velocities of pronation compared to healthy controls. Instead, they demonstrate more prolonged pronation. Ways in which prolonged pronation could lead to the development of these two injuries were discussed in Chapter IV. However, one limitation to this study is that the injured subjects were already symptomatic with their injuries. Thus, we cannot be sure whether their movements actually lead to the injury or simply reflected adaptations to the injury. To being addressing this limitation, Chapter V utilized musculoskeletal modeling techniques to examine whether these movement patterns could be associated with muscular loading thought to lead to the development of the injuries.

Musculotendinous Kinematics in Injured, Healthy, and Runners with Prolonged Pronation

In the final study musculoskeletal modeling software was used to examined musculotendon kinematics in individuals currently symptomatic with Achilles tendinopathy or medial tibial stress syndrome, healthy matched control subjects, and healthy individuals previously identified as demonstrating prolonged pronation. Musculotendinous lengths, velocities, percent elongations, and percent elongation rates were examined. The only variable which was actually statistically significant between groups was the peak musculotendinous length and percent elongation in the soleus muscle, and only when comparing individuals with medial tibial stress syndrome to their controls. Numerous additional comparisons resulted in large between groups effect sizes, suggesting inter-individual variability was possible affecting the ability to detect significant differences in peak values.

However, when musculotendinous percent elongation was averaged across 0 – 20%, 20 – 60%, and 60 – 100% stance, more between group differences were observed, with the number of comparisons resulting in significant between group differences increasing with the later time periods. Assuming musculotendinous strain is involved in the development of these injuries, then observation that the greatest number of between group differences in percent elongation occurred late in stance suggests this might be the primary period on which to focus for injury risk assessment or intervention. Since prolonged pronation inherently implies actions occurring later in stance phase, these findings also support the theory that prolonged pronation, not necessarily excessive amounts or velocities of pronation, may be an important variable for future studies on these injuries.

Limitations to the Study

Several limitations in the current study should be noted. In Chapter II all the subjects were habitually shod runners who were participating in an acute bout of barefoot running. In this regard, it is unknown if the observed differences in COP trajectories would have remained had the subjects been given more time to adapt to BF running. Similarly, it is unknown whether these differences would still have been observed had habitually BF runners been used as subjects. Finally, it should be noted that subjects were included in this study only if they naturally transitioned from a RFS when SH to a

MFS when BF. While this transition in foot strike patterns is commonly observed when habitually SH runners are asked to run BF [127,128] other data from our laboratory suggests this might not always be the case. Thus, the differences in COP trajectories between SH and BF running reported in this study may be dependent on whether or not subjects change their foot pattern when switching from SH to BF running.

A limitation common throughout all the studies, but especially relevant to Chapters III and IV, is the foot model used in the studies and the accompanying assumptions involved in calculation foot motion. The movement of interest in these studies was foot pronation. As previously described, foot pronation is a complex motion, involving movement of multiple bones and occurring at multiple joints within the foot. Thus, without the use of invasive techniques such as intra-cortical bone pins [37,170–172] or dynamic imaging techniques such as bi-planar fluoroscopy [173,174] it is impossible to truly quantify all the motions taking place. While this difficulty can partially be alleviated by using multi-segmented foot models [175,176], even these models present challenges, especially for running studies. These include the following.

First, commonly used multisegement foot models divide the foot into anywhere from three to five segments. While this is undoubtedly better than simply using a single rigid segment to model the foot, one still describes the motion at each joint independently from the motion taking place at other joints. Similarly, motions are described in a single plane, or about a given rotational axis. While some authors have attempted to combine these three rotations by calculating the vector sum of the rotations about the three axes and calling this the “3D pronation” [95], this nomenclature has seen little other use.

A second difficulty with multi-segmented foot models, especially in running studies where subjects are wearing running shoes, is placing the markers on the foot in a way that they are visible to the motion capture cameras. To solve this issue some authors have used specialized running sandals [43,177,178]. However, given that these sandals have different midsole properties then most running shoes and lack standard running shoe features such as a heel counter, it is not certain whether they accurately would reflect kinematic and kinetic conditions encountered when running in shoes. Thus, in the current study each subject wore their own running shoes and holes were cut in the shoes to facilitate marker placement directly on the foot. While a similar method has been used in previous studies [42,134,136,179,180], it inherently limits the number of markers, and thus the number of foot segments, which can be used. Therefore, the foot model used in this study was a simplified version of the Leardini [176] model with only a rearfoot and a forefoot segment. The joint center for the joint between the rearfoot and forefoot segments was assumed to be midway between the navicular and 5th metatarsal markers. Thus, while the descriptions of rearfoot eversion as a measure of pronation are similar to what has been previously reported in the literature (see Table 3.5), it is unknown whether the other descriptors of foot kinematics such as forefoot abduction would also closely match previously reported data.

In addition to the foot model used, there were limitations with the methodologies used to actually calculate the joint kinematics. The current series of studies used the static trial to establish the anatomic coordinate systems for each segment and then calculated joint angles during stance phase using Cardan rotations. In this scenario the joint angles represent movement of the distal segment relative to the proximal segment.

While the coordinate systems and rotation sequences used were established according to the recommendations of the International Society of Biomechanics [97], and are commonly used in biomechanics research, they are not without limitation. For instance, the plantar flexion-dorsiflexion, inversion-eversion, and internal-external rotation for the ankle were modeled to occur at the ankle joint center, defined as the midpoint between the medial and lateral malleoli markers. From an orthopedic perspective this could be problematic for several reasons.

First, based on the bony geometry of the ankle mortise there really is not much internal or external rotation that can take place. Secondly, the eversion-inversion of the calcaneus takes place predominantly at the subtalar joint, not the ankle. Similar arguments could be made for modeling the motion of the midfoot joint. In this regard, perhaps the use of Cardan angles to model plantar flexion and dorsiflexion combined with approaches such as a helical axis for the subtalar joint [181] would yield more anatomically relevant joint motions. However, such approaches are currently not commonly seen in the running literature, and thus would reduce the ability to compare the results of the current series of studies to those published previously.

Another limitation that applies to the Chapters III and IV, is the choice of parameters to examine for quantifying prolonged pronation. The variables presented in these chapters either intuitively made sense (period of pronation), where based on the clinicians observations and classification of subjects (heel eversion at heel off), or were commonly reported variables in biomechanical studies on running injuries. A small subset of the variables was able to differentiate between non-prolonged and prolonged pronators. However, it is quite likely there could be other parameters which would have also differentiated between non-prolonged and prolonged pronators which were not examined in this study. For instance, several authors have discussed the implications of congruity between the actions of the knee and the actions of the subtalar joint during stance [104,105,182–184]. While pronation duration was not explicitly measured, implication in these studies is that if pronation is prolonged beyond midstance, when the knee started extending, then additional torsional stress would be placed on the soft tissue structures at the knee. The authors hypothesized this could be one reason for the knee being one of the most commonly sites for the occurrence of running injuries.

In addition to the actions of the subtalar and knee joints, coordination between the rearfoot and the shank [185–190] as well as between the rearfoot and forefoot [178,191–193], have been investigated. Since foot pronation involved movements of the bones comprising the shank, rearfoot, and forefoot segments, it seems likely that prolonged pronation could be identified through differences in these coupling patterns. Thus, while the variables examined in the current studies did demonstrate differences between non-prolonged and prolonged pronators, it is highly likely that there are additional biomechanical variable involving inter-joint coordination or coupling behaviors which could also be used to identify individuals with prolonged pronation.

There are other limitations to this dissertation that are only relevant for the study presented in Chapter V. The goal of this study was to examine musculotendinous kinematics in injured and healthy runners and runners with prolonged pronation. In this regard, the main parameters reported in this study were musculotendinous lengths and velocities as well as peak percent elongations and percent elongation rates. While these are informative, they all describe the entire musculotendinous unit. As discussed in

Chapter V, the musculotendinous unit consists of the actual muscle fascicles and the tendon component. Thus, the information presented in Chapter V would be more informative if the kinematics of these individual components could be examined. Additionally, it would have been beneficial to gain some insights into the actual amount of force being produced by each muscle.

Given recent advancements in the OpenSim software [107], these types of comparisons are currently possible. For instance, several studies have reported individual muscle fascicle as well as tendon parameters [151,156,158]. However, the main limitation in these studies is they required simplified foot models where the foot consists of a single rigid segment with the only degree of freedom being plantar and dorsiflexion at the ankle. Thus, while similar techniques could have been employed in the current study, it is questionable whether the results would have much meaning given the importance of all the other joints and degrees of freedom in foot pronation.

Other authors have eschewed the OpenSim [107] platform and estimated forces produced in the extrinsic foot muscles based on parameters such as their physiologic cross sectional area [21]. While this type of approach would yield the individual muscle forces, it loses the ability to differentiate between the actions within the muscle fascicle and the tendon components. Ultimately, is either further refinements of the foot models used in OpenSim [107], or a the development of hybrid modeling technique incorporating motion and length data from OpenSim [107] with external force calculations such as those as those by Hamill et al. [21] will be required to provide a more detailed analysis of muscle kinematics and kinetics while maintaining anatomically valid joint movements.

Finally, an additional limitation for the study presented in Chapter V was the high inter-individual variability observed in those results. This may be one reason for the lack of statistical differences, despite the large effect sizes, observed in this study. For many of the peak musculotendinous percent elongation values the standard deviations are close to or above 50% of the mean value. Large standard deviations were also observed for the percent stance at which peak musculotendinous percent elongation and percent elongation rate occurred. In this case this may be due to the fact that some subjects demonstrated peak values very early in stance while other demonstrated peak values late in stance. Thus, any actual differences between the groups may have been masked by the high inter-subject variability. It is possible this inter-subject variability could be accounted for by adding additional subjects or trials [194]. If this is the case then overall variability in these measures would be reduced and they may become statistically significant. However, it is also possible that the inter-subject variability could be resulting from subject's using different neuromusculoskeletal solutions to achieve the movement task [194]. Indeed, this appears highly likely given the redundancy within the musculoskeletal system. In such a case additional subjects would not yield clearer results and a single subject analysis approach might be warranted [194]. Further research is required to clarify exactly which may be the case.

Future Research

There are several future research studies that naturally follow from the results presented in this dissertation. First, as just discussed, it is likely there are additional biomechanical parameters involving joint coordination or coupling that could distinguish between non-prolonged and prolonged pronators. However, there are numerous techniques available for quantifying inter-joint coordination or joint coupling patterns. Thus, future studies should examine exactly which, if any, of these methods are also capable of identifying individuals with prolonged pronation. The natural follow up to this would be to also examine whether individuals with injuries such as Achilles tendinopathy or medial tibial stress syndrome also display these coupling patterns.

While the data presented in this dissertation suggests there may be a relationship between prolonged pronation and common running injuries, it cannot conclude that definitively. Therefore, a prospective study following individuals with prolonged to see if they actually develop these injuries is needed. Though the time constraints for this study prevented it from achieving that goal, several small steps were made in this direction. Figure 6.1 shows the center of pressure trajectories from two subjects who were evaluated both pre and post injury. Subject one was first evaluated when he was healthy, with no issues. He was subsequently seen about 6 months later after he had developed Achilles tendinopathy. As is shown in Figure 6.1A, from the pre injury evaluation to the post injury evaluation this subject's center of pressure trajectory shifted medially. While not evident from the COP trajectory alone, this individual also demonstrated several kinematic markers of prolonged pronation. His mean period of pronation during his first visit was 77.4 % of stance and his heel was everted, on average, 4.2° at heel off. At his second visit his period of pronation had increased to 85.6 % of stance phase while his heel position at heel off had not changed.

The second individual was first seen as an injured patient with bilateral medial tibial stress syndrome. As can be seen in the center of pressure trajectories in Figure

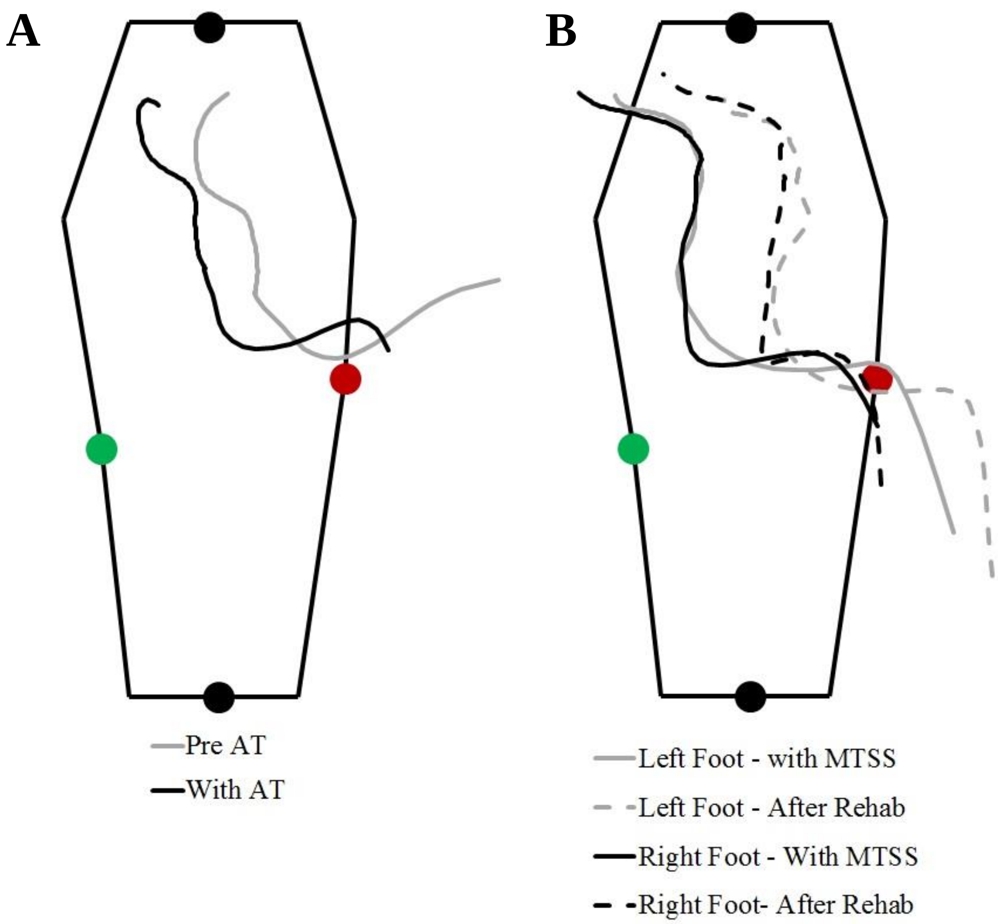

Figure 6.1. Examples from two subjects showing changes in the center of pressure trajectories between pre and post injury evaluations.

6.1B, the center of pressure on the first visit was located medial to the long axis of his foot. The period of pronation on the first visit was 74.9 % of stance for the left foot and 81.38 % of stance for the right foot. Similarly, the heel was everted 2.4° at heel off on the left foot and 3.2° at heel off on the right foot. However, after going through several weeks of rehabilitation his center of pressure trajectory shifted back lateral, on both feet, as can be seen in Figure 6.2B. The periods of pronation decreased to 66.2 % of stance on the left foot and 51.2 % of stance on the right foot while the eversion of the heel at heel off changed to 1° of inversion on the left foot and 3.2° of inversion on the right foot.

Thus, when considering both individuals, the presence of prolonged pronation perhaps signaled the impending injury in the first subject while it served as a quantitative marker of recovery in the second subject. Based on this small sample of pilot data it certainly would seem that a longitudinal study on the relationship between the period of pronation and common overuse running injuries is warranted.

However, suppose such a study is performed and its findings support the relationship between prolonged pronation and the development of common running injuries. The logical question stemming from such a finding would be "can prolonged pronation be easily addressed and what are the best methods for doing so?" Data such as that presented in Figure 6.2B suggests that periods of pronation certainly can be changed and it more a question of what are the best methods for doing so. For instance, the subject in Figure 6.2B was involved with a rehabilitation program consisting of general core and hip muscle strengthening as well as targeted strengthening and retraining of the tibialis posterior muscle. However, this is only one option. Equally effective might be the use of an orthotic intervention. However, should the orthotic focus on controlling rearfoot motion or forefoot motion? Perhaps a combination of hip strengthening and orthotics yields best results? Recently gait retraining has shown promise for changing biomechanical factors thought to lead to running injuries [195–197]. Perhaps a target gait retraining program, when combined with proper strengthening leads to the best clinical outcomes? These are just a few of the many follow up questions needing answering regarding the involvement of prolonged pronation and running injuries. Answering them may finally, after 40 years of research, help start making a dent in the occurrence and frequency of running related overuse injuries.

APPENDIX A

CONTENTS OF THE CLINICAL EVALUATION EXAM

Running Study Subject Questionnaire and Clinical Evaluation Form

Subject Code:________________ Date:________________

Age: _________

Year in college, if applicable: __________

Number of Years Running: ____________

Approximate Mileage Run per Week: _______________

Over the course of your running career, have you sustained any running related injuries? Y N

If Yes then please describe the nature of the injury, diagnosis by a physician, extent or duration of the injury, and treatment protocols you underwent to relieve symptoms:

____

____

____

____

____

____

Other Comments of History Information:

General Lower Body Alignment and Mobility Assessment

1. Angle of Gait
   a. Left: ____________
   b. Right: ___________

2. Leg Varus to Floor
   a. Left: ____________
   b. Right: ___________

3. Standing Arch Type:
   a. Left: ____________
   b. Right: ___________

4. Tibial Torsion?
   a. Left: ____________
   b. Right: ___________

5. Extremity Length (cm)
   a. Left: ____________
   b. Right: ___________

6. Ankle Dorsi Flexion

	extended	flexed
Left		
Right		

7. Ankle Plantar Flexion (flexed)
   a. Left: ____________
   b. Right: ___________

8. Prone Hip Rotation

	Internal	External
Left		
Right		

9. Hamstring Flexibility (from 0)
   a. Left: ___________
   b. Right: __________

10. Quadriceps Flexibility (from straight)
    a. Left: ____________
    b. Right: ___________

11. Gastroc Flexibility (@ STN)
    a. Left: ____________
    b. Right: ___________

12. Obers Test
    a. Left: ____________
    b. Right: ___________

13. General Foot Motion (Loose, Tight, Normal)
    a. Left: ____________
    b. Right: ___________

14. Subtalar Joint Inversion
    a. Left: ____________
    b. Right: ___________

15. Subtalar Joint Eversion
    a. Left: ____________
    b. Right: ___________

16. Forefoot Alignment (Neutral, Varus, Valgus)
    a. Left: ____________
    b. Right: ___________

17. 1st Ray Position (Pflexed, Dflexed, Neutral)
   a. Left: ___________
   b. Right: __________

18. 1st Ray Motion (Normal, Mod Restricted, Restricted)
   a. Left: ___________
   b. Right: __________

19. 1st MPJ Joint ROM (dorsiflexion)
   a. Left: ___________
   b. Right: __________

20. Heel Varus @ STN (Toe Position)
   a. Left: ___________
   b. Right: __________

Clinician Notes/Comments:

Measurement of Arch Height

(from Williams, McClay, Hammill, and Buchanan (2001). Lower Extremity Kinetic and Kinematic Differences in High and Low Arched Runners. *J. Applied Biomechanics.* Vol. 17, pp. 153-161)

1. Full Foot Length (cm): L __________ R___________
2. 50% Full foot length (cm): L __________ R___________
3. Truncated Foot Length ____________L _____________R

   (measured from most posterior point of calcaneus to medial joint space of first metatarsal phalangeal joint).

4. Height of Dorsum of foot @ 50% foot length: __________L ____________R
5. Arch Height Ratio: _________ L __________R

   (measurement 4 divided by measurement 5)

Height (cm)____________ Weight (Kg)_____________

APPENDIX B

INFORMED CONSENT FORM

You are invited to participate in a research study conducted by Drs. Li-Shan Chou, Louis Osternig, Stan James, and graduate student James Becker regarding the role of foot pronation in running injuries. We hope to understand how the duration of foot pronation can be quantified from both clinical and biomechanical perspectives, how it may be different in injured and uninjured runners, and how it may affect muscle forces in the lower limb while running. You are being invited to participate because you are either a currently healthy runner or because you are currently an injured runner with either medial tibial stress syndrome or an Achilles tendon injury**.**

If you decide to participate, you will be tested in the Motion Analysis Laboratory at the University of Oregon. This study is longitudinal in nature, meaning we will follow up and retest you at regular intervals throughout the year. It is expected that the follow up visits will be conducted approximately every three months; however there is some leeway in this time frame depending on your individual schedule and needs. We anticipate recruiting a total of 150 subjects for this study, 20 currently symptomatic with medial tibial stress syndrome or Achilles tendinopathy, and 130 healthy subjects.

TESTING PROCEDURES: The assessments in the Motion Analysis Lab will include both clinical and biomechanical evaluations. The clinical evaluation will include measures of your body alignment, joint range of motion, and muscle strength. For the running gait analysis reflective markers will be placed on selected bony landmarks to record the motion of each individual body segment. You will run laps around the laboratory space and your body movement (indicated by motion of reflective makers) during running will be recorded by our cameras for further analysis. We will also record your running with traditional video cameras and, with your permission, may take photographs of the marker set up placed on your body. Your will run both while wearing running shoes and barefoot. When you run with shoes on we will cut holes in the shoe to allow us to place markers directly on your foot. Therefore you will be required to provide an old pair of running shoes you do not mind having cut up. It is expected that you will run approximately 30 short laps around the laboratory under each condition, with each lap being approximately 25 meters. You will also be asked to complete short bouts of treadmill running, also with and without shoe. Finally, while running on a treadmill we will measure the pressure distribution under your foot using specialized insoles which we will place inside your shoe.

You will wear normal running shoes for these procedures. You will be asked to wear a pair of paper physical therapy shorts and sleeveless shirt (tank top) or equivalent clothing of your choice during testing to allow the cameras to clearly see the markers. It is expected each testing session will require approximately 2.5 hours of your time.

COMPENSATION: You will be compensated $20 for each visit to the laboratory. You should understand that your old shoes will no longer be usable after your participation in the study.

RISKS AND DISCOMFORTS: We expect that there will be no more risk for you during these tests than there normally is for you when outside of the laboratory. However, running in the laboratory is different than running outside. You will be asked to speed up then slow down over a 25 meter distance. Running laps in the laboratory will require negotiating tight corners. We will do our best to arrange the lab equipment and furniture to minimize any discomforts and provide as much room as possible. If you are not comfortable you may stop the trials at any time. Additionally, running on a treadmill is also not the same as running outside, however you may stop the treadmill at any time if you feel uncomfortable. You may feel fatigue during or after the testing. Our staff member will check with you frequently and provide any required assistance. You will be given frequent breaks as requested. Cutting the holes in your running shoes will require the removal of the inner lining so there is the possibility of rubbing or discomfort on your feet. We will do our best to reduce these effects, and should they still be present you may request additional modifications or stop the trials at any time. There is also the possibility of discomfort involved in removing adhesive tape (used for marker placement) from skin at the end of the experiment.

ADDITIONAL INFORMATION: Any information that is obtained in connection with this study and that can be identified with you will remain confidential and will not be shared without your permission. Subject identities will be kept confidential by coding the data as to study, subject pseudonyms, and collection date. The code list will be kept separate and secure from the actual data files. Your participation is completely voluntary. Your decision whether or not to participate will not affect your relationship with the Department of Human Physiology or University of Oregon. You do not waive any liability rights for personal injury by signing this form. In spite of all precautions, you might develop medical complications from participating in this study. If such complications arise, the researchers will assist you in obtaining appropriate medical treatment. In addition, if you are physically injured because of the project, you and your insurance company will have to pay your doctor bills. If you are a University of Oregon student or employee and are covered by a University of Oregon medical plan, that plan might have terms that apply to your injury. If you have any questions about your rights as a research subject, you can contact Research Compliance Services, 5237 University of Oregon, Eugene, OR 97403, (541) 346- 2510. This office oversees the review of the research to protect your rights and is not involved with this study.

If you decide to participate, you are free to withdraw your consent and discontinue participation at any time without penalty. This includes discontinuing your participation anytime during the initial visit or not returning for follow up visits. If you choose not to return for follow up visits the researchers may discontinue your participation in the study. Additionally, the researchers may discontinue your participation in this study if you are not able to provide an old pair of shoes, or are not capable of running the amount required to complete the testing, either on the treadmill or overground.

If you have any questions, please feel free to contact Dr. Li-Shan Chou, (541) 346-3391, Department of Human Physiology, 112C Esslinger Hall, University of Oregon, Eugene OR, 97403-1240. You will be given a copy of this form to keep. Your signature indicates that you have read and understand the information provided above, that you willingly agree to participate, that you may withdraw your consent at any time and discontinue participation without penalty, that you will receive a copy of this form, and that you are not waiving any legal claims, rights or remedies.

Name:____

Signature:____

Date: __________________________________

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