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DOES FOOTFALL PATTERN IN FOREFOOT RUNNERS CHANGE OVER A

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A Thesis Presented

by

CARL M. JEWELL

Submitted to the Graduate School of the
University of Massachusetts Amherst in partial fulfillment
of the requirements for the degree of

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Department of Kinesiology

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ABSTRACT

DOES FOOTFALL PATTERN IN FOREFOOT RUNNERS CHANGE OVER A
PROLONGED RUN?

MAY 2014

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There has been much debate on the benefits of a forefoot versus rearfoot strike pattern in distance running in terms of performance and injury prevalence. Shock attenuation occurs more prominently in soft tissues at impact in forefoot runners compared to the passive skeletal loading in rearfoot runners. Recent studies indicate that a forefoot strike pattern may not be maintainable over long distance efforts. Therefore, this study tested the hypothesis that habitual forefoot runners could not maintain their strike pattern throughout a prolonged, intensive run.

Fourteen forefoot runners ran to voluntary exhaustion on an instrumented force treadmill (average run duration: 15.4 ± 2.2 minutes). Kinematic and kinetic data were sampled each minute at 200Hz and 1000Hz, respectively. Ankle plantar-flexor torque was measured during pre- and post-run isometric contractions, during which electromyographic activity was measured in the soleus, lateral, and medial gastrocnemius.

Loading rate (49.95 ± 14.83 to 57.40 ± 22.53 BW*s^{-1}, p=0.0311) and impact peak (1.35 ± 0.43 to 1.50 ± 0.51, p=0.0207) increased significantly throughout the run. Both peak knee flexion ($-33.93\pm3.67°$ to $-36.21\pm3.48°$, p=<0.0000) and sagittal ankle angle at touchdown ($-11.83\pm5.33°$ to $-9.33\pm6.29°$, p =0.0202) increased significantly. Ankle

torque decreased significantly from pre- to post-run (120.57±33.57 to 110.76±32.91 Nm, p = 0.0154). This was accompanied by a decrease in medial and lateral gastrocnemius integrated electromyographic activity (iEMG) (p=0.0387 and 0.0186, respectively).

The results indicated that there were significant changes in landing mechanics in the habitual forefoot runners with increased levels of exertion, as they shifted towards strike patterns more similar to rearfoot runners throughout the run. These changes are in line with metabolic findings of other studies. There is increased eccentric loading of the ankle plantar-flexor muscles at touchdown in forefoot runners that may contribute to a decreased torque output by the end of the run. The decline in iEMG may indicate altered central drive of the system and a decline in the impact attenuation ability of the triceps surae, leading to the changes exhibited up the kinematic chain. These findings suggest that while forefoot strike patterns are good for speed, the onset of fatigue may affect the ability to maintain this pattern during a prolonged, intensive effort.

TABLE OF CONTENTS

APPENDICES

LIST OF TABLES

LIST OF FIGURES

CHAPTER 1

INTRODUCTION

1.1 Background

Running is one of the most common forms of exercise due to its ease of access, low cost, and beneficial health effects. In the United States alone, there are more than thirty million active runners for competition and recreation (Novacheck, 1998). This population of runners exhibits one of three distinct footfall patterns: heel strike (RF), midfoot strike (MF), and forefoot strike (FF) (Hasegawa, Yamauchi, & Kraemer, 2007; Novacheck, 1998). Over the course of a stride cycle, RF runners first make contact with the ground with their heel or the posterior third of the foot before transitioning to midfoot stance and the propulsive toe-off (Hasegawa et al., 2007). MF runners make initial ground contact simultaneously with the ball and rear of the foot, while FF runners make ground contact solely on the front half of the foot, with little to no heel contact later in the stance phase prior to toe-off (Hasegawa et al., 2007). Previous literature has indicated that the RF group is much more prominent in the running population, with approximately 75-80 percent of all runners falling in this group (Hasegawa et al., 2007; Kerr, Beauchamp, Fisher, & Neil, 1983). The remaining 20 percent consists mostly of MF runners, with sources reporting as few as 1.4 percent as true FF runners that make no heel contact with the ground (Hasegawa et al., 2007).

In previous literature, it has not been proven that FF running is a more economical style in terms of metabolic energy expenditure (Ardigo, Lafortuna, Minetti, Mognoni, & Saibene, 1995; Gruber, 2012). However, despite this lack of evidence, it has been shown that with increased running velocity, there is a trend to shift towards a more pronounced

FF running style (Hasegawa et al., 2007; Kerr et al., 1983; Novacheck, 1998; Payne, 1983). This has been attributed to the fact that FF runners tend to exhibit shorter ground contact times and shorter acceleration periods, as well as a toe-off that begins earlier within the stance phase; in some sprinters, this occurs as early as 25 percent through stance (Hasegawa et al., 2007; Hayes & Caplan, 2012; Novacheck, 1998). FF runners also exhibit decreased peak dorsiflexion and increased peak plantarflexion, which may allow for a greater amount of force production at toe-off (Novacheck, 1998). With shorter contact times and increased plantar flexion angles, a greater amount of energy must be absorbed at the ankle (Hasegawa et al., 2007). In FF running, the ankle takes on an additional role than compared to RF running; instead of being primarily an energy producer, it assumes the additional task of absorbing energy at touchdown (Dickinson, Cook, & Leinhardt, 1985). While greater shock attenuation at the ankle may appear to be a beneficial aspect of FF running, it places greater stresses on the soft tissue of the lower extremity.

Throughout the gait cycle of running, the peak forces do not occur at the time of contact, but at the initiation and during the propagation of toe-off (Novacheck, 1998; Winter, 1983). During RF running, the initial vertical ground reaction force loading rate is higher than that of FF running, contributing to a greater lower extremity shock wave compared to FF running. (Laughton, Davis, & Hamill, 2003; Pratt, 1989). However, in FF running, much of the force at impact is placed on the soft tissues surrounding the ankle, namely the tibialis posterior, gastrocnemius, and the soleus, causing greater Achilles' tendon loading compared to RF running (Laughton et al., 2003; Novacheck, 1998). The Achilles' tendon behaves as a spring during running, stretching during the

first portion of stance, storing elastic energy, and then later recoiling to return roughly 90 percent of the prior absorbed energy at the propulsive toe-off (Novacheck, 1998). In FF running, significantly higher eccentric loading on the Achilles' tendon and triceps surae; forces in the Achilles' tendon have been estimated between 6-8 body weights (Laughton et al., 2003; Pratt, 1989; Williams, McClay, & Manal, 2000). Due to these extreme, rapid loads on these soft tissue structures in the posterior of the lower limb, the force attenuation capacity of the muscle can be decreased, potentially altering gait under excessive fatigue (Dickinson et al., 1985; Winter, 1983). Loss of coordination and joint motion can be caused by localized muscle fatigue, causing increased and altered loading rates and internal loading that can cause changes in stride mechanics (Hayes, Bowen, & Davies, 2004).

In addition to localized muscle fatigue, the repeated stretch-shorten cycling of the elastic elements in the triceps surae and Achilles' tendon can be affected by exaggerated eccentric loading (Gollhofer, Komi, Fujitsuka, & Miyashita, 2008; Komi, 2000). Because of the high levels of muscular force produced eccentrically in these posterior leg muscles, previous literature suggests that the elastic elements of these muscles, as well as neuronal control, is altered with fatigue (Gollhofer et al., 2008; Komi, 2000). This means that in FF running, where the storage and recoil of elastic energy in the triceps surae and Achilles' tendon is of greater importance, the energy return is reduced after many repeated stretch-shorten cycles. Due to this reduced energy return capability, the body adjusts leg stiffness as well as overall limb coordination via activation couplings and segment positioning at foot strike (Bosco, Komi, & Ito, 1981; Nicol, Komi, & Marconnet, 1991). As the stiffness of the leg is reduced during this eccentric loading, the force

attenuation capacity of the musculo-tendon complex is also reduced (Dalleau, Belli, Bourdin, & Lacour, 1998). Over the course of a run with many thousand repeated footfalls, this fatiguing effect could be greatly magnified, giving a high potential for compensation of the body via altered kinematics and foot strike pattern changes.

Fatigue has been defined as any reduction in the maximum force generating capacity of a muscle regardless of the forces required in any given situation (Madigan & Pidcoe, 2003). Several studies have already examined the effect of specific muscle and functional lower limb fatigue on performance, as well as the effect on overall foot strike pattern (Hasegawa et al., 2007; Hayes & Caplan, 2012; Kerr et al., 1983; Larson, Higgins, Kaminski, Decker, Preble, Lyons, McIntyre, & Normile, 2011; Laughton et al., 2003; Madigan & Pidcoe, 2003; Mizrahi, Verbitsky, & Isakov, 2000a; Mizrahi, Verbitsky, Isakov, & Daily, 2000b; Mizrahi, Voloshin, Russek, Verbitski, & Isakov, 1997). In one study, single leg squats and stands were repeated until exhaustion while kinetic and kinematic data was collected (Madigan & Pidcoe, 2003). After exhaustion, vertical ground reaction peak forces decreased 12 percent, and overall flexion at the knee and dorsiflexion at the ankle increased (Madigan & Pidcoe, 2003). These changes were accompanied by significantly different hip and ankle impulse values, indicating changes occurred throughout the entire lower limb chain to compensate for increased levels of fatigue (Madigan & Pidcoe, 2003). Another study focusing on local muscular fatigue studied runners during an exhaustive run at VO_2max, and their findings showed that local muscular endurance of hip extensors and knee flexors are important for maintaining running form in a fatigued state as well as maintaining a stable stride length (Hayes et al., 2004). With increasing fatigue, stride length has been shown to decrease while frequency

remains relatively constant (Elliot & Ackland, 1981). Other authors have found that stride length is variable with fatigue, with some runners shortening their stride while others lengthening, which was closely correlated with hip extensor local muscular endurance (Hayes et al., 2004). Decreased stride length has been accompanied by increased contact times, greater knee extension, and decreased heel acceleration at foot strike (Elliot & Ackland, 1981; Hayes & Caplan, 2012).

Previous studies have been performed on site at recreational and competitive distance races to investigate how foot strike patterns change over the course of a race (Hasegawa et al., 2007; Hayes & Caplan, 2012; Larson et al., 2011). In the longer distance races, it has been shown that the number of FF strikes decreases the farther into a race they are observed (Hasegawa et al., 2007; Larson et al., 2011).

1.2 Statement of the Problem

Many studies have investigated the effects of fatigue during an exhaustive or prolonged run (Abt, Sell, Chu, Lovalekar, Burdett, & Lephart, 2011; Bates, Osternig, & James, 1977; Cheung & Ng, 2010; Derrick, Dereu, & McLean, 2002; Elliot & Ackland, 1981; Friesenbichler, Stirling, Federolf, & Nigg, 2011; Hasegawa et al., 2007; Hayes et al., 2004; Hayes & Caplan, 2012; Larson et al., 2011; Lepers, Pousson, Maffiuletti, Martin, & Van Hoecke, 2000; Mizrahi et al., 2000a; Mizrahi et al., 2000b; Mizrahi et al., 1997; Nicol et al., 1991). Several different protocols have been used to study and induce fatigue, ranging from a 400-meter maximum effort to marathon-length runs (Bates et al., 1977; Larson et al., 2011). Nearly every study shows some form of kinematic or kinetic adaptation to an exhaustive or prolonged run, indicating that there are changes in

muscular function on mechanical and neuronal levels that influence this. Despite all the previous research performed, there appears to be a lack of empirical investigation of prolonged runs in populations consisting entirely of FF runners.

For this study, it seems logical to take the next step in comparing RF and FF running. In the pre-existing literature, there appears to be a trend of fatigue causing runners to switch to a more RF-based running style. As the triceps surae in FF runners is subjected to repeated stresses of many footfalls, there is evidence that the high levels of eccentric loading can cause a reduction in the shock attenuation capability of the lower limb (Williams et al., 2000). In order to account for these fatigued muscles and decreasing leg stiffness, it can be expected for these runners to begin to make contact with the ground with more of a RF style, using greater knee flexion in early stance to attenuate the force of landing.

The purpose of this study is to examine kinetic, 3-dimensional kinematic, and electromyographic (EMG) data to discern how FF runners alter their foot strike patterns while performing a prolonged run on a treadmill. The participants will complete a run of one half hour at an intensity chosen by self-reported previous competition and training data. A run of this length and intensity is expected to be appropriate to induce local muscular fatigue of the triceps surae, which has been previously shown to alter overall running mechanics.

1.3 Hypotheses

It is hypothesized that the participants will gradually show a decreased FF running style and begin to shift towards a RF pattern. Three distinct hypotheses have been formulated to test if the shift occurs.

Hypothesis 1: There will be significant kinematic differences between each collection period of the prolonged run in the peak values and excursions of the sagittal ankle angle, sagittal knee angle, forefoot-rearfoot inversion-eversion angles, and tibial internal-external rotations.

Hypothesis 2: The vertical ground reaction force curve will begin to exhibit a steeper loading rate as well as a more pronounced impact peak in the first 30 percent of stance. Vertical and leg stiffness will decrease throughout the prolonged run.

Hypothesis 3: There will be a leftward shift in the frequency domain of the EMG signal in the triceps surae between pre- and post- exercise isometric contractions, indicating lower frequency activity in the muscles commonly seen with the onset of local muscular fatigue. There will also be a decrease in the integrated EMG in the pre- and post-exercise collections, accompanied by a decline in overall force production.

1.4 Significance of the Study

This study will add empirical evidence and depth to the ongoing, heated debate about the benefits of a FF running style when compared to a RF style. No studies in the past have looked specifically about the effects of fatigue on landing mechanics of FF runners. By collecting and analyzing this data, it is expected that some distinguishable differences in gait during a prolonged run will be observed. These findings could further

be used to provide more insight into injury onset and prevention, as well as to increase performance in all foot strike types.

1.5 Summary

There are more than 30 million people in the United States alone who run for exercise. Over 80 percent of these runners first make ground contact with the posterior third of their foot, a heel strike, at the beginning of stance phase (Hasegawa et al., 2007). The other remaining runners are made up of midfoot strikers, who land flat on their feet, and forefoot strikers, who land on the ball of the foot or anterior half with no heel touch through stance (Hasegawa et al., 2007). Previously, heel striking was touted as the most economical stride pattern for distance running, but recently, a trend towards recommending forefoot running has occurred. There has been little evidence on forefoot running being a more economical pattern (Ardigo et al., 1995; Gruber, 2012); however, runners tend to transition to a more pronounced forefoot running style at higher speeds (Hasegawa et al., 2007; Kerr et al., 1983; Novacheck, 1998; Payne, 1983). The transition from a rearfoot pattern to a forefoot one causes the ankle to take on an additional role as both a shock absorber and an energy generator, while only the latter primarily occurs in rearfoot running (Dickinson et al., 1985).

The added role of the ankle as an energy absorber causes the triceps surae and other posterior plantar flexors of the ankle to act eccentrically in early stance, which have been associated with high forces experienced in the Achilles' tendon (Laughton et al., 2003; Pratt, 1989; Williams et al., 2000). These high eccentric loads and repeated submaximal stretch-shorten cycling of the elastic elements of these muscles has been

shown to alter muscle force producing capability and shock attenuation, the effect increasing with an increase in fatigue (Dickinson et al., 1985; Gollhofer et al., 2008; Komi, Hyvärinen, Gollhofer, & Mero, 1986; Winter, 1983). Previous studies conducted in competitive race environments have observed a decrease in the amount of forefoot runners present at each successive stage in long-distance races (Hasegawa et al., 2007; Kerr et al., 1983; Larson et al., 2011). This study will involve the collection of kinematic, kinetic, and electromyographic data to determine whether or not forefoot runners will alter their strike patterns over the course of an intense, prolonged treadmill run to more closely resemble characteristics of a heel strike.

CHAPTER 2

REVIEW OF LITERATURE

2.1 Introduction

In running, much emphasis has been placed on determining the most efficient and injury preventative form. This investigation began by analyzing how the majority of people run, and continued with several comparative studies. One of the main focuses recently has been on foot strike pattern, as this is the only point of contact and link between the runner and the ground. The main argument has revolved around which foot strike pattern is more desirable in terms of injury prevention, running economy, and optimal performance.

The differences between FF and RF strike patterns have been investigated from kinetic and kinematic perspectives to the point where each style is well understood. However, there has been little comparative research on the effects of each of these patterns over long, intense bouts of exercise. This review of literature will be aimed to first define each foot strike pattern and then compare the main differences between the two. It will also delve into the literature of fatigue-inducing protocols, examining the different methods of inducing fatigue and the effects of localized muscle fatigue on muscle activity and performance in running, specifically.

2.2 Strike pattern types

A foot strike occurs whenever the foot makes contact with the ground, marking the beginning of the stance phase in the gait cycle. Historically, most of the literature has been conscious of two types of foot strike patterns: RF and FF. However, in reality, foot

strike patterns exist on a continuum encompassing landing points across the entire length of the foot (Cavanagh & Lafortune, 1980). Currently, there exist three groups generally referred to as RF, midfoot strike (MF) and FF (Hasegawa et al., 2007). The RF pattern initiates contact with the ground in the posterior third of the foot (Hasegawa et al., 2007). After this initial contact, the foot transitions to become completely flat on the ground during midstance (Hasegawa et al., 2007). The heel is then lifted off the ground and the propulsive toe-off phase occurs in the last portion of stance (Hasegawa et al., 2007). In the MF pattern, the foot lands flat on the ground, or lands one the front third of the foot and allows heel touch to occur, before entering the propulsive toe-off (Hasegawa et al., 2007). In forefoot running, no heel contact occurs (Hasegawa et al., 2007). Each of the styles is depicted in Fig. 2.1 below.

Figure 2.1. Strike pattern definitions. The top two panels depict a RF strike, the middle panels depict a MF strike, and the bottom panels depict a FF strike. (Hasegawa et al., 2007)

The occurrence of each foot strike pattern in the general population consist of 75-80 percent RF runners, with the remaining percentage filled mostly by MF runners (Hasegawa et al., 2007; Kerr et al., 1983). The actual percentage of pure FF runners, who have no heel contact with the ground throughout stance phase, has been reported to be as low as 1.4 percent in an elite level distance race (Hasegawa et al., 2007). However, the percentage of RF runners in a more recreational marathon even has been shown to be significantly higher, upwards of 88 percent (Larson et al., 2011).

2.3 Running speed

As was briefly mentioned in the previous section, there exists a trend in which the overall ratio of RF runners to other strike patterns is lower with increased running velocity. Indeed, most people walk with a RF pattern, and in a previous study, at speeds below 5 ms^{-1}, the majority of runners studied were shown to possess a RF pattern (Keller, Weisberger, Ray, Hasan, Shiavi, & Spengler, 1996). However, above 3.0 ms^{-1}, subjects began to show a marked decrease in RF patterns, with 86 percent of subjects utilizing a MF or FF running style above 6.0 ms^{-1} (Keller et al., 1996). However, despite the strong trend, speed was not the only determining factor in strike pattern, as one subject maintained a RF gait at speeds above 8.0 ms^{-1}. With an increase in speed, and transitioning from a walk, to a run, to a sprint, it appears that subjects transition to a more FF or MF biased strike pattern (Keller et al., 1996; Novacheck, 1998). Part of the proposed reason for this transition is an increase in pelvic tilt and plantar flexion at higher speeds to maximize propulsion (Novacheck, 1998). A consequence of these increased speeds are increased peak force magnitudes and loading rates (Keller et al., 1996). Thus,

as the foot strike pattern shifts to more FF at higher speeds, the loads and stressors that

the body is subjected to will also be increasing due to the increased velocity. In turn, this

may cause certain active muscle groups, such as the triceps surae, to fatigue more rapidly.

Although speed on its own is not an accurate predictor for foot strike pattern, the

percentage of FF and MF runners relative to RF has been shown to increase in distance

races with decreasing finish time (Hasegawa et al., 2007). In the elite-level half

marathon race studied by Hasegawa et al., RF runners made up 74.9 percent of all 415

runners; however, if only the top 50 finishers of the race were looked at, this percentage

dropped to 62 percent (Hasegawa et al., 2007). Another large scale study indicated that

while the majority of runners again fell into the RF group, at faster speeds, there was a

greater percentage of MF and FF runners (Kerr et al., 1983). This relationship was

presented well by Kerr et al., and is presented in Fig. 2.2 below. The causal relationship,

if any, between foot strike pattern and finish time is unclear. There has merely been

speculation of whether FF runners are able to run faster as a result of their strike pattern,

or that they are simply able to maintain a FF strike pattern throughout a race due to their

enhanced fitness.
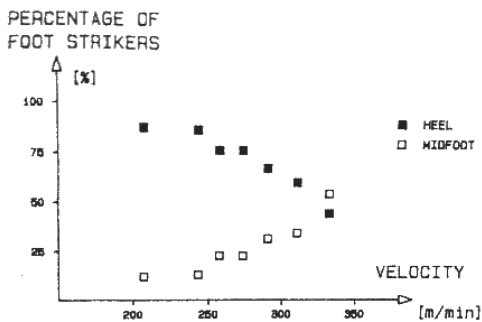
Figure 2.2. Percentage of each foot strike pattern with increasing velocity (Kerr et al., 1983). With increasing velocity, a greater percentage of runners switch from a RF pattern to a MF. The X-axis is shown with increasing velocity in m/min, while the Y-axis is given as a percentage of Foot Strikers.

13

Along with the trend between running velocity and strike pattern exists another

related relationship. With increased velocity there is a decrease in ground contact time,

defined as the amount of time the foot is touching the ground (Hasegawa et al., 2007;

Hayes & Caplan, 2012; Novacheck, 1998). In a study conducted on 181 male and female

middle distance runners, it was shown that in 800 and 1500 meter race time had a

significant relationship with ground contact time, with a correlation of -.586 and a p-

value of <0.001 (Hayes & Caplan, 2012). This study also showed that elite 800 and 1500

meter athletes are a much more homogenous group in terms of foot strike pattern than

longer distance runners, with RF, MF, and FF runners making up 27, 42, and 31 percent

of the population, respectively (Hayes & Caplan, 2012). This coincides with conclusions

made from previous literature regarding middle distance specialists exhibiting less RF

strike pattern characteristics (Payne, 1983).

<u>2.4 Running economy</u>

In previous literature, it has long been proposed that a RF running style is the

most economical style for longer distances in regards to the metabolic cost of transport

(Payne, 1983; Williams & Cavanagh, 1987). However, more recently, there has been a

rise in favor of FF and MF running styles. This is a direct result of correlations made

between foot strike pattern and race finish, leading many to believe that to run fast, a non-

RF style must be adopted (Hasegawa et al., 2007; Hayes & Caplan, 2012; Kerr et al.,

1983). Several of these authors have used this observation to make assumptions about

running economy, believing FF and MF running styles to be more economical than RF

running (Hasegawa et al., 2007). Further comparisons between foot strike patterns in

barefoot and shod conditions have led others to the same conclusion (Lieberman,

Venkadesan, Werbel, Daoud, D'Andrea, Davis, Mang'eni, & Pitsiladis, 2010). It has

been suggested that FF running is more economical because it allows for greater storage

and utilization of energy in elastic elements of the Achilles' tendon and triceps surae,

which will be discussed in a later section.

Despite the evidence from race results, there have been no human studies that

show a difference in running economy between the foot strike patterns, as depicted in Fig.

2.3 (Ardigo et al., 1995; Perl, Daoud, & Lieberman, 2012). Recent work has indeed

shown that FF runners exhibited a slightly, though statistically significant, greater rate of

oxygen consumption and cost of transport running at their natural FF running pattern than

a converted RF strike pattern (Gruber, 2012). This data has limited biological

significance; however, it may be more relevant in highly trained athletes, for whom

switching to a RF pattern from a FF may be advantageous (Gruber, 2012). For the

majority of the running world, this research indicates that there is no universal

economically optimal foot strike pattern.
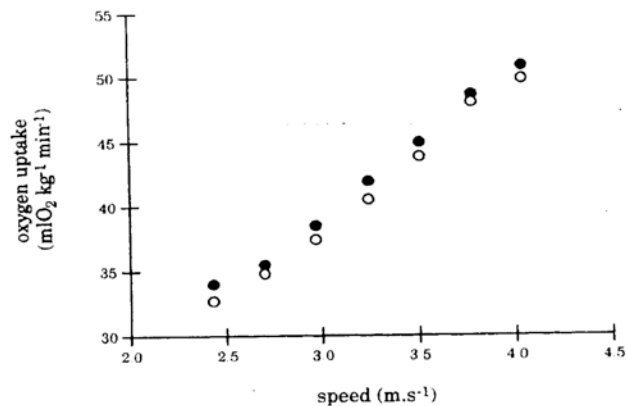
Figure 2.3. Relationship between oxygen uptake and foot strike pattern. This figure, adapted from Ardigo et al. 1995, shows that there is no discernable relationship in oxygen uptake between RF and FF running groups. The closed circles represent RF runners, and the open circles represent FF runners. On the X-axis is speed in ms^{-1}, while oxygen uptake in mlO$_2$kg^{1}min-^{1} is on the Y-axis.

15

2.5 Kinetic differences in strike patterns

There are several kinetic differences between FF and RF strike patterns. For this literature review, the focus will be on the differences in vertical ground reaction forces for the two different patterns. The vertical ground reaction force is the force of the ground pushing up at the runner during the stance phase, which is equal and opposite the force the runner applies to the ground (Novacheck, 1998). During running, this force has been shown to be a 2-3 fold increase over forces experienced during walking, with peak values of about 3 bodyweights (Cavanagh & Lafortune, 1980). Both RF and FF running styles exhibit different characteristics over the time trace of stance.

In RF runners, the vertical ground reaction force curve exhibits two peaks: the impact, or passive, peak, and the active peak while FF runners only exhibit a second active peak (Cavanagh & Lafortune, 1980; Derrick, Hamill, & Caldwell, 1998; Dickinson et al., 1985; Novacheck, 1998). The two force traces can be seen in Fig. 2.4. The impact peak marks the middle of the impact phase, defined by Derrick et al. (1998) as the twice the time to the peak force value experienced at impact. This can be seen in Fig. 2.5. The shock attenuation during the impact phase is considered passive, in that the majority of attenuation occurs without active input from the musculature, with the heel pad of the foot, cushioning in the shoe, and skeletal tissue absorbing most of the shock (Derrick et al., 1998). This peak is typically around 2 body weights, occurs at about 25 ms after initial ground contact, and is not present in FF runners in nearly the same magnitude (Cavanagh & Lafortune, 1980; Dickinson et al., 1985; Laughton et al., 2003; Munro, Miller, & Fuglevand, 1987). The impact peak increases with increased running velocity (Cavanagh & Lafortune, 1980). This peak has been suggested to be a cause for increased

injury rates, and a major talking point for promoters of FF running; however, the mere

existence of this impact peak has not been shown to have any direct correlation with

injury (Bates, Osternig, Sawhill, & James, 1983; Nigg, Cole, & Brüggemann, 1995).

Rearfoot Strike
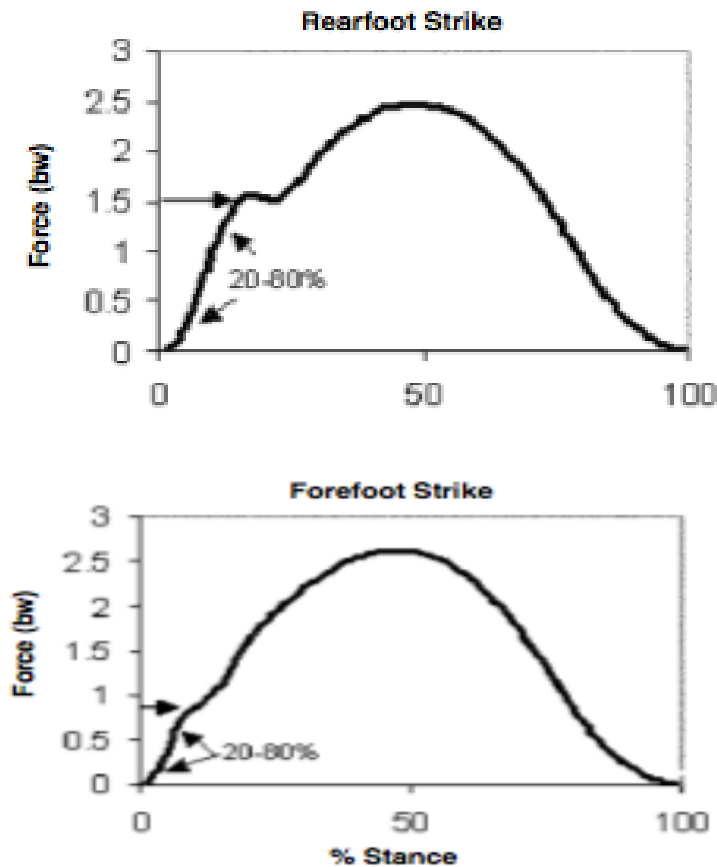
Forefoot Strike

Figure 2.4. Ground reaction force in foot strike patterns. The top image depicts a force trace of a typical vertical ground reaction force for a RF runner, in which two distinguishable peaks can be seen. The bottom image is the typical vertical ground reaction force curve for a FF runner, with a minor fluctuation in the slope of the loading rate curve and no real distinguishable impact peak. On the Y-axis is Force in bodyweights, while the X-axis is a percentage of normalized stance. The horizontal arrows mark the impact peaks, while the 20-80 percent arrows demonstrate the slope used to calculate loading rate. Adapted from (Laughton et al., 2003).

The active peak represents energy generation and occurs during propulsion (Derrick et al.,

1998). In FF runners, it has been shown that despite the lack of a prominent impact peak,

the overall peak magnitude of the active peak is higher than in RF running, with values of

2.64 BW for FF running and 2.48 BW for RF running

17
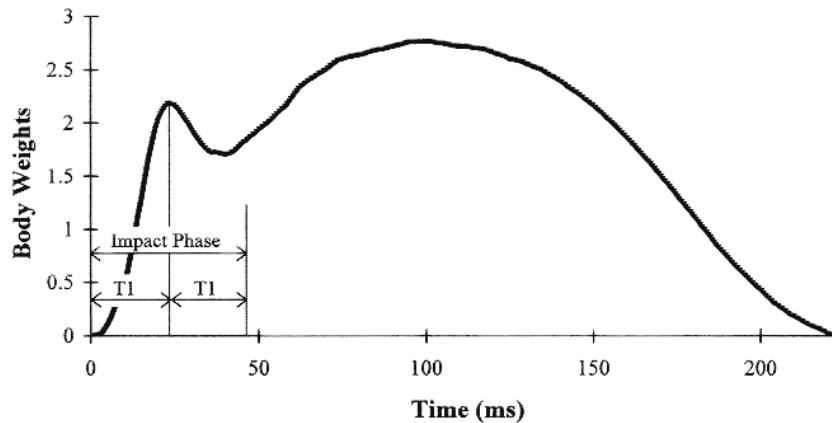
Figure 2.5. Rearfoot vertical ground reaction force profile. The impact peak and impact phase labeled. On the Y-axis is Force in bodyweights, while the X-axis is time in ms. Adapted from (Derrick et al., 1998)

(p=.0002) (Dickinson et al., 1985; Laughton et al., 2003). Another factor that can be compared between different vertical ground reaction curves is the loading rate, which is the slope of the line of force leading up to the impact peak and thus the time derivative of the force-time function (Laughton et al., 2003; Nigg et al., 1995; Novacheck, 1998). There has been contradictory results on loading rate differences between footfall patterns; however, generally the trend has shown that loading rate is higher in RF runners than FF runners (Laughton et al., 2003; Oakley & Pratt, 1988). Loading rate has been calculated as the slope of the force curve during 20-80 percent of the time to the impact peak in RF runners; in FF runners, loading rate was calculated from 20-80 percent of the time to the first change in slope of the force curve (Laughton et al., 2003). Representative measurements can be seen in Fig. 4 above. Another calculation method for loading rate involves calculating the slope of the force curve between the time it rises above 50 N to when it reaches BW plus 50 N (Munro et al., 1987). Loading rate is believed to be more

18

telling in terms of the actual instantaneous stressors the body is experiencing rather than the presence or magnitude of an impact peak (Laughton et al., 2003)

2.6 Kinematic differences in strike patterns

As the strike patterns are markedly different, the kinematics of RF and FF running differ greatly in the sagittal, frontal, and transverse planes. RF running is marked by foot contact with the ground with ankle dorsiflexion, allowing the heel to make first impact with the foot in a supinated position (Williams & Cavanagh, 1987). In FF runners, the foot lands plantar flexed, also in a supinated position, causing increased inversion of the foot at touchdown (Williams et al., 2000; Williams & Cavanagh, 1987). This initial plantar flexion angle causes peak and average plantar flexion values to be higher in FF runners compared to RF, indicating a higher level of strain on these plantar flexor muscles in FF running (Pohl & Buckley, 2008). However, overall dorsiflexion excursion is greater in FF runners than RF, with some studies reporting FF dorsiflexion excursion as 31.57±4.31˚ compared to 19.24±4.77˚ (Laughton et al., 2003). It is possible to determine foot strike pattern between FF and RF runners via ankle angle at foot strike alone with a success rate of 87 percent, thus indicating an accurate measure of foot strike patterns in the absence of force data, as shown in Figure 6 below (Altman & Davis, 2012). It has also been shown that while peak rearfoot eversion angles are significantly higher (p<0.05) in RF runners than FF runners, -11.1˚±2.8˚ to -6.2˚±2.6˚, the overall rearfoot eversion excursion is significantly higher for FF runners, 17.6˚±3.2˚ to 12.0˚±2.7˚, respectively, indicating a greater amount of rearfoot motion with a FF strike pattern (Pohl & Buckley, 2008). Another study indicated similar values of FF and RF rearfoot

eversion excursion, 16.39±3.94˚ and 13.66±4.08˚, respectively (Laughton et al., 2003)

Shank internal rotation, forefoot dorsiflexion and abduction excursions were all found to

be significantly higher in the FF running group than the rearfoot group, along with a

decreased time to peak forefoot dorsiflexion (Pohl & Buckley, 2008). These results can

be observed in the Figure 7, while a table of results is given in Table 1.
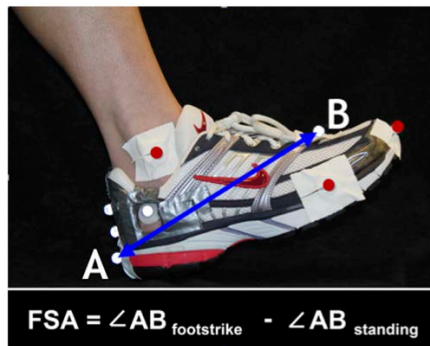
Figure 2.6. Foot strike angle definitions. This figure depicts how foot strike angle (FSA) is defined based on a marker place on the heel and second metatarsal head of the foot. FSA can be used in place of kinetic data when it is not readily or easily available. (Altman & Davis, 2012)

Table 2.1. Foot and shank kinematic variables for FF and RF runners. Adapted from (Pohl & Buckley, 2008).

Variables	HFS	TFS
Rearfoot peak EVE (°)	−11.1 (2.8)	−6.2[heel, fore] (2.6)
Shank peak IR (°)	−8.4 (2.9)	−5.0[heel, fore] (3.4)
Rearfoot EVE excursion (°)	−12.0 (2.7)	−17.6[heel] (3.2)
Shank IR excursion (°)	−5.8 (2.0)	−7.4[heel] (2.2)
Forefoot DF excursion (°)	7.3 (1.8)	13.9[heel] (2.1)
Forefoot ABD excursion (°)	−3.7 (1.2)	−6.5[heel] (1.4)
Time to peak rearfoot EVE (% stance)	43.6 (4.8)	43.9 (5.4)
Time to peak shank IR (% stance)	37.7 (10.9)	41.2 (8.9)
Time to peak forefoot DF (% stance)	58.2 (4.9)	52.1[heel] (6.8)
Time to peak forefoot ABD (% stance)	53.8 (11.5)	45.9[heel, fore] (8.4)
Rearfoot EVE/shank IR excursion ratio	2.0 (0.4)	2.4 (0.6)

Significant differences between conditions ($P < 0.05$) are shown by superscript.

The kinematic differences shown between RF and FF runners are not limited to the foot

and ankle. There have been shown to be differences at the knee as well, although not as

clearly reported in the literature. RF runners have been shown to have greater peak knee
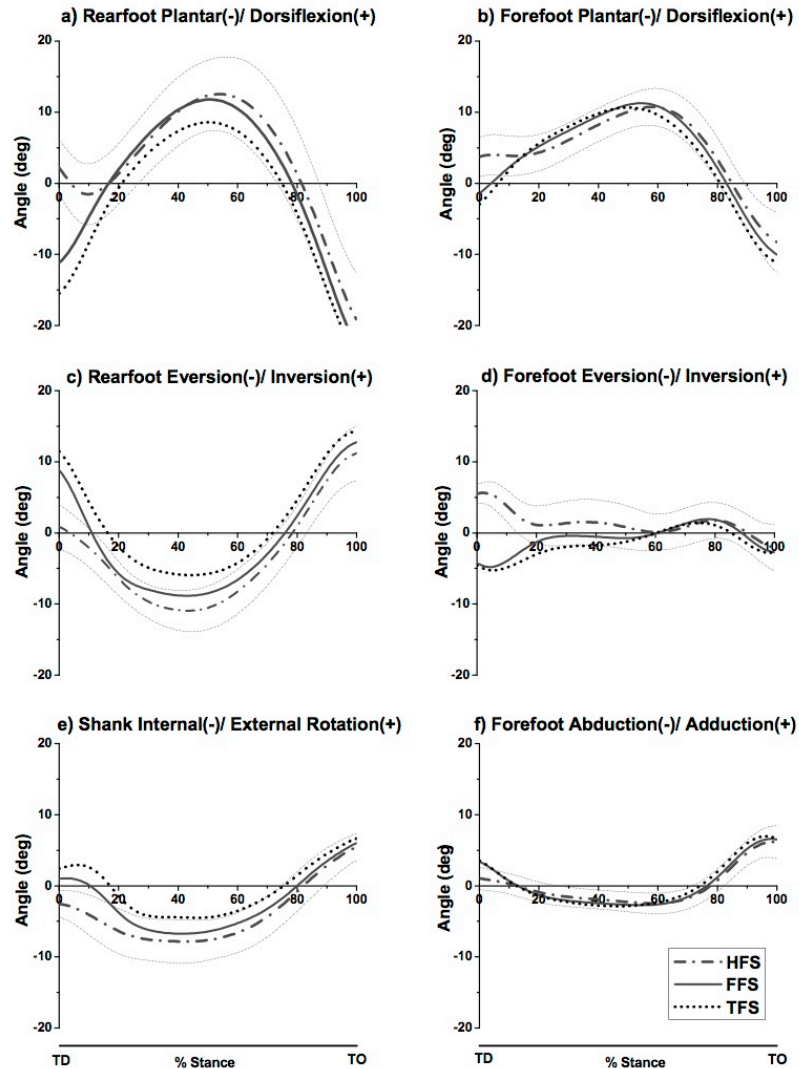
Figure 2.7. Angular displacement curves of rearfoot, forefoot, and shank. This figure depicts angular displacement curves for the forefoot, rearfoot, and shank for RF (HFS) runners, MF (FFS) runners, and FF (TFS) runners. The light gray solid lines represent means for all subjects of HFS, shown for illustrative purposes only. Adapted from (Pohl & Buckley, 2008).

flexion velocities in the first 30 percent of stance, while the remaining flexion velocity is

similar between both patterns (Williams et al., 2000). In the transverse plain, RF runners

experiencing greater peak internal rotation at the knee during the middle portion of stance,

while FF runners exhibit greater external rotation throughout stance, landing about three

degrees more externally rotated at touchdown (Williams et al., 2000).

2.7 Stretch-shorten cycle

Running with a FF pattern requires the plantar flexor muscles of the lower limb to contract eccentrically at ground contact, acting as a shock absorber during this active force attenuation phase (Laughton et al., 2003; Pratt, 1989; Williams et al., 2000). Most of the literature regarding repeated stretch-shorten cycling has shown a decreased capacity for muscular function, indicating that this mechanism of energy return in FF running could be reduced over time at high intensities. In a study on a submaximal fatiguing stretch-shorten cycle activity using subjects' arms, integrated EMG was shown to increase for both the concentric and eccentric phases of muscle activity, while the ratio between the two did not change, indicating an increase in neural output (Gollhofer et al., 2008). The authors believed that the increased neuronal output after the submaximal cycling was a compensation for reduced contractile function of the muscle due to fatigue (Gollhofer et al., 2008). When maximal drop loads were initiated following the repeated submaximal cycling, the eccentric level of activity stayed the same, while the concentric dropped significantly (p<.005), causing the ratio of eccentric to concentric integrated EMG to rise from .62 to .78 (Gollhofer et al., 2008). Thus, with fatigue in stretch-shorten cycles, the concentric contractile ability of the muscle can be decreased (Gollhofer et al., 2008). This indicates a potential loss in energy return from elastic muscular components in these cyclic activities with increased fatigue.

When runners fatigue, it has been shown that ground contact time increases (Hayes et al., 2004). The greater amount of time spent in stance phase causes the coupling time between eccentric and concentric loading of the plantar flexor muscles of the triceps surae to increase. An increase in coupling time between eccentric and

22

concentric loading of a muscle could potentially cause a dissipation of stored elastic

energy as heat rather than concentric force production (Bosco et al., 1981). In a study

utilizing countermovement and squat jumps to investigate the effect of prestretch velocity

and coupling time on end force, a negative correlation was shown (r = -.35) between

greater coupling time and end force, indicating that a longer time spent in between

contractile phases reduces force potentiation at the end of the activity (Bosco et al., 1981).

Another study examined the neuromuscular effects of a marathon length running

bout (Nicol et al., 1991). Both maximal force production and the rate of force onset were

both shown to be diminished following the run, and further running and jump testing

indicated that the marathon run modified the performance of muscles involved in the

stretch-shorten cycle (Nicol et al., 1991). These findings were supported by Komi et al.

(1986), which found that after a marathon run, there was a decrease in the ability of the

body to attenuate impact forces (Komi et al., 1986). The evidence for this was shown in

decreased average vertical braking forces in running, indicated by a rapid drop of vertical

force after the initial impact in sprinting tests (Komi et al., 1986). The authors believed

this was caused by a decreased capacity of musculotendon system to absorb the impact

shocks. Post marathon, knee flexion has been shown to increase just after heel strike in a

fatigued state, indicating the normal stretch-shorten cycles present in running operate in a

modified way (Nicol et al., 1991).

2.8 Fatigue protocols

There have been several studies examining the effects of fatigue on running and

other exercise tasks. Due to the wide breadth of research, there exist a large variety of

protocols differing in length, intensity, and mode. As this study is aimed at inducing local muscular fatigue via running, the majority of this review will revolve around this topic. Individual results for fatiguing studies will be examined in the next section.

When establishing a running fatigue protocol, there are several different variables that must be taken into account. The dependent variables in question focus the fatigue protocol: does the experimenter induce fatigue with increased duration, velocity, or some other limiting factor? In order to avoid selecting a random speed for a prolonged run, several different methods of calculating speed have been previously used in the literature. Some studies have been limited by their equipment, and have been forced to select slower than optimal speeds. In one particular study, it was shown that maximal treadmill speed caused the prolonged run speed to differ from self-selected speed overground trials by a significant margin, with self-selected overground speed recorded at 4.61 ms^{-1} while maximal treadmill speed was 2.75ms^{-1} (Dickinson et al., 1985). This was not the only limitation of this study; only three overground trials were collected for every 15 minutes of the prolonged run as more time off of the treadmill would cause too much recovery from any noticeable treadmill induced fatigue (Dickinson et al., 1985). It had previously been recommended that at least eight trials were needed in order to obtain stable force plate values and make appropriate comparisons between fatigued states (Bates et al., 1983).

Apart from equipment limitations, speeds have been determined based on initial physiological tests and presence of specific fatigue markers (Abt et al., 2011; Dutto & Smith, 2002; Lepers et al., 2000; Mizrahi et al., 2000b; Mizrahi et al., 1997). In a study performed by Mizrahi et al. (1997), there were two visits involved: a testing session and

an exhaustive run. First, the anaerobic threshold (AT) of each runner was determined

from oxygen data, determined non-invasively using end-tidal carbon dioxide pressure, a

proven and reliable method for measuring AT as well as overall fatigue, collected while

subjecting the runner to increasingly difficult loads (Mizrahi et al., 1997; Muza, Lee,

Wiley, McDonald, & Zechman, 1983). This method has been shown to be a more

accurate measure of performance loss and thus fatigue than blood lactate levels

(Wasserman, Whipp, Koyal, & Beaver, 1973). The prolonged run was conducted for

thirty minutes, during which seven different collection periods occurred at five-minute

intervals while accelerometer and EMG data was collected (Mizrahi et al., 1997).

Another study was later performed by the same group with an identical protocol,

although the participants were worked at a slightly higher intensity of 5 percent above the

determined AT speed (Mizrahi et al., 2000b).

In a similar fashion, four other studies conducted their experimental protocol on

separate visits (Abt et al., 2011; Derrick et al., 2002; Dutto & Smith, 2002; Lepers et al.,

2000). The study performed by Abt et al. (2011) first had participants undergo a

VO_2max test using a Modified Astrand protocol to exhaustion while recording heart rate

(Abt et al., 2011). These results were used to identify the ventilatory threshold (VT) and

its corresponding heart rate (Abt et al., 2011). During their fatigue protocol, participants

were ramped up to a speed that would elicit this heart rate response, and then were asked

to run to exhaustion (mean duration: 17.8±5.7 minutes) (Abt et al., 2011). Kinematic and

accelerometer data were collected before and after the protocol, but not during. In

another study, a similar incremental test was used to establish maximal aerobic speed

(Lepers et al., 2000). However, the length of the fatigue protocol used in this study was a

2-hour run conducted at 75 percent of maximum aerobic speed, over which the participants traveled an average of 28.4±1.3 km (Lepers et al., 2000). EMG, isokinetic peak torque of knee extensor muscles, and maximal counter movement jump height were the measures examined in this study (Lepers et al., 2000). Dutto and Smith (2002) conducted a study in similar fashion, using a maximal test to establish a baseline measure. A run was conducted at 80 percent of the speed corresponding with peak oxygen consumption to examine the effects of fatigue on leg stiffness (Dutto & Smith, 2002). The run time ranged from 31 to 90 minutes for fifteen subjects, a very wide variation, during which vertical ground reaction forces were measured (Dutto & Smith, 2002). A less-traditional test was conducted in the form of a 3200 meter time trial on an a week prior to the prolonged run to establish a pace baseline (Derrick et al., 2002). The average speed of the time trial was used as the set pace on the treadmill, and participants ran until they were unable to continue (Derrick et al., 2002). This form of maximal test has the benefit of establishing an appropriate baseline speed for testing but does not carry the same subject risks inherent with conducting an incremental test on a treadmill. This study also controls for only one factor, speed, and does not depend on physiological markers as the other protocols reviewed thus far have.

Another study completed their testing on one day, at a pace that was chosen as a self-selected, subject specific, typical training pace (Dierks, Davis, & Hamill, 2010). The run was conducted on a treadmill and continued until the subject either reached 85 percent maximum estimated heart rate or had a score of 17 on the Borg RPE scale (Borg, 1971; Dierks et al., 2010). The focus of investigation here was potential changes in joint kinematics.

In addition to the research performed on a treadmill in a laboratory setting, several studies have been conducting in outdoor venues in both experimental and competitive environments. One study conducted by Freisenbichler et al. (2011) asked recreational runners to maintain a constant pace, based off of previous 10 kilometer race performances, and perform as many laps around a 230-meter course as possible. Splits were given every lap to ensure pace adherence, and the laps were continued until the participants were no longer able to maintain the pace or continue voluntarily (Friesenbichler et al., 2011). Fatigue levels were measured throughout the run based on rate of perceived exertion levels (RPE) each lap (Figure 8) (Borg, 1971), and each subject had to reach a high degree of fatigue by the end of the test to ensure appropriate fatigue (Friesenbichler et al., 2011).

Table 5.2A The original Borg Scale Rating Perception of Effort (RPE)	
Rating	Perception of effort
6	
7	Very, very light
8	
9	Very light
10	
11	Fairly light
12	
13	Somewhat hard
14	
15	Hard
16	
17	Very hard
18	
19	Very, very hard
20	

Figure 2.8 Borg Scale Rating Perception of Effort (RPE). Adapted from (Borg, 1971).

In the competitive realm, there are no experimental measures of fatigue other than the goal of finishing a race in the shortest possible amount of time. In recreational and elite athletes, competitive environments suggest a nearly maximal effort in performance

(Bates et al., 1977). The range of competitive distances studied has been widespread, with fatigue effects examined from short distances of 400 meters, through middle distance track events, up to half marathon and marathon endurance races (Bates et al., 1977; Elliot & Ackland, 1981; Hasegawa et al., 2007; Hayes & Caplan, 2012; Kerr et al., 1983; Larson et al., 2011). In several of these studies, cameras were used to collect data at various checkpoints in the race. One study examined elite-level female 400-meter runners in a competitive event at distances of 169 and 370 meters from the start line to observe "fresh" running and running in a fatigued state (Bates et al., 1977). Stride frequency and several temporal factors were examined (Bates et al., 1977). Hayes et al. (2012) examined foot strike pattern and ground contact times of 181 male and female elite-level 800- and 1500-meter runners on a lap-by-lap basis.

The other studies reviewed performances in longer distance events. In one, an examination was performed on eight skilled 10,000 meter runners, whom were recorded on lap 2, lap 10, lap 17, and lap 24 of the 25 lap race on a track to observe fatigue effects on running velocity, stride length and rate, as well as several kinematic measures (Elliot & Ackland, 1981). The final three competitive studies focused on road races from 10 kilometers to 42.2 kilometers (marathon). In the 10 km race, participants were recorded once at the 9km mark in their fatigued state (Kerr et al., 1983). In the half marathon, runners were recorded at the 15 km mark (Hasegawa et al., 2007), while checkpoints of 10k and 32k and 20k and 35k, respectively, were used in the marathon studies (Kerr et al., 1983; Larson et al., 2011).

Apart from fatigue induced directly through running, the effect of fatigue brought on prior to an exercise bout on running mechanics has been investigated. These studies

typically focus on fatiguing a specific set of muscles inherent to running (Christina, White, & Gilchrist, 2001). Christina et al. (2001) chose to directly induce local muscle fatigue in either the dorsiflexors or invertors of the right foot, using the rational that "exhaustive running protocols fail to distinguish when kinematic or kinetic changes are a direct result of localized muscle fatigue". This study consisted of two visits separated by two weeks to allow for each subject to be subjected to the separate muscle group fatigue protocols (Christina et al., 2001). The running portion occurred in a pre-fatigue and post-fatigue setting on a treadmill at a speed of 2.9 ms^{-1} during which kinetics and kinematics were measured (Christina et al., 2001). Fatigue was induced using repetitive concentric and eccentric contractions on an Elgin leg/ankle exerciser, using torque data to ensure appropriate fatigue (Christina et al., 2001). Torque was also monitored after the second running bout to examine muscle recovery (Christina et al., 2001). This type of test is reliable for inducing local muscular fatigue, but does not account for overall fatigue in the other major muscle groups used during running.

A similar study was performed to examine the effects of specific muscle group fatigue on ground reaction forces, kinematics, and EMG activity during a landing exercise (Madigan & Pidcoe, 2003). In this protocol, subjects warmed up on an ergometer for 10 minutes and then performed a cyclic fatigue exercise of two single leg drops and three single leg squats to exhaustion (Madigan & Pidcoe, 2003).

2.9 Fatigue effects

Local muscular fatigue has been shown to affect both the functions of the fatigued muscle as well as the functions of related parts (Mizrahi et al., 1997), and local muscular fatigue

has been shown to induce several changes in individual running mechanics. This section will expand on the results of the fatigue studies addressed in the previous section.

2.9.1 Kinetic

In the study performed by Dickinson et al. (1985), it was shown that the magnitude of the heel strike impact peak increased from 186 percent bodyweight prior to the fatiguing treadmill run to an average peak force of 203 percent bodyweight, with a peak force during the run of 211 percent bodyweight 30 minutes in. This increase in peak impact force was believed to be caused by decreased muscle force attenuation capacity and altered proprioception, leading to a fatigue-altered gait (Dickinson et al., 1985). In this study, it was assumed that fatigue of the lower extremity muscles caused more loading to occur on the heel at impact, potentially caused by a decrease capacity of the soft muscular tissue to attenuate the shock (Dickinson et al., 1985). In a study examining vertical and leg stiffness changes over the course of a prolonged run, it was shown that the stiffness of both significantly decreased (p<0.001) by the end of the run (Dutto & Smith, 2002). Vertical stiffness was defined as the ratio of peak force to peak center of mass displacement, and it was shown to decrease despite relatively constant peak forces due to an increase in overall ground contact time (Dutto & Smith, 2002). This increase in contact time caused a greater displacement of the center of mass, and was confirmed in other studies (Dutto & Smith, 2002). Leg stiffness was calculated as a ratio of peak vertical force to leg length change during stance, which was calculated with Equation 1 and 2 (Dutto & Smith, 2002).

$$\Delta L = \Delta y_{cm} + L_0 (1 - cos \emptyset) \qquad \text{Eq. 1}$$

$$Leg\ Stiffness = \frac{Peak\ Vertical\ Force}{\Delta L} \qquad \text{Eq. 2}$$

In this equation, L_0 is the resting leg length and $\emptyset$ is the angle of the leg at initial contact relative to the vertical. ΔL is the leg length change used in calculating leg stiffness in Eq. 2. With fatigue, leg stiffness was shown to decrease as well (Dutto & Smith, 2002). This change in stiffness, when coupled with eccentric loading, has been shown to decrease the force attenuating capability of a musculotendon complex, indicating that alternative attenuation strategies such as altered gait or neuromuscular control must be adopted (Dalleau et al., 1998)

In the two studies designed to induce muscle fatigue in specific groups, kinetic differences between pre- and post-fatigue were observed. With fatigue induced in the dorsiflexors, there was an increase shown in impact peak force loading rate by 20.09 BW/second (Christina et al., 2001). No difference was shown in peak pushoff force or impact peak force (Christina et al., 2001). In the case of invertor fatigue, there were significant decreases shown in peak impact force, by .1 BW, and peak pushoff force (.06 BW) (Christina et al., 2001). These results are shown in Figure 9. In the study by Madigan et al. (2003), vertical ground reaction peak forces decreased in the fatigued group along with the overall vertical impulse, despite no change in loading rate (Madigan & Pidcoe, 2003). The reduction in vertical force was accompanied by an increase in joint flexion (Madigan & Pidcoe, 2003).
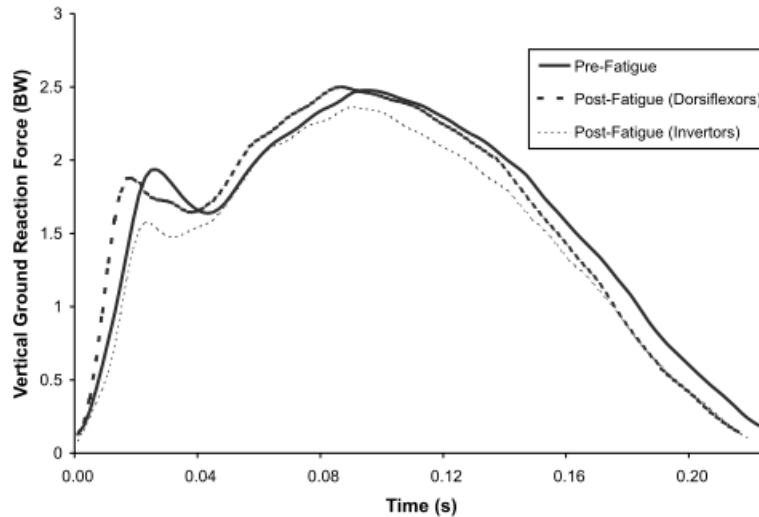
Figure 2.9. Fatigue related changes in vertical ground reaction forces. Pre-fatigue levels are given as the solid black line, and the other two lines represent the dorsiflexor fatigue condition and the invertor fatigue condition. Adapted from (Christina et al., 2001).

The kinetic evidence presented indicates that with fatigue, there may be decreases in peak vertical forces and an increase in impact loading rate and ground contact time, along with the potential for decreased leg and overall vertical stiffness.

2.9.2 Kinematic

The majority of the studies reviewed with fatigue protocols indicated some changes in kinematic data somewhere along the lower limb chain, although some showed little or no significant changes in kinematic variables despite altered kinetics (Abt et al., 2011; Christina et al., 2001). In the study by Mizrahi et al. (2000), several significant changes were found between the beginning of running and the 30th minute of the prolonged run. Average stride rate was shown to decrease by .07 strides per second, while average knee flexion at foot strike decreased from 13.6±6.3° to 8.1±3.0° (Mizrahi et al., 2000b). Possibly most notable was the increase in maximum knee extension position prior to foot strike, which increased from the first to 15th to 30th minutes with

32

values of 13.8±3.1˚, 16.8±3.7˚ and 17.2±4.2˚, respectively (Mizrahi et al., 2000b).

Similar results in knee extension with fatigue had been shown previously by Elliot and

Ackland (1981), where a greater knee extension at touchdown was shown to require a

greater hip moment to maintain similar foot strike kinematics. In addition, it was shown

that the relative backward velocity of the ankle at foot strike significantly decreased,

indicating less preparation of the body for placing the site of ground contact directly

under the center of mass and thus, a greater propensity for a heel strike (Elliot & Ackland,

1981).

After
While peak knee extension was shown to increase with fatigue, it has also been

found that an increase in both knee flexion at heel strike and peak knee flexion during

stance occurs with fatigue (Derrick et al., 2002; Madigan & Pidcoe, 2003). It has been

speculated that the increase in knee flexion may cause a reduced contribution of the

biarticular gastrocnemius muscle, reducing plantar flexion (Madigan & Pidcoe, 2003).

The increase in knee flexion comes at a potential metabolic cost, as decreased in leg

stiffness can cause as high as a 20 percent increase in metabolic cost of an activity as well

as a decrease in overall elastic energy return (Dalleau et al., 1998; Derrick et al., 2002).

One major consideration in fatigue studies is the potential for altered coordination

of segment and joint timing. A conclusion, despite few kinematic changes with local

muscular fatigue, by Christina et al. (2001), is that with increased muscle fatigue and

decreased force potentiation comes a change in the temporal relationship of segment

orientation just after ground contact. The authors postulated that when a muscle acts

eccentrically at heel strike across a joint, a fatigued state may prevent a "desired" foot

angle from being achieved due to the muscle's inability to contract concentrically at an

appropriate level (Christina et al., 2001). Dierks et al. (2010) found evidence of increased rearfoot eversion and peak knee internal rotation angles, excursions, and velocities while running in an exerted state, along with increased peak tibial internal rotation and hip adduction velocity. Despite this fact, there were no significant timing differences in joints at this exerted state, although rearfoot eversion throughout stance was shown to be significantly different in excursion, peak angle, and peak velocity (Dierks et al., 2010). It was noted in this study that the exertion level was not to exhaustion, and that even moderate fatigue was shown to cause subtle differences in kinematics at the ground contact point during a prolonged run (Dierks et al., 2010)

Evidence from previous fatigue studies has indicated that increased knee extension, increased peak knee flexion, decreased rearwards ankle acceleration prior to touch down, and decreased leg stiffness are all common in fatigued runners. While potentially altered joint timing and coordination mechanisms have been met with contradicting studies, it appears that further investigation into this matter is well warranted.

2.9.3 Electromyography

Muscular fatigue has been measured with change in EMG activity in terms of temporal activation, overall root means squared amplitude (RMS), total integrated EMG, and frequency analysis. Previous literature has suggested that fatigue of a runner may lead to mechanical changes in landing patterns due to muscle preactivation level changes (Bobbert, Yeadon, & Nigg, 1992). Despite this statement, a study by Mizrahi et al. (1997) showed little to no difference in mean power frequency or amplitude with fatigue. The

34

authors believed this to be caused by a level of muscle activation and force production too far below maximum voluntary contraction levels, at which changes in frequency and amplitude have decreased correlation with fatigue (Mizrahi et al., 1997). In the cyclic squatting exercises from another study, it was shown that the quadriceps muscles showed a 15.4 percent decrease in EMG mean frequency data, indicating a significant ($p<0.01$) level of fatigue (Madigan & Pidcoe, 2003). Cyclic, repetitive loads in running have been shown to produce neuromuscular changes over a marathon distance in running, namely in the overall integrated EMG of the vastus medialis and vastus laterali muscles (Nicol et al., 1991). Another fatiguing running study indicated that RMS activity was decreased in these muscles after a 2 hour run at nearly every isokinetic velocity tested (Lepers et al., 2000). These total losses in RMS activity were measured at 13 percent isometrically, 11 to 17 percent eccentrically and 12 to 21 percent under concentric conditions (Lepers et al., 2000). These losses can be viewed in the Figure 10. Another study examined the difference in fatigue in running between a neutral and motion control shoe (Cheung & Ng, 2010). These authors found that over the course of a 10 km run, the normalized RMS activity in the motion control shoe was significantly less than that in the neutral shoe, indicating that the motion control shoe limited local muscular fatigue compared to the neutral shoe (Cheung & Ng, 2010). Median frequency was also shown to be significantly different between the two footwear types, and also shown to decrease in both shoe types over the course of a prolonged run (Cheung & Ng, 2010).
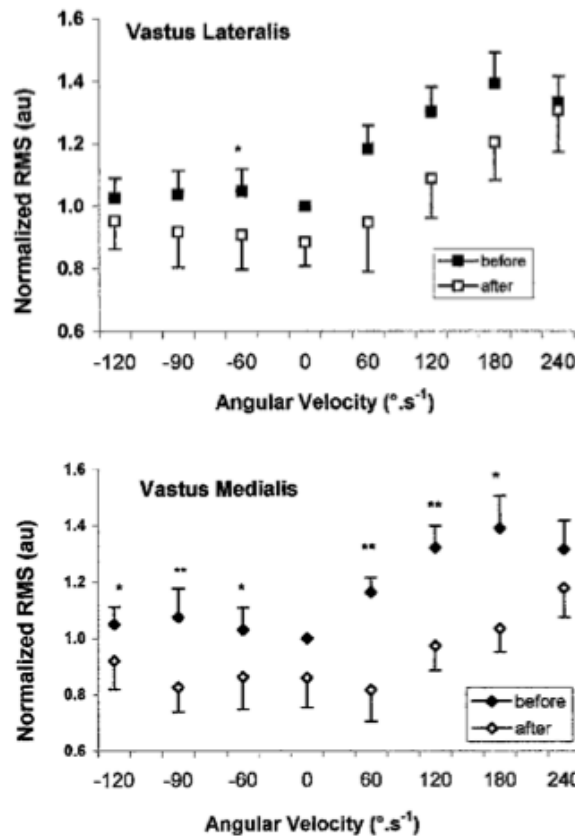
Figure 2.10. Pre- and post- normalized root means squared (RMS) values of Vastus Lateralis and Vastus Medialis muscles. Adapted from (Lepers et al., 2000).

The use of surface EMG in fatigue analysis of activities outside of constant-force, isometric contractions presents potential problems of non-stationarity in the signal, which can be caused by accumulation of metabolites in the muscle or due to actual motion of the muscle being studied (Bonato, Roy, Knaflitz, & De Luca, 2001). Dynamic activities can cause the skin sensors to move relative to the muscle belly, and changes in fiber length and velocity have an effect on how the signal is transmitted (Bonato et al., 2001). One way this has been accounted for has been the use of a time-frequency spectrum of the EMG signal as opposed to a power density spectrum (Bonato et al., 2001). From the time-frequency spectrum, instantaneous median and mean frequencies can be defined and

estimated using a Choi-Williams transform (Bonato et al., 2001). After the transform, only a small "burst" of EMG signal is analyzed at similar biomechanical event times throughout the cyclic motion in order to reduce movement causes of frequency shifts while averages can be used across consecutive cycles if quasi-cyclostationarity is assumed (Bonato et al., 2001). While this technique is an advancement over what has been previously proposed, it is still new and limited in its scope.

2.9.4 Foot strike patterns

In the majority of the competitive studies performed outside a laboratory setting, there were no motion capture systems or force platforms to collect kinetic and kinematic data. Most of the studies simply utilized video cameras and photographs to determine the strike pattern each runner was utilizing at specific checkpoints in each race.

Several of these competitive studies were performed around races on a track, with the competitive distances ranging from 800-10,000 meters (Elliot & Ackland, 1981; Hayes & Caplan, 2012). In the study conducted with male and female middle distance runners in 800-1500 meter races, 181 athletes were examined on each lap of their respective races to record ground contact time and foot strike pattern (Hayes & Caplan, 2012). The results in terms of foot strike pattern were much different at this middle distance than in some of the longer race performances measured previously (Hasegawa et al., 2007; Larson et al., 2011). Amongst all groups, the percentages for FF, MF, and RF runners were 31, 42, and 27 percent respectively; this is significantly skewed from the typical 80 percent RF group typically seen amongst runners (Hayes & Caplan, 2012). Changes in strike pattern were not monitored over the course of each race; however, it

37

was found that after the first lap, there was a significant increase in ground contact time for subsequent laps (Hayes & Caplan, 2012). Increased ground contact time has been previously indicative of a RF pattern, including greater knee extension at touch down (Williams & Cavanagh, 1987). When all runners were combined, it was shown that FF and MF runners had faster average race speeds than heel strikers, but when event was taken into account, the only significant difference was shown between MF and RF runners in the men's 1500 meter run (Hayes & Caplan, 2012). In another competitive track study, focused on the biomechanics of elite and sub-elite 10,000 meter runners, exact foot strike parameters were not recorded (Elliot & Ackland, 1981). However, notable results indicated that rearwards velocity decreased at each stage of the race, indicating difficulty in placing the foot underneath the runner's center of mass (Elliot & Ackland, 1981). This tends to lead to a more RF strike.

Three of the most notable large-scale studies on foot strike pattern over an endurance competitive event investigated races of 10 km, half marathon, and marathon distances (Hasegawa et al., 2007; Kerr et al., 1983; Larson et al., 2011). In the 10 km road race, it was shown that 81 percent of the runners utilized a RF pattern 9km into the race, and the other 19 percent were MF runners (Kerr et al., 1983). As the finish time in the race increased from 29 to 45 minutes, the percentage of RF strikers rose from 50 percent to 81 percent (Figure 2) (Kerr et al., 1983). In the marathon studied by the same authors, checkpoints were recorded at 20 and 35 km into the race (Kerr et al., 1983). At the 20k point in the race, 79 percent of the runners exhibited a RF pattern, while at the 35 km point, there was an increase of 3 percent in RF runners (Kerr et al., 1983). The authors showed a correlation between running velocity and the percentage of RF strikers,

and were unable to make a causal statement on whether the increase in proportion of RF runners was due to fatigue or decreased running velocity (Kerr et al., 1983). A similar result was shown by Hasegawa et al. (2007), where the top 50 male runners at the 15km checkpoint of a half marathon saw percentages of 62.0, 36.0, and 2.0 for RF, MF, and FF runners respectively, while the last 48 runners in order of finish showed percentages of 79.0, 19.0, and 2.0 percent (Hasegawa et al., 2007). The results were even more skewed for the women, with the top 20 percent of women showing percentages of 43.0, 43.0, and 14.0 percent, respectively, and the bottom 20 percent having a break down of 86.0, 14.0, and 0 percent (Hasegawa et al., 2007). Ground contact time tended to be greater in RF runners than MF and FF runners for nearly all groups, while both groups showed strong correlation coefficients for increased contact time relative to order of finish (r=.80 for RF runners, and r=.71 for MF+FF) (Hasegawa et al., 2007).

The last large-scale competitive study looked at foot strike patterns of recreational and sub-elite marathon runners (n=286) between the 10km and 32km check point (Larson et al., 2011). It was shown that the percentage of RF runners increased from 87.8 to 93.0 percent of the study population, while the FF runner percentage was reduced from 1.4 percent to 0 percent (Larson et al., 2011). Out of the total individual FF strikes observed at the 10 km point (n=13), all but one were shown to switch to MF or RF strike by 32km, while 59.5 percent of MF strikes were converted to RF strikes over this same distance (Larson et al., 2011). No significant differences in race finish time were shown between foot strike groups; however, the small sample size of non-RF runners causes the authors to caution further interpretation of this result (Larson et al., 2011). In fact, the authors believed that a potential reason for the larger percentage of RF runners reported in this

study compared to previous literature may have been caused by the more recreational race population observed (Larson et al., 2011).   It should be noted that this study was performed at video frame sampling rates more than double the previous studies (Larson et al., 2011). Thus, the difference in foot strike pattern classification between studies could be caused by an insufficient capture rate of the previous studies, with the information reported here potentially more accurate in representing naturally-occurring proportions of the global running population.

2.10 Summary

The foot is the first point of contact with the ground during running, and whatever happens during this initial contact point is relayed up the lower extremity chain. The foot strike pattern of most runners is divided up into three groups: rearfoot, midfoot, and forefoot runners. The most common pattern is rearfoot, while the least common is forefoot, making up less than five percent of all runners. Forefoot (FF) and rearfoot (RF) strike patterns each possess unique vertical ground reaction force profiles. RF vertical ground reaction forces contain a greater initial loading rate to a passive impact peak before a small decline leading in to the propulsive active peak. FF vertical ground reaction force curves have little or no impact peak, looking much like an upside-down "U" shape. Each strike pattern also lends itself to certain specific kinematic features, based around foot, ankle, and knee motion.

The differences between these strike patterns have been studied from a kinetic and kinematic standpoint to the point where each style is well understood.   In addition to the kinetic and kinematic variables, this review also focused on electromyography and the

stretch-shorten cycle. Different patterns of electrical muscle activity are observed with different exercise bouts; with fatigue, temporal relationships between muscle signals and the frequency content of those signals have been shown to change. FF running involves greater eccentric loading of the triceps surae and other plantar flexor muscles of the leg, and thus repetitive stretch-shorten cycles occur while running with this pattern. The dynamics of local muscular fatigue and repeated stretch-shorten cycles has been proposed to cause reduced muscular force production and attenuation in the concentric state, and could be a reason for altered running mechanics at higher exertion levels.

Despite the vast amount of research existing on these strike patterns and the factors influencing them, there has been little comparative research on the effects of each of these patterns over long, intense bouts of exercise. Several competitive and field studies exist on the topic, but no empirical evidence has been given. Many fatigue protocols were discussed in this review to induce fatigue in the musculature of the lower extremity. An understanding of each of these protocols is necessary to design a protocol to answer this specific research question. Having this knowledge will allow for a more detailed analysis of the possible gait changes in a forefoot strike pattern runner over the course of a prolonged run, add insight to what mechanisms could be causing this shift, and indicate specific markers and signs to pay attention to that could help quantify this potential change.

CHAPTER 3

METHODS

3.1 Introduction

The purpose of this study is to examine the 3D kinematics, kinetics, and electromyography of forefoot strike (FF) runners during a prolonged run. It is hypothesized that over the course of a prolonged run:

Hypothesis 1: There will be significant kinematic differences between each collection period of the prolonged run in the peak values and excursions of the sagittal ankle angle, sagittal knee angle, forefoot-rearfoot inversion-eversion angles, and tibial internal-external rotations.

Hypothesis 2: The vertical ground reaction force curve will begin to exhibit a steeper loading rate as well as a more pronounced impact peak in the first 30 percent of stance. Vertical and leg stiffness will decrease throughout the prolonged run.

Hypothesis 3: There will be a leftward shift in the frequency domain of the EMG signal in the triceps surae between pre- and post- exercise isometric contractions, indicating lower frequency activity in the muscles commonly seen with the onset of local muscular fatigue. There will also be a decrease in the integrated EMG in the pre- and post-exercise collections, accompanied by a decline in overall force production.

3.2 Participants

The participants that will be recruited for this study will be healthy males between the ages of 18-45. An "a priori" sample size calculation yielded a necessary sample of n=15 in order to achieve significance for the collected variables (α=0.05, β=0.20), which

was calculated based on previous reported means in the literature (Laughton et al., 2003). The population of true forefoot runners is a small percentage of the available running population, less than 2 percent in some reported studies (Hasegawa et al., 2007). Due to this reason, runners will be accepted into this study depending on availability if they exhibit foot strike characteristics that fall in between forefoot and midfoot striking as the hypotheses should still be applicable. Every participant must complete a modified physical activity readiness questionnaire (PAR-Q) and sign an informed consent approved by the Institutional Review Board. All participants must be experienced runners (minimum 15-20 mi/week) who have been injury free for the previous year with no major surgical histories. An initial overground screening process will be conducted for all participants to ensure that they exhibit the appropriate foot strike patterns necessary for participation in this study.

3.3 Experimental setup

There will be several different measurement systems in place to collect the kinetic, kinematic, and electromyographic data necessary to complete this study. All collections will be conducted in the University of Massachusetts Biomechanics Lab.

3.3.1 Instrumented Treadmill

In order to collect kinetic data, an instrumented treadmill will be used (Athletic Republic, Park City, UT). The treadmill rests on four load cells (AMTI, Watertown, MA) and vertical ground reaction forces will be determined by summing the forces recorded by each load cell. Force data will be sampled at 1000 Hz, and will be collected via Qualisys Track Manager (Qualysis, Gothenburg, Sweden).

To collect the 3- dimensional kinematic data, the participants will be fitted with 31 retro-reflective markers that will be tracked by eight infrared cameras (Pro Reflex MCU 240, Qualysis, Gothenburg, Sweden) sampling at 200 Hz. A unilateral maker set up will be used on the right leg, with toe and heel markers on the left foot to be used to determine stride length. The pelvic segment will be demarcated by markers on the right and left anterior superior iliac spines (ASIS), iliac crests, and greater trochanters as well as a sacral marker. Plates of 4 non-colinear markers will be place on the right thigh and shank, and calibration markers will be placed on both the right lateral and medial mallelioli and knee. A modified multi-segment foot model will be used, with markers on the hallux, first, second, and fifth metatarsal heads and bases, and a heel triad (Leardini, Benedetti, Berti, Bettinelli, Nativo, & Giannini, 2007; Myers, Wang, Marks, & Harris, 2004). Motions of the pelvis, thigh, shank, and foot segments will be tracked in 3-dimensional space using a minimum of 3 non-colinear markers per segment. Examples of the marker setup are given in Figures 11-13. A calibration of the camera system will be performed using a known-distance T wand with an L-bracket placed at the lab coordinate system origin.
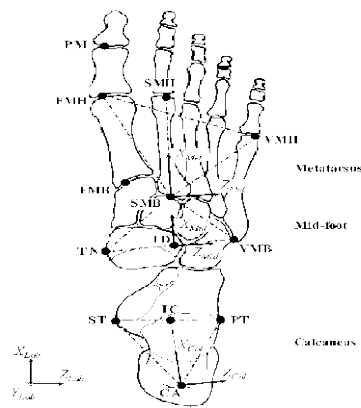
Figure 3.1. Multi-segment foot model. An adapted version will be used in this study. Adapted from (Leardini et al., 2007).
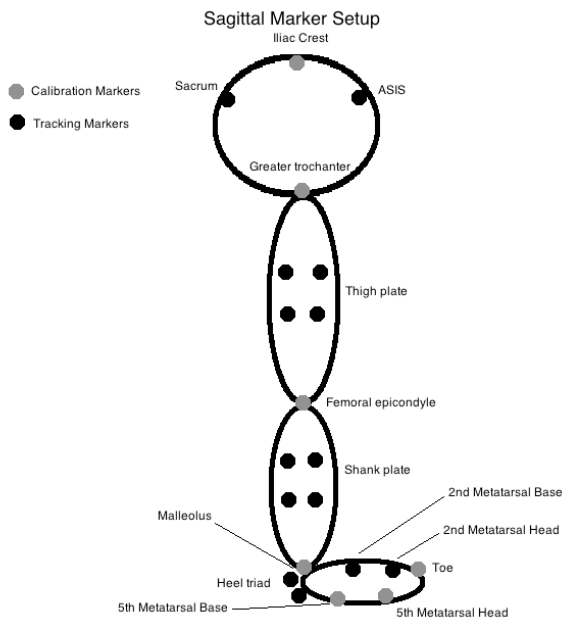
Figure 3.2. Sagittal marker setup. The grey markers represent calibration markers that will be used to define segments, and the black markers represent markers used to track segments. Some calibration markers will be used for both.
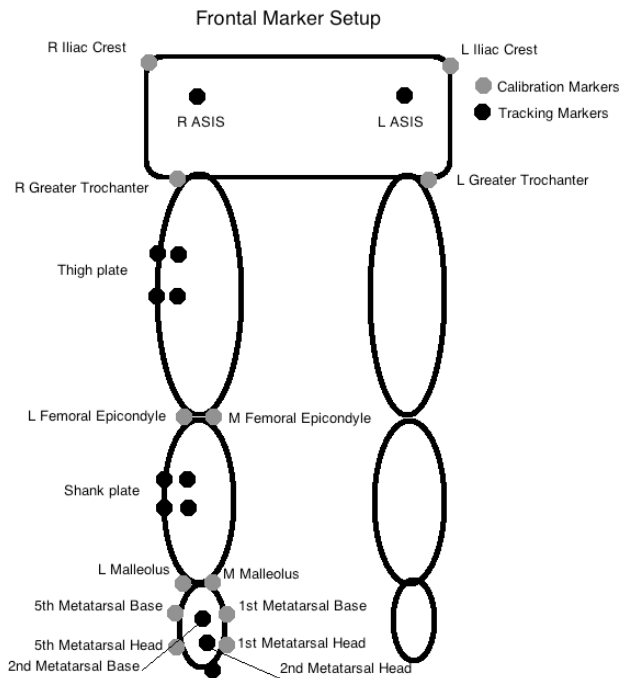
Figure 3.3. Frontal marker setup. The grey markers represent calibration markers that will be used to define segments, and the black markers represent markers used to track segments. Some calibration markers will be used for both.

45

3.3.3 Electromyography

Electromyographic data will be collected over the duration of the prolonged run on six different sites: the tibialis anterior, medial and lateral gastrocnemius, soleus, rectus femoris, and biceps femoris. At each of these sites, a wireless electrode will be placed, which transmits data to a Trigno Wireless EMG System base station (Delsys, Boston, MA). Each electrode has a mass of less than 15 grams, and records both EMG data and tri-axial accelerations. The location of each electrode will be determined by palpating participants' muscles and placing each sensor on a shaved and abraded site on the muscle belly to ensure an accurate, clean signal. The EMG data will be sampled at 2000 Hz, and data acquisition will occur with EMGWorks (Delsys, Boston, MA). The collection of all data will be initiated using a Delsys trigger module, which simultaneously begins a collection of all the data.

3.4 Experimental protocol

Participants for this study will be recruited primarily from the University of Massachusetts Track and Field teams, as well as other competitive university, club and individual races around the Amherst, MA area. When the participant first arrives at the lab, they will fill out an informed consent document approved by the local Institutional Review Board (IRB), outlining the procedures and associated risks of the study, as well as a modified Physical Activity Readiness Questionnaire (PAR-Q) commonly used in the department. They will fill out a participant demographic to determine their previous running history and race performances, which will later be used to calculate the appropriate experimental velocity. Any participant questions will be answered prior to

data collection. On the occasion that any participant answers, "No" to any question on the PAR-Q without medical approval, they will be excluded from the study.

Once the participant has been cleared for the study of appropriate health and running status, they will be screen to ensure that they exhibit a FF running pattern both over ground and on the instrumented treadmill. To perform the screening process, retro-reflective markers will be placed on their heels and toes. They will be asked to run through the collection window over the force platforms (AMTI, Watertown, MA) at their preferred running speed. For the over ground trials, they will have their starting position adjusted to ensure that they land on a force platform without altering their natural stride. From collected kinetic and kinematic data it will be possible to tell if they are utilizing a FF or RF strike pattern. Five screening trials will be performed to achieve a consistent average between the trials and accurately assign the participant into a FF or RF group. Once they have competed the screening process, they will either be scheduled for a treadmill collection session if placed into the FF running group, or they will be thanked and excluded from the remainder of the study if they utilize a RF pattern.

For those in the FF running group, the screening process will then move to the instrumented treadmill. The participant will be allowed to warm up on the instrumented treadmill at a self-selected, comfortable pace for up to 10 minutes. After this warm up period, the participant will be given ample recovery time to ensure that they enter the collection period in a fresh, prepared, and unfatigued state. The projected experimental velocity will then be set on the treadmill with the participant standing on the belt, allowing a short period of time to ramp up to full speed. The participant will then be asked whether or not they believe they can sustain the speed for 30 minutes. The speed

will be increased or decreased based on this response. Once the exact condition speed is selected, the screening session will be terminated, and the participant will be scheduled for the next collection session.

Upon arrival for the second session, a similar protocol to the screening process on the treadmill will be followed. The participant will be allowed a ten minute warm up period, after which the pre-selected condition speed will be tested. Further adjustment to the speed can be made at this time based upon the participant's current subjective status. After this period, data acquisition hardware will be placed on the participant in accordance with the experimental set-up above.

Once the participant has been fitted properly with all the acquisition hardware, the participant will be asked to perform an isometric calf raise while shouldering a load of 40 percent body mass while standing on an in ground force platform. The contraction will be held for as long as voluntarily possible, with the time of the contraction and muscle activity being recorded throughout. After this contraction, the participant will be given ample time to recover. Once fully recovered, the participant will be asked to stand on the treadmill and be ramped up to the experimental speed. This ramp up period will last for under one minute. Once the peak experimental speed is reached, the collection clock will begin, with this point marked as time zero. Every 5 minutes, a data collection of 15 seconds will occur. Collections will occur at time=0, 5, 10, 15, 20, 25, and 30 minutes. After the 4th collection period at 15 minutes, the participant will step off the treadmill, the belt will be stopped, and the load cells will be zeroed to ensure the forces stay within the collection range. The participant will then step back onto the treadmill, be brought back up to experimental speed, and the remainder of the prolonged run will continue.

Throughout the prolonged run, the participant will be monitored closely for safety, and every 5 minutes after the collection period, their rate of perceived exertion will be asked for and recorded (Borg, 1971). The participant will be asked each time if they are able to continue, and in the event they are not able to continue, a final collection will occur (if able) to mark the end of the collection period. Immediately after the prolonged run is completed, or the participant is unable to continue, another weighted, isometric calf raise will be performed, during which kinetic and EMG data will be recorded.

3.5 Data analysis

Kinematic data was tracked and labeled in Qualysis Track Manager and exported to .c3d files (Qualysis, Inc., Gothenburg, Sweden). Data were filtered using a dual-pass, low pass Butterworth filter with a cutoff frequency of 12 Hz. Sagittal ankle and knee angles, tibial internal-external rotations, and forefoot-rearfoot eversion inversion were calculated using a Cardan Xyz rotation sequence in Visual 3D(C-Motion Inc., Germantown, MD). The data were interpolated, normalized to 101 data points, and tracked over five consecutive strides and averaged to create a mean for each collection period of the experiment.

Vertical kinetic data was analyzed in both Visual 3D and a custom designed Matlab program (Mathworks, Natick, MA). Peak impact and active forces as well as loading rate were calculated (Christina et al., 2001). The kinetic data were interpolated and normalized to 101 data points for each stance phase to allow accurate comparison between separate foot strikes and separate collection periods. A low pass, 4th order zero-phase lag Butterworth filter with a 30 Hz cutoff frequency was used to filter the data

(Laughton et al., 2003). Vertical stiffness was calculated as the ratio between maximum

displacement of the center of mass and peak vertical force (Eq. 1) (Dutto & Smith, 2002;

McMahon & Cheng, 1990).

$$Vertical\ Stiffness = \frac{\Delta y_{cm}}{Peak\ Vertical\ Force} \qquad Eq.\ 1$$

Leg stiffness was defined as the ratio of peak vertical force to the change in leg length,

defined by Equations 2 and 3 (Dutto & Smith, 2002) Equation 2 gives the change in leg

length used in the equation to calculate leg stiffness, Equation 3.

$$\Delta L = \Delta y_{cm} + L_0(1 - cos\emptyset) \qquad Eq.\ 2$$

In this equation, L_0 is the resting leg length and $\emptyset$ is the angle of the leg at initial contact

relative to the vertical. The change in leg length used in both Equations 2 and 3 is ΔL.

$$Leg\ Stiffness = \frac{Peak\ Vertical\ Force}{\Delta L} \qquad Eq.\ 3$$

EMG signals were bandpass filtered (20-500 Hz) (Madigan & Pidcoe, 2003) and

EMGworks 4.1 (Delsys, Boston, MA, USA) was used to calculate the mean frequency

and integrated EMG (iEMG) during the pre- and post- exercise isometric contractions.

3.6 Statistical analysis

The kinematic variables that were assessed for significance were the minimum,

maximum, and initial value as well as overall excursion of the sagittal ankle and knee

angles, tibial internal-external rotation, and forefoot-rearfoot inversion-eversion, as well

as ground contact time. The kinetic variables of active and impact peaks, loading rate,

vertical and leg stiffness were all assessed for significance. Finally, pre- and post-run

strength measures, mean frequency, and iEMG were assessed for the plantarflexor

muscles.

For each variable, significance between the first and last minute of the run were tested for significance using a one-way, paired sample repeated measures Student's t test ($\alpha=0.05$). For each measure that showed significance, a linear mixed effects model was applied accounting for the percent time of the run and participant number to test for non-zero changes in slope over time. Significance was tested for the effect of change in time. Effect sizes were calculated to determine if any differences found between trials or pre- and post-run measures were biologically meaningful. An effect size (*d*) greater than 0.3 indicated a small effect, greater than 0.5 a moderate effect, and an effect size larger than 0.8 indicated a large effect (Cohen, 1992).

3.7 Summary

The purpose of this study is to determine whether or not a prolonged run will cause a shift in the forefoot strike pattern of an experienced forefoot runner during stance to that more similar to a rearfoot strike pattern. To accomplish this, kinetic, kinematic, and EMG data will be collected and compared for each stage of a high intensity prolonged run. The prolonged run will be performed on an instrumented treadmill, set to a predetermined speed based off of each participant's most recent race performances. If the change in strike pattern does exist, then this will add to the literature that at varying levels of fatigue, runners will alter their stride pattern in order to adjust for the continued demands on the body.

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CHAPTER 4

DOES FOOTFALL PATTERN IN FOREFOOT RUNNERS CHANGE OVER A PROLONGED RUN?

ABSTRACT

There has been much debate on the benefits of a forefoot versus rearfoot strike pattern in distance running in terms of economy, competitive performance, and injury prevalence. Shock attenuation occurs more prominently in soft tissues at impact in forefoot runners compared to the more passive skeletal loading in rearfoot runners. Recent studies have shown that a forefoot strike pattern may not be maintainable over long distance efforts. Therefore, this study tested the hypothesis that habitual forefoot runners could not maintain their strike pattern throughout a prolonged, intensive run.

The prolonged run was performed to voluntary exhaustion in fourteen forefoot runners on an instrumented force treadmill. The average duration of the run was 15.4 ± 2.2 minutes. Kinematic and kinetic data were sampled each minute at 200Hz and 1000Hz, respectively. Ankle plantar-flexor torque was measured during pre- and post-run isometric contractions, during which electromyographic activity was measured in the soleus, lateral, and medial gastrocnemius.

Loading rate (49.95 ± 14.83 to 57.40 ± 22.53 BW*s^{-1}, p=0.0311) and impact peak (1.35 ± 0.43 to 1.50 ± 0.51, p=0.0207) increased significantly throughout the run while there was no significant change in active peak (p=0.08). Both peak knee flexion (-33.93 ± 3.67° to -36.21 ± 3.48°, p=<0.0000) and sagittal ankle angle at touchdown (-11.83 ± 5.33° to -9.33 ± 6.29°, p =0.0202) increased significantly. Ankle torque decreased significantly from pre- to post-run (120.57 ± 33.57 to 110.76 ± 32.91 Nm, p = 0.0154). This was accompanied by a decrease in medial and lateral gastrocnemius integrated electromyographic activity (iEMG) (p=0.0387 and 0.0186, respectively).

The results indicated that there were significant changes in landing mechanics in the habitual forefoot runners with increased levels of exertion, as they shifted towards strike patterns more similar to rearfoot runners throughout the run. These changes are in line with metabolic findings of other studies. There is increased eccentric loading of the ankle plantar-flexor muscles at touchdown in forefoot runners that may contribute to a decreased torque output by the end of the run. The decline in iEMG may indicate altered central drive of the system and a decline in the impact attenuation ability of the triceps surae, leading to the changes exhibited up the kinematic chain. These findings suggest that while forefoot strike patterns are good for speed, the onset of fatigue may affect the ability to maintain this pattern during a prolonged, intensive effort.

4.1 Introduction

Running is one of the most common forms of exercise due to its ease of access, low cost, and beneficial health effects. In the United States alone, there are more than thirty million active runners for competition and recreation (Novacheck, 1998). Three distinct footstrike patterns exist in these runners: forefoot (FF), midfoot (MF), and rearfoot (RF) strikes (Hasegawa et al., 2007). These patterns are based upon the foot position at touchdown and the subsequent foot movement throughout the stance phase (Hasegawa et al., 2007). When initial touchdown occurs with the anterior half of the foot, and the heel remains elevated throughout stance, a FF strike occurs. If the same touchdown conditions occur, but the heel makes ground contact throughout the stance phase, or if the foot touches down flat, it is defined as a MF strike. A RF strike occurs when the heel or posterior third of the foot makes the initial ground contact. The latter group makes up approximately 80 percent of all runners, while the remaining runners predominantly utilize a MF pattern (Hasegawa et al., 2007; Kerr et al., 1983). As few as 1.4 percent of all runners have been reported as "true" FF runners (Hasegawa et al., 2007).

Even though the percentage of FF runners in distance events is low, the greatest percentage of FF runners has been found in the fastest-finishing groups (Hasegawa et al., 2007; Hayes & Caplan, 2012). Studies have also found that runners make contact on the more anterior aspects of the foot with increases in running speed (Hasegawa et al., 2007; Kerr et al., 1983; Novacheck, 1998; Payne, 1983). Due to these performance differences in footstrike groups, it has been suggested that FF running is more economical and that more runners should train and adapt this running style (Hasegawa et al., 2007; Perl et al., 2012). However, recent research has suggested that FF running has a similar or even

decreased economy compared to RF running (Gruber, Umberger, Braun, & Hamill, 2013; Ogueta-Alday, Rodríguez-Marroyo, & García-López, 2013). When sub-elite runners were studied, RF runners were shown to be greater than five percent more economical than FF runners across a range of submaximal speeds (Ogueta-Alday et al., 2013). Given some of the current perceptions regarding performance and strike pattern, this sub-elite group of runners should have shown the greatest benefits of a FF strike pattern as this population consists of a greater percentage of FF runners than the general population. Thus, the decline in economy for even this population is important to note when considering strike pattern benefits.

Due to the relatively uncommon nature of FF running, most of the literature involving running mechanics has focused primarily on RF running (Cavanagh & Lafortune, 1980; Hasegawa et al., 2007; Novacheck, 1998; Williams & Cavanagh, 1987). These previous studies have focused on understanding the forces and joint coordination at play during normal, healthy running, as well as both pathological gait and the effects of fatigue. However, differences have been shown between footfall patterns in both kinematics and kinetics. Further study of FF running is appropriate to create a more complete understanding of running mechanics. Forefoot runners exhibit shorter ground contact times and a more rapid transition to toe-off within the gait cycle than RF runners (Hasegawa et al., 2007; Hayes & Caplan, 2012; Novacheck, 1998). Forefoot runners also exhibit decreased peak dorsiflexion and increased peak plantarflexion throughout stance (Hasegawa et al., 2007). The vertical ground reaction force profile of FF runners is marked by a decreased loading rate and poorly defined impact peak when compared to RF runners, as well as by the presence of an increased active peak (Dickinson et al.,

1985; Laughton et al., 2003; Oakley & Pratt, 1988). The active peak marks the peak component of the vertical ground reaction force associated with active mechanisms of force production, while the impact peak is the passive transient force experienced in the first 30 percent of stance (Novacheck, 1998). The loading rate is the rate at which the force is applied in early stance (Novacheck, 1998). The ankle in FF runners is forced to behave as both a shock absorber at impact and a propulsive force generator at toe-off (Dickinson et al., 1985). This extra demand placed on the ankle may cause greater shock attenuation to occur in the soft tissues of the lower extremity, which may be accompanied by altered loading over time. Injuries may result from these attenuation strategies and altered loading (Mizrahi et al., 2000b). This effect could be magnified during periods of increased fatigue, which are frequently accompanied by loss of joint and muscular coordination leading to changes in stride mechanics (Dierks et al., 2010; Hayes et al., 2004; Mizrahi et al., 2000b).

During the running cycle, the lower extremities are subjected to multiple submaximal loads with each successive footstrike. Repeated stretch-shorten cycling of the elastic elements of the triceps surae and Achilles' tendon can be affected by the exaggerated eccentric loading that occurs at touchdown in FF runners, which leads to a diminished ability to both attenuate and produce force at impact (Gollhofer et al., 2008; Komi, 2000). The energy return of these elastic elements may become diminished over time, subjecting the tissues to greater impact stresses and leading to an increased metabolic cost to maintain velocity and mechanics. Over the course of several thousand footfalls in a distance run, this effect is greatly magnified and may lead to gait adaptations to account for the changing loads placed on the lower extremity.

The current state of running lends itself to constant changes in footwear designs and styles, training methodologies, and new techniques and products touted by the lay literature to enhance performance, improve economy, and reduce injury risk. However, many of these claims cannot be back up by scientific research. This is especially true of running form suggestions, and more specifically, on the most desirable foot strike patterns to employ. The recent fad in the running community touts FF and minimalist running as the latest and greatest advancement in running performance. On the contrary, several studies have indicated that injury risk is not reduced in FF runners, and that economy is at best equivalent to that of RF runners (Gruber et al., 2013; Ogueta-Alday et al., 2013). Furthermore, numerous studies have indicated that runners who begin races utilizing a FF pattern are unable to maintain it for the duration of the race (Kerr et al., 1983; Larson et al., 2011). If these runners are switching to a more posterior strike pattern as they become more tired, then why focus on running with a pattern that is possibly not maintainable for these longer efforts?

The purpose of this study was to investigate the effects of a prolonged, intensive run on FF runners on muscle fatigue, kinematics, and vertical ground reaction forces. The hypotheses tested were: 1) strength of the ankle plantar flexors will decrease from the beginning to end of the run, and are associated with decreased mean electromyographic (EMG) frequency and decreased integrated EMG (iEMG) ; 2) there will be significant kinematic changes in the sagittal ankle and knee angles, tibial internal-external rotation, and forefoot-rearfoot eversion-inversion throughout the run, indicating a shift from a FF to a RF strike pattern, as well as an increase in ground contact time; and 3) there will be a decreased active peak and loading rate in the vertical ground reaction

force component, as well as an increased impact peak, increased vertical stiffness, and increased leg stiffness from the beginning to the end of the run. These hypotheses all contribute to the overall belief that foot strike pattern will change from a FF to a RF pattern by the end of the run.

4.2 Methodology

4.2.1 Participant Selection

A sample size estimation was calculated from previous research, and it was determined that to achieve appropriate power ($\beta = 0.80$), 14 participants were necessary (Christina et al., 2001; Williams et al., 2000). Fourteen healthy, natural FF runners participated in this study (8 male, 6 female) (Table 4.1). Participants were required to run at least 15 miles (24 km) per week. Participants were excluded from the study if they had any history of cardiovascular or neurological problems or lower extremity injury or surgery in the previous year. Each participant was screened to ensure they utilized a FF pattern both overground and on an instrumented treadmill via vertical ground reaction force measurements further validated by high-speed motion analysis. Participants were included once determination had been made that their ankles were plantarflexed at touchdown, and that there was little or no visible impact peak present in the vertical ground reaction force. All participants read and completed an informed consent document and questionnaires approved by the University of Massachusetts Amherst Institutional Review Board prior to participating.

Table 4.1: Mean ± SD participant characteristics.

Age (years)	Height (m)	Mass (kg)	Treadmill Running Speed (m/s)	Miles/Week	Time to exhaustion (min)
26.5 ± 8.6	1.73 ± 0.1	67.5 ± 7.8	3.89 ± 0.6	31.1 ± 15.8	15.4±2.2

4.2.2 Experimental setup

All data collection for this study was performed in the University of Massachusetts Biomechanics Lab. All participants wore a neutral, low-profile shoe with minimal heel raise, generally classified as a racing flat. Unilateral, three-dimensional kinematic data was collected using an eight-camera Qualysis ProReflex motion capture system (Qualysis, Inc., Gothenberg, Sweden) sampling at 200 Hz. On each participant, 31 retro-reflective markers were placed to track segment motion. Markers were placed on the right foot, leg, and pelvis. The pelvic segment was defined by markers on the iliac crests, greater trochanters, anterior superior iliac spines (ASIS), and the sacrum. Lateral and medial knee and ankle markers were used to define lower extremity segments, with tracking plates of four non-colinear markers placed on the thigh and the shank. A modified multi-segment foot model was used, with a heel cluster, toe marker, and markers on the first, second, and fifth metatarsal heads and bases (Leardini et al., 2007; Myers et al., 2004). The collection space was calibrated prior to each data collection.

The cameras surrounded an instrumented treadmill (Athletics Republic, Park City, UT), which rests on four tri-axial load cells (AMTI, Inc., Watertown, MA, USA). Resultant forces were calculated by summing together the output from each load cell, which were sampled at 1000 Hz. Isometric strength data were collected pre- and post-run using a Humac Norm dynamometer (CSMi, Stoughton, MA, USA).

Electromyography data were collected using a Trigno Wireless EMG system (Delsys, Boston, MA, USA). Each sensor has a mass of less than 15 grams and records both EMG and tri-axial accelerations. Seven muscles were collected unilaterally: the tibialis anterior, medial and lateral gastrocnemius, soleus, rectus femoris, vastus lateralis,

and biceps femoris. Prior to the placement of each electrode, each site was shaved, abraded, and cleaned to ensure an accurate, clean signal. EMG data were sampled at 2000 Hz, and synchronization between kinematic, kinetic, and EMG collection was performed utilizing an external trigger to simultaneously start all data collection.

4.2.3 Protocol

This study was performed in one visit. Participants arrived in the lab, went through the informed consent process, and then went through the screening process. Once footfall pattern was confirmed, participants were fitted with EMG electrodes. All signals were checked for quality, and then maximum voluntary contractions (MVC) were performed for each muscle to allow for signal normalization. Next, participants completed pre-strength testing consisting of 3x5 second isometric maximal plantar flexor contractions with the ankle in a neutral, 90º position. Markers were then placed on the participant, and they were given a period to warm up as needed. After the warm up period, the speed for the run was confirmed as determined during the screening based upon previous race performances and training paces. Speed was confirmed as one that "could be maintained for 15 minutes, but not more than 20". After the warm-up period, participants completed a standing calibration, and then were ramped up to the prolonged bout speed. Data were collected for 15-second windows every minute of the run. The rate of perceived exertion (RPE) of each participant was asked every other collection period (Borg, 1971). All participants gave a verbal indication of RPE between 18-19 (on a scale of 6-20) prior to voluntary termination of the prolonged run, indicating that all participants reached similar end stage levels of effort. Upon voluntary termination of the

63

run by the participant, post-run strength testing was conducted in the same manner as the pre-run test.

4.2.4 Data Reduction

Kinematic data were tracked and labeled in Qualysis Track Manager and exported to .c3d files (Qualysis, Inc., Gothenburg, Sweden). Marker data were filtered using a dual-pass, low pass Butterworth filter with a cutoff frequency of 12 Hz. Sagittal ankle and knee angles, tibial internal-external rotations, and forefoot-rearfoot eversion inversion were calculated using a Cardan Xyz rotation sequence in Visual 3D (C-Motion Inc., Germantown, MD). The data were interpolated, normalized to 101 data points, and tracked over five consecutive strides for each collection period of the experiment. Individual outcome measures were calculated for each stride, and then a resultant mean was calculated to represent the participant at each time point in the run.

Vertical kinetic data were analyzed in both Visual 3D and a custom designed Matlab program (Mathworks, Natick, MA). Peak impact and active forces, as well as loading rate, were calculated (Christina et al., 2001). In FF runners with no distinct impact peak, relative impact phase magnitude were calculated at similar time points to RF events. The kinetic data were interpolated and normalized to 101 data points for each stance phase to allow accurate comparison between separate foot strikes and separate collection periods. A low pass, 4th order zero-phase lag Butterworth filter with a 30 Hz cutoff frequency was used to filter the data (Laughton et al., 2003). Vertical stiffness was calculated as the ratio between maximum displacement of the center of mass and peak vertical force (Eq. 1) (Dutto & Smith, 2002; McMahon & Cheng, 1990).

$$Vertical\ Stiffness = \frac{\Delta y_{cm}}{Peak\ Vertical\ Force} \qquad\qquad Eq.\ 1$$

Leg stiffness was defined as the ratio of peak vertical force to the change in leg length, defined by Equations 2 and 3 (Dutto & Smith, 2002) Equation 2 gives the change in leg length used in the equation to calculate leg stiffness, Equation 3.

$$\Delta L = \Delta y_{cm} + L_0(1 - cos\phi) \qquad \text{Eq. 2}$$

In this equation, L_0 is the resting leg length and ϕ is the angle of the leg at initial contact relative to the vertical. The change in leg length used in both Equations 2 and 3 is ΔL.

$$Leg\ Stiffness = \frac{Peak\ Vertical\ Force}{\Delta L} \qquad \text{Eq. 3}$$

EMG signals were bandpass filtered (20-500 Hz) (Madigan & Pidcoe, 2003) and EMGworks 4.1 (Delsys, Boston, MA, USA) was used to calculate the mean frequency and integrated EMG (iEMG) during the pre- and post- exercise isometric contractions.

4.2.5 Statistical Analysis

The kinematic variables that were assessed for significance were the minimum, maximum, and initial value at footstrike as well as overall excursion of the sagittal ankle and knee angles, tibial internal-external rotation, and forefoot-rearfoot inversion-eversion. Ground contact time was also measured between touchdown and toe-off. The kinetic variables of active and impact peaks, loading rate, vertical and leg stiffness were all assessed for significance. Finally, pre- and post-run strength measures, mean frequency, and iEMG were assessed for the plantar flexor muscles.

For each variable, significance between the first and last minute of the run were tested for significance using a one-way, paired sample, repeated measures Student's t-test (α=0.05). For each significant measure, a linear mixed effects regression model was applied. Time point of the run was defined as a fixed effect in the model, while subject differences were designated as a random factor. Two statistical tests were performed

65

using the model. First, a Chi-square test was used to determine whether the time point in

the run had a significant effect on the outcome measure being modeled. Next, a T-test

was performed to determine whether or not differences found by time demonstrated a

significant non-zero slope over the course of the run. Both tests are needed to identify

the relationship between time and outcome measure. It would be possible for time to

have a significant effect on each measure tested, and not show a trend throughout the

entire run; this would return a significant Chi-square test but not a significant test for

non-zero slope. Effect sizes were calculated to determine if any differences found

between trials or pre- and post-run measures were biologically meaningful. An effect

size (d) greater than 0.3 indicated a small effect, greater than 0.5 a moderate effect, and

an effect size larger than 0.8 indicated a large effect (Cohen, 1992).

4.3 Results

4.3.1 Kinematics

4.3.1.1 Sagittal Ankle Angle

The initial sagittal ankle angle at foot strike was significantly less plantar flexed at

the end of the run, while the angle at toe-off, or peak plantar flexion, was significantly

more plantar flexed ($\rho= 0.0202$ and 0.0399, respectively) (Table 4.2). A time series of a

representative subject indicating the trend shift in sagittal ankle angle is presented in

Figure 4.1. The linear mixed effects regression model showed no significant effect of

time point in the run to changes in the initial and peak plantar flexion sagittal ankle angle

values. The data did not show a significant, non-zero slope over run time point for the

variables determined to be significant between the first and last trials. The peak plantar

flexion angle effect size was less than 0.3, indicating poor biological significance, while

there was a small effect for the initial ankle angle (ES = 0.430).

Table 4.2: Mean ± SD for initial, maximum, angle at toe-off, and excursion of sagittal ankle angle. Differences between beginning and end of run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

Sagittal Ankle Angle (Degrees)	Beginning of Run Mean±SD	End of Run Mean±SD	ρ-value	Effect size
Initial (angle at footstrike)	-11.83±5.33	-9.33±6.29	0.0202*	0.430
Maximum (Peak Dorsiflexion)	14.90±5.27	14.31±4.50	0.1345	0.121
Angle at toe-off (peak plantar flexion)	-24.94±5.42	-26.15±5.05	0.0399*	0.231
Excursion	39.84±3.91	40.87±5.33	0.1492	0.223

Table 4.3: Sagittal ankle angle linear mixed effects model results. Chi-Square value and significance test (α = 0.05) are for the effect of run time point on the measured variable. Slope of Time Point defines the slope for the given measure with respect to the time point of the run, and the significance test (α = 0.05) is to test the difference from 0.

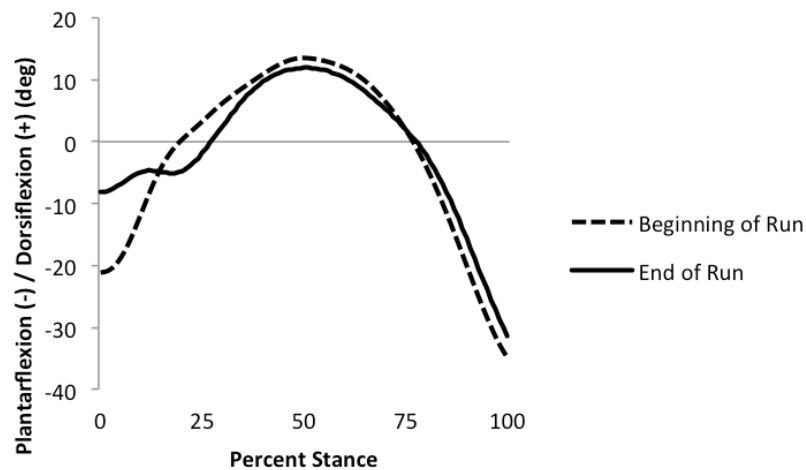| Linear Mixed Effects Model - Effect of Percent Run – Ankle Angle				
	Chi-Square value	p-value of Chi-Square	Slope of Trial (Degrees)	p-value
Toe-Off	0.1877	0.6648	-0.092	0.6533
Initial	0.3225	0.5701	0.1201	0.3027

Figure 4.1: Sagittal ankle time series. Time series of sagittal ankle profile from beginning (dashed) to end (solid) of the run for a representative participant. Plantar flexion is represented by negative angle values, and dorsiflexion is positive.

4.3.1.2 Sagittal Knee Angle

Minimum value, or peak flexion, and excursion of the sagittal knee angle were found to be significant ($\rho=$ <0.0000 and 0.0081) between the beginning and end of the run. Both had moderate effect sizes of 0.638 and 0.707, respectively. A scatterplot of all data for peak knee flexion is shown in Figure 4.2. A time series of the sagittal knee angle over stance for a representative participant is presented in Figure 4.3. From the beginning to the end of the run, the participants had greater knee flexion at toe-off, and peak flexion. The linear mixed effects regression model showed significant effects of time on the minimum value and excursion of the sagittal knee angle. Time affected minimum knee angle ($\chi^2(1)$=5.866, p=0.0154), lowering it by 0.28º per time point of the run. Each successive run time point increased excursion by .20º ($\chi^2(1)$=3.901, p=0.0483). Only the slope of the linear fit for the minimum value was significantly different from zero (p=0.0356).

Table 4.4: Mean ± SD for initial, maximum, minimum, and excursion of sagittal knee angle. Differences between beginning and end of run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

Sagittal Knee Angle (Degrees)	Beginning of Run Mean±SD	End of Run Mean±SD	ρ-value	Effect size
Initial (angle at foot strike)	-9.45±4.58	-10.43±4.09	0.1440	0.226
Maximum	-6.27±2.48	-5.99±2.93	0.3792	0.104
Minimum (Peak knee flexion)	-33.93±3.67	-36.21±3.48	<0.0000*	0.638
Excursion	27.66±3.71	30.22±3.53	0.0081*	0.707

Table 4.5: Sagittal knee angle mixed effects model results. Chi-Square value and significance test (α = 0.05) are for the effect of run time point on the measured variable. Slope of Time Point defines the slope for the given measure with respect to the time point of the run, and the significance test (α = 0.05) is to test the difference from 0.

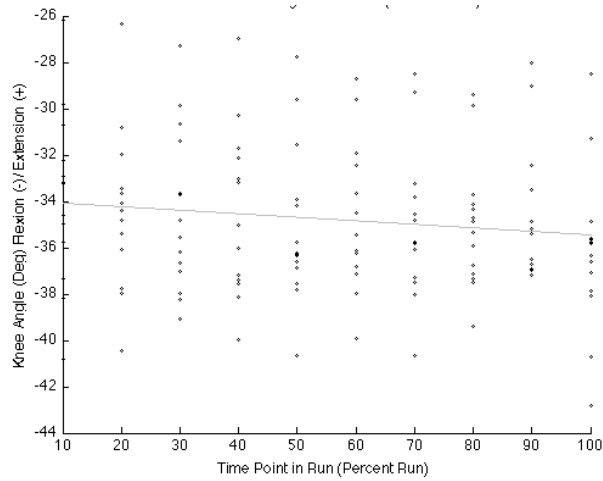| Linear Mixed Effects Model - Effect of Percent Run – Knee Angle				
	Chi-Square value	p-value of Chi-Square	Slope of Trial (Degrees)	p-value
Minimum	5.866	0.0154*	-0.2812	0.0356*
Excursion	3.901	0.0483*	0.1951	0.0593

Figure 4.2: Scatter plot of peak knee flexion versus time point of the run. The gray line is a best fit of the data, representing the slope from the linear model.
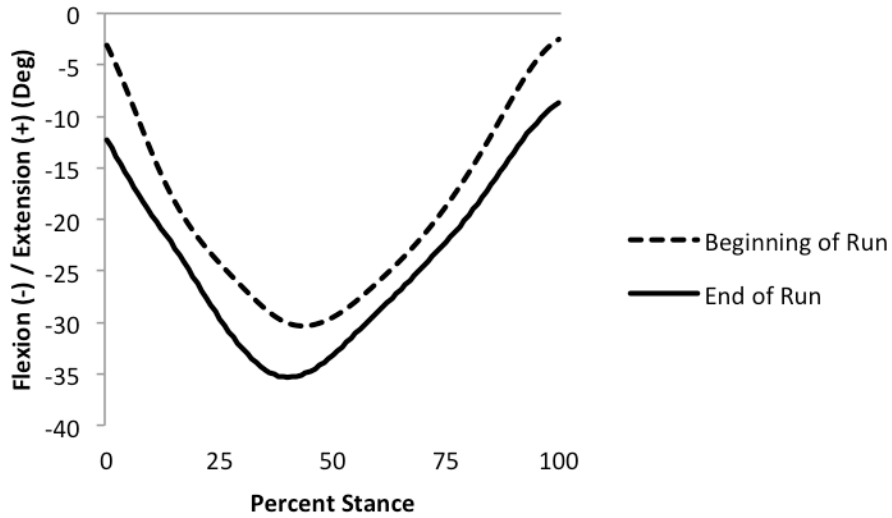

Figure 4.3: Sagittal knee time series. Time series of sagittal knee profile from beginning (dashed) to end (solid) of the run for a representative participant. Flexion is represented by negative angle values, and extension is positive.

4.3.1.3 Tibial Internal-External Rotation

For the tibial internal-external rotation measures, only the maximum angle (peak internal rotation) and the excursion were found to be significantly different from the beginning to the end of the run (p=0.0011 and 0.0017) (Table 4.6). Both had moderate effect sizes of 0.596 and 0.563, respectively. All data for peak internal rotation are displayed in Figure 4.4, and all data for excursion are in Figure 4.5. A time series for a representative participant is displayed in Figure 4.6. Both measures were found to have a significant time effect, as determined by the linear mixed effects regression model (Table 4.7). Run time point affected the peak internal rotation angle ($\chi^2(1)$=29.977, p= <0.0000) increasing it 0.1965° per time point. The same was true of the excursion of the internal-external rotations ($\chi^2(1)$=39.665, p=<0.0000), causing an overall increase in excursion of 0.2863° per time point. The fit of the model with both data sets was found to be significantly non-zero (p = 0.0022 and 0.0122).

Table 4.6: Mean ± SD for initial, maximum, minimum, and excursion of tibial internal-external rotation. Differences between beginning and end of run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

Tibial Internal-External Rotation (Degrees)	Beginning of Run Mean±SD	End of Run Mean±SD	ρ-value	Effect size
Initial (angle at foot strike)	-7.99±5.40	-7.69±5.23	0.3189	0.056
Maximum (peak internal rotation)	8.08±3.96	10.32±3.56	0.0011*	0.596
Minimum (peak external rotation)	-9.09±5.27	-9.90±4.98	0.1325	0.158
Excursion	17.17±4.62	20.22±6.21	0.0017*	0.563

Table 4.7: Tibial internal-external rotation linear mixed effects model results. Chi-Square value and significance test (α = 0.05) are for the effect of run time point on the measured variable. Slope of Time Point defines the slope for the given measure with respect to the time point of the run, and the significance test (α = 0.05) is to test the difference from 0.

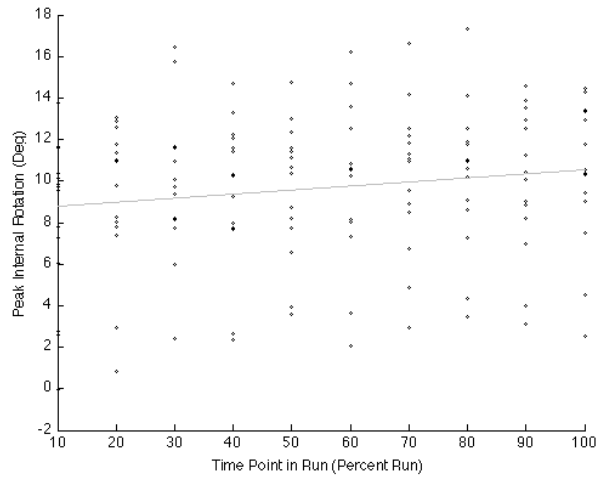| Linear Mixed Effects Model - Effect of Percent Run – Tibial Internal/External Rotation				
	Chi-Square value	p-value of Chi-Square	Slope of Trial (Degrees)	p-value
Maximum	29.977	<0.0000*	0.1965	0.0022*
Excursion	39.665	<0.0000*	0.2863	0.0012*

Figure 4.4: Peak tibial internal rotation versus time scatterplot. The gray line is a best fit of the data, representing the slope from the linear model.
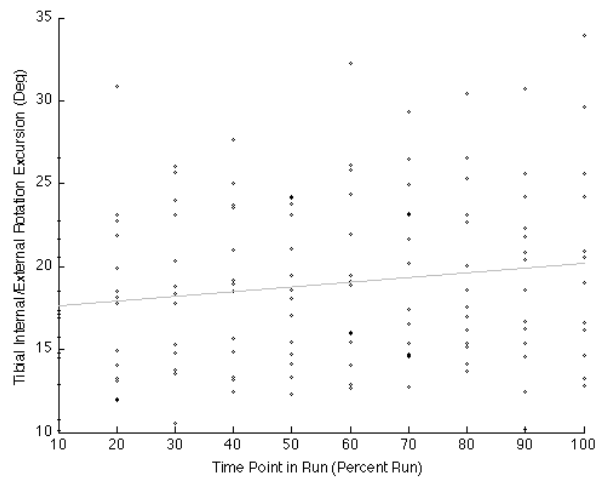

Figure 4.5: Tibial rotation excursion versus time scatterplot. The gray line is a best fit of the data, representing the slope from the linear model.
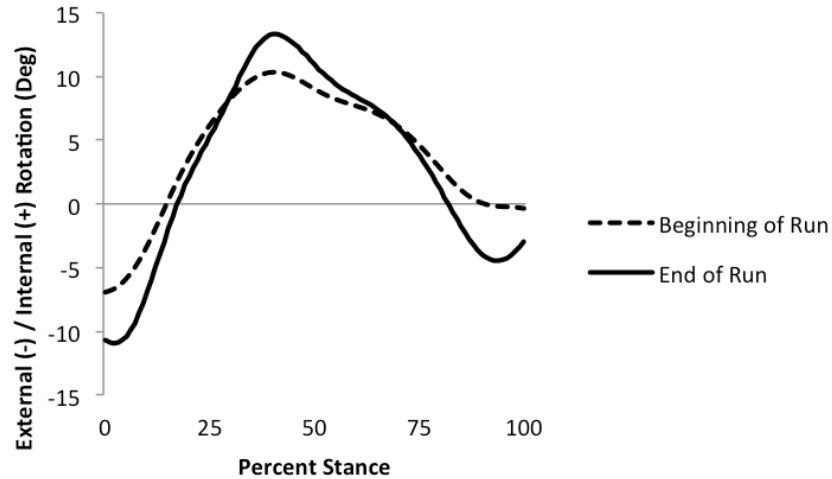
71

Figure 4.6: Tibial rotation time series. Time series of tibial internal-external rotation from beginning (dashed) to end (solid) of the run for a representative participant. External rotation is represented by negative angle values, and internal rotation is positive.

4.3.1.4 Forefoot-Rearfoot Inversion-Eversion

Forefoot-rearfoot inversion-eversion angle minimum (peak inversion) and excursion were both found to be significant between the beginning and end of the run (p = 0.0080 and 0.0261) (Table 4.8). The effect size of the angle excursion was small (ES = 0.397) while the minimum angle effect size was only 0.166. There were no significant findings of the effect of run time point on these measures determined by the linear mixed effects regression model (Table 4.9). A time series of forefoot-rearfoot inversion-eversion angles is presented in Figure 4.7 for a representative participant.

Table 4.8: Mean ± SD for initial, maximum, minimum, and excursion of forefoot-rearfoot inversion-eversion. Differences between beginning and end of run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

Forefoot-Rearfoot Inversion-Eversion (Degrees)	Beginning of Run Mean±SD	End of Run Mean±SD	ρ-value	Effect size
Initial (at foot strike)	0.26±2.05	0.12±1.94	0.0984	0.070
Maximum (peak eversion)	1.42±1.81	1.35±1.86	0.2339	0.038
Minimum (peak inversion)	-1.15±2.13	-1.50±2.09	0.0080*	0.166
Excursion	2.57±0.69	2.85±0.72	0.0261*	0.397

Table 4.9: Forefoot-rearfoot inversion-eversion angle linear mixed effects model results. Chi-Square value and significance test (α = 0.05) are for the effect of run time point on the measured variable. Slope of Time Point defines the slope for the given measure with respect to the time point of the run, and the significance test (α = 0.05) is to test the difference from 0.

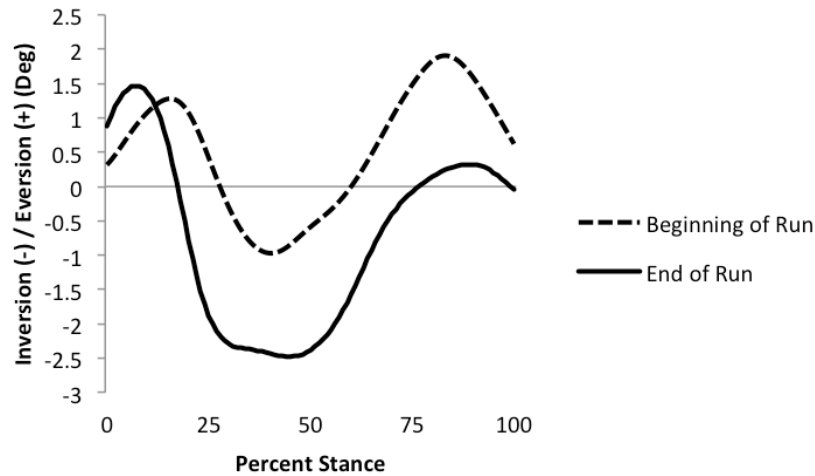| Linear Mixed Effects Model - Effect of Percent Run – Forefoot/Rearfoot Inversion-Eversion				
	Chi-Square value	p-value of Chi-Square	Slope of Trial (Degrees)	p-value
Minimum	0.1451	0.7032	-0.0472	0.3651
Excursion	0.0634	0.8012	-0.0073	0.4121

Figure 4.7: Forefoot-rearfoot inversion/eversion time series. Time series of forefoot-rearfoot inversion-eversion from beginning (dashed) to end (solid) of the run for a representative participant. Inversion is negative angle values, and eversion is positive.

4.3.1.5 Ground Contact Time

From the beginning to the end of the run, ground contact time, the duration of the stance phase, increased by 0.007 seconds from 0.220±0.022 to 0.227±0.022. This difference was significant (p = 0.0141). These results are presented, with standard deviation bars, in Figure 4.8. This effect size was small (ES = 0.318).
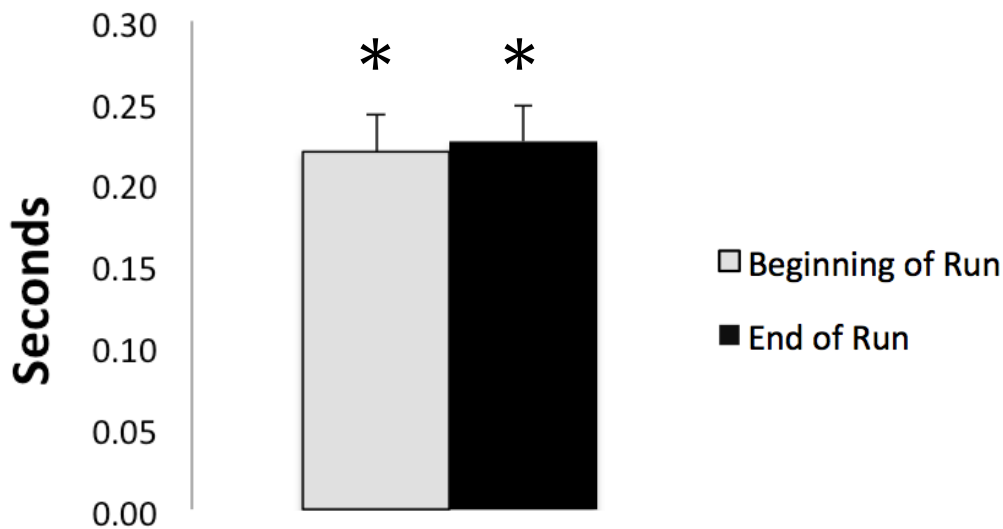
Figure 4.8: Pre- and post-run ground contact time. Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top. * = significance ($\alpha = 0.05$).

4.3.2 Kinetics

All of the force data presented in this paper were determined from the vertical ground reaction force profile for each participant. The profile for the beginning and end of run are presented, for a representative participant, in Figure 4.9. The curve for the beginning of the run is what would be expected for a FF runner, while the curve for the end of the run more closely resembles the vertical ground reaction force profile that would be exhibited by a RF runner. Seven of the fourteen participants showed qualitative changes in foot strike pattern over the course of the run, while the other participants remained relatively constant in strike pattern for the entire run. This was determined both by the sagittal ankle angle at touchdown, as well as the following kinetic results.
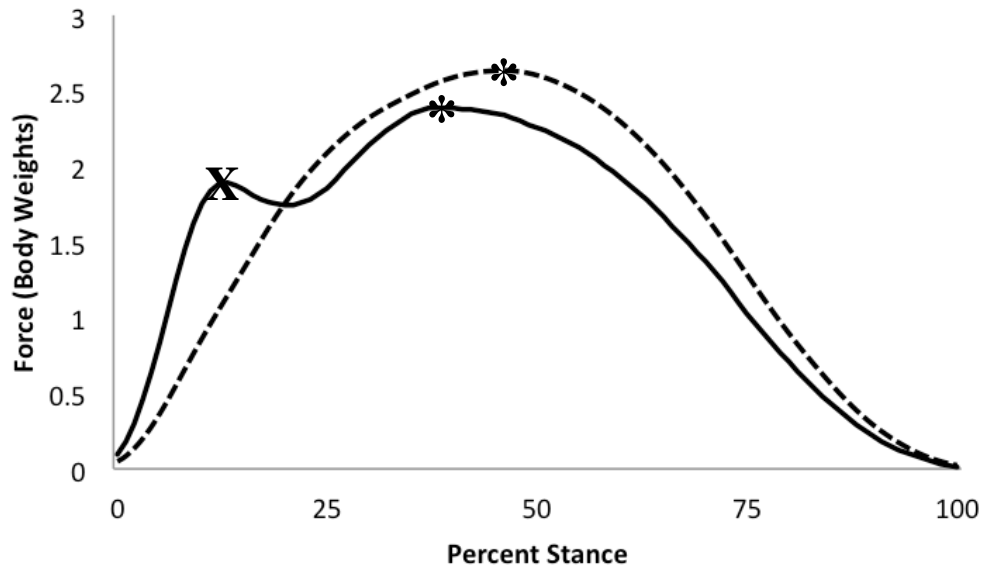
Figure 4.9: Vertical ground reaction force time seriesTime series of vertical ground reaction force from beginning (dashed) to end (solid) of the run for a representative participant. X marks the impact peak presence at the end of the run, and is indicative of a rearfoot strike pattern. The two *'s mark the active peak location.

No changes from beginning to end of run were found in the active peak, or the peak magnitude, of the vertical ground reaction force over the stance phase (2.61 ± 0.18 to 2.57 ± 0.23 BW, $p = 0.0830$, ES $= 0.195$) (Figure 4.10).
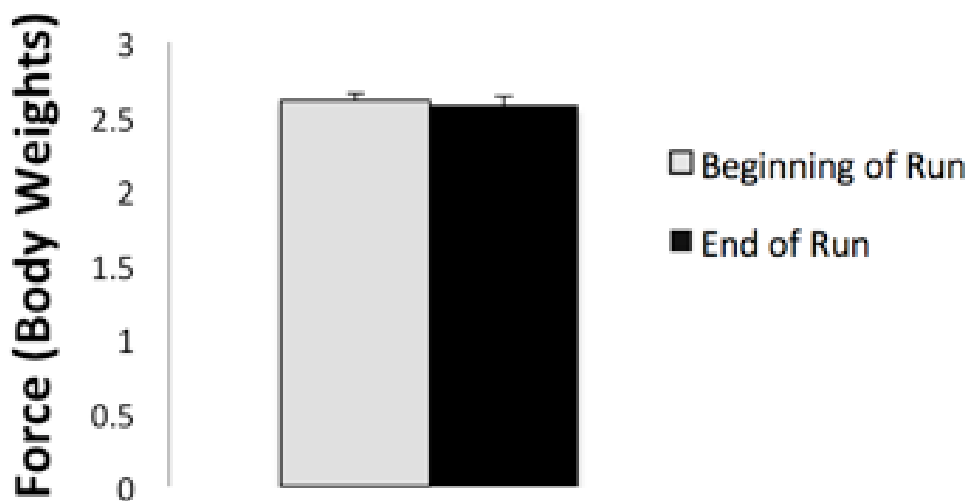

Figure 4.10: Pre- and post-run active peak magnitude. Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top.

Impact phase magnitude increased significantly from 1.35±0.43 to 1.50±0.51 body weights between the beginning and end of the run (p = 0.0207) (Figure 4.11). The impact phase effect size was small (ES = 0.319). When the linear mixed effects regression model was applied, it was determined that run time point had a significant effect on impact phase magnitude ($\chi^2(1)$=17.42, p=<0.000), causing an increase in impact phase magnitude of 0.0121 BW per time point. The slope of this relationship was found to be significantly different from zero (p = 0.0063). The slope is depicted in Figure 4.12 in a scatterplot with the data from each participant.
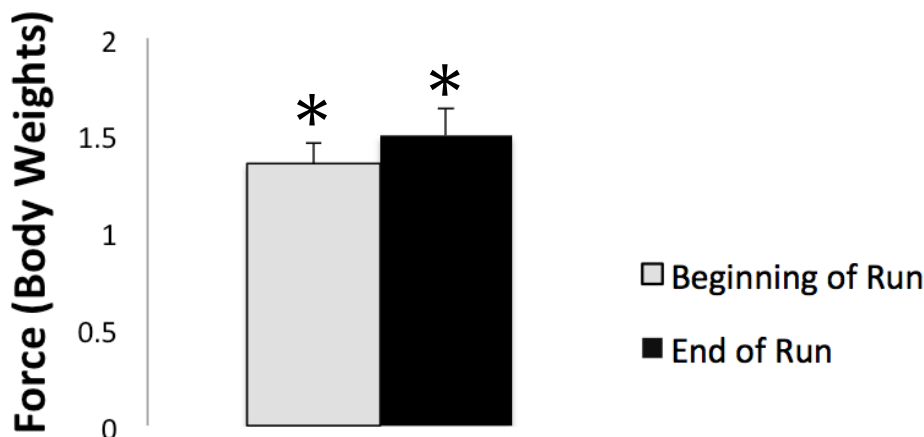

Figure 4.11: Pre- and post-run impact phase magnitude. Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top. * = significance (α = 0.05).

Loading rate changed from 49.95±14.83 to 57.40±22.53 BW*s^{-1} over the course of the run (p=0.0311). Loading rate effect size was small (ES = 0.399). When the linear mixed effects regression model was applied, time was determined to have a significant effect on loading rate ($\chi^2(1)$=16.863, p=<0.000), leading to an increased loading rate per time point by 0.562 BW*s^{-1}. The fit of the model with the data was found to be significantly nonzero (p=0.0067). Group means and standard deviations are presented in Figure 4.13. A scatterplot of the data, including the slope, is shown in Figure 4.14.
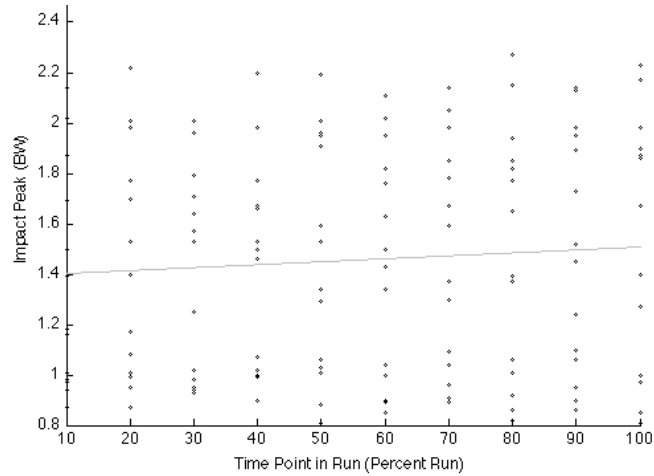
76

Figure 4.12: Impact phase magnitude versus time scatterplot. The gray line is a best fit of the data, representing the slope from the linear model.
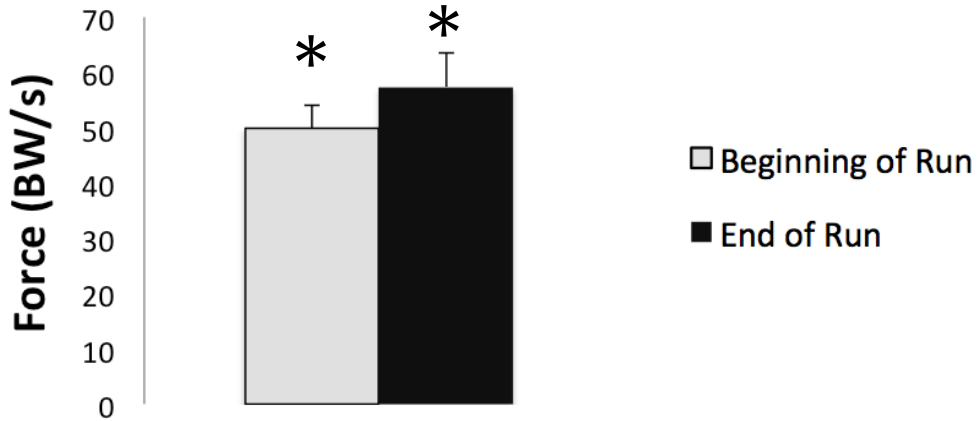

Figure 4.13: Pre- and post-run vertical loading rate. Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top. * = significance ($\alpha = 0.05$).
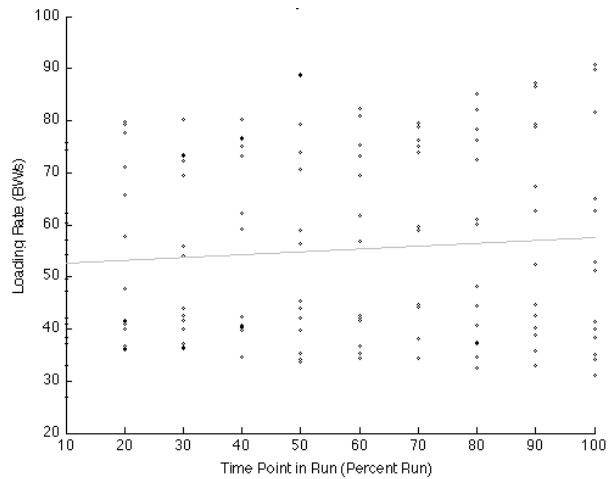

Figure 4.14: Vertical loading rate versus time scatterplot. The gray line is a best fit of the data, representing the slope from the linear model.

Between the beginning and end of the run, the vertical stiffness did not change significantly (Table 4.10). However, leg stiffness decreased significantly over the same time frame (p=0.0029). Means and standard deviations for the calculated stiffness values are presented in Figure 4.15. The effect size of leg stiffness was small (ES = 0.435).

Table 4.10: Mean ± SD of vertical and leg stiffness. Differences between beginning and end of run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

Stiffness Constant Mean±SD	Beginning of Run	End of Run	ρ-value	Effect size
Vertical Stiffness	34.35±6.19	33.56±5.72	0.1026	0.133
Leg Stiffness	14.47±1.70	13.73±1.70	0.0029*	0.435
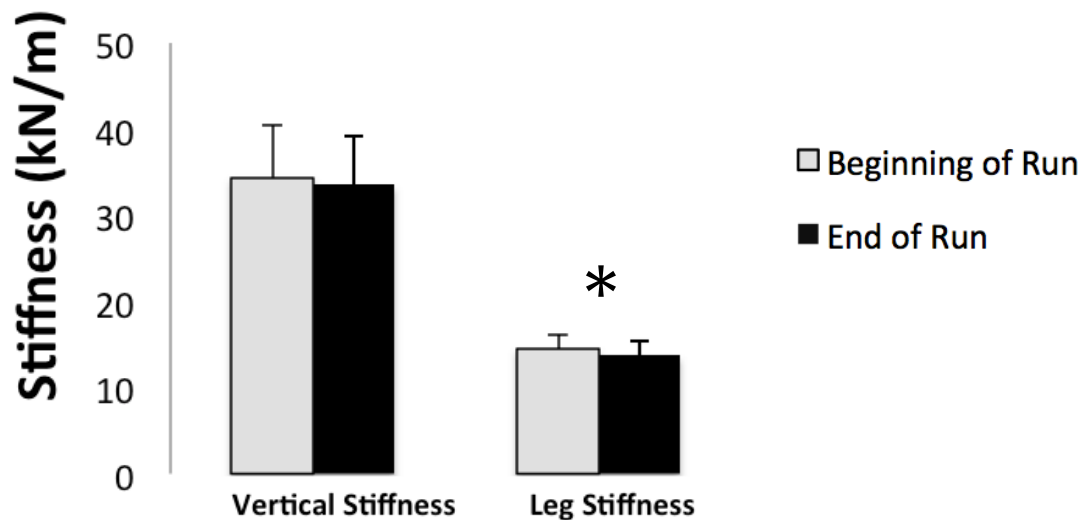
Figure 4.15: Pre- and post-run stiffness measures. Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top. * = significance (α = 0.05).

4.3.3 Strength and EMG

The mean ankle plantar flexion torque before the run was 120.57±33.57 Nm, while the end of the run torque decreased significantly to 110.76±32.91 (p = 0.0154). The effect size was small (ES = 0.300) and the results are displayed in Figure 4.16.
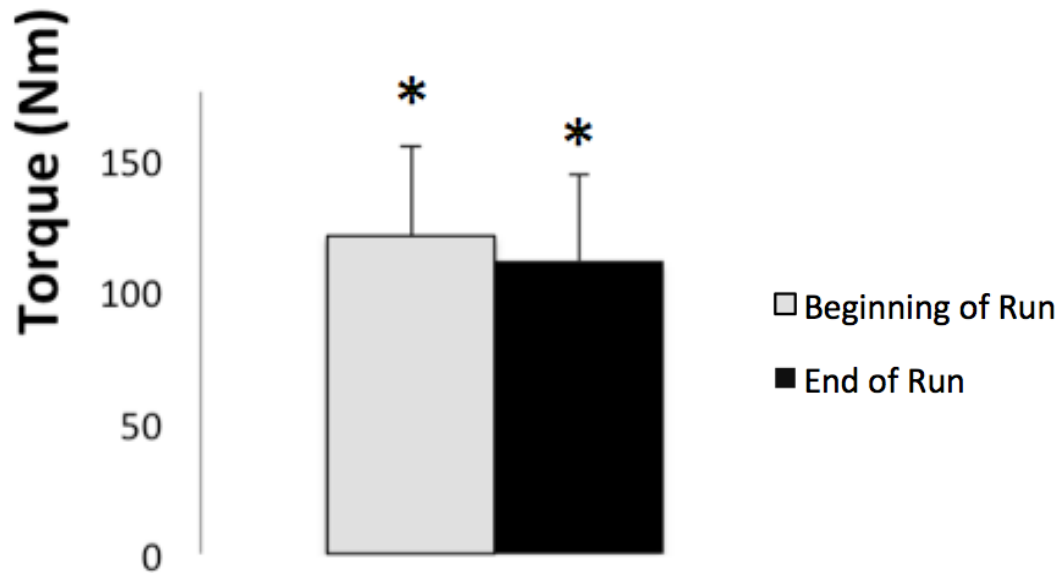
Figure 4.16: Pre- and post-run ankle plantar flexor torque. Light gray bar is pre-run, and black bar is post-run. Standard deviation bars are shown on top.   * = significance (α = 0.05).

There were no significant differences between pre- and post-run mean EMG frequency in the soleus, medial and later gastrocnemius muscles (Table 4.11, Figure 4.17). Integrated EMG for the medial and lateral gastrocnemius muscles were found to decrease significantly between the pre- and post-run measures (p = 0.0387 and 0.0186) (Table 4.12). The results are presented in Figure 4.18. Only the effect size for the lateral gastrocnemius was moderate (ES = 0.650).

Table 4.11: Mean ± SD of mean frequency of EMG signal for three ankle plantar flexor muscles. Differences between pre- and post-run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

| Mean Frequency	Pre - Mean±SD in Hz	Post - Mean±SD in Hz	ρ-value	Effect size
Soleus	133.95±19.43	138.37±18.62	0.2146	0.232
Medial Gastrocnemius	172.62±22.94	173.43±30.24	0.4393	0.030
Lateral Gastrocnemius	147.56±22.05	153.10±21.30	0.1721	0.256
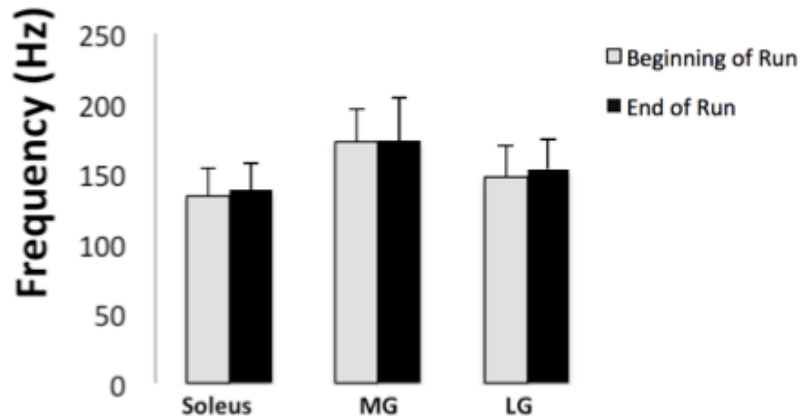
Figure 4.17: Pre- and post-run EMG mean frequency. Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top. * = significance (α = 0.05).

Integrated EMG	Pre - Mean±SD in mV	Post - Mean±SD in mV	ρ-value	Effect size
Soleus	0.0991±0.0412	0.0842±0.0434	0.0945	0.352
Medial Gastrocnemius	0.1803±0.0872	0.1597±0.0896	0.0387*	0.233
Lateral Gastrocnemius	0.1680±0.1141	0.1180±0.0397	0.0186*	0.650

Table 4.12: Mean ± SD of iEMG for three ankle plantarflexor muscles. Differences between pre- and post-run were assessed by a student's t-test (α = 0.05). * = significance. Effect size is defined by Cohen's *d*.

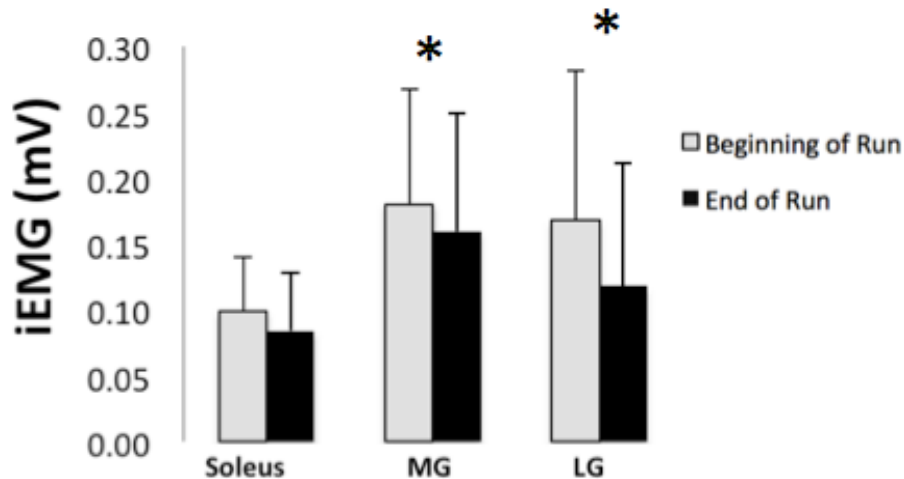**Figure 4.18: Pre- and Post-run iEMG of soleus, medial gastrocnemius (MG), and lateral gastrocnemius (LG). Light gray bar is beginning, and black bar is end. Standard deviation bars are shown on top. * = significance (α = 0.05).**

4.4 Discussion

Three distinct hypotheses were made prior to data collection. The first hypothesis, that there would be significant kinematic changes in the sagittal ankle and knee angles, tibial internal-external rotation, and forefoot-rearfoot inversion-eversion throughout the run, was supported. Parts of the second hypothesis were supported, as there was an increased presence and magnitude of impact peak, an increased loading rate, and a decline in vertical leg stiffness over the course of the run. Finally, ankle plantar flexor torque output decreased between pre- and post-run test as hypothesized; however, only the medial and lateral gastrocnemius muscles showed any difference in iEMG.

The increased ground contact time exhibited throughout the run may indicate a greater tend towards a RF strike pattern. FF and MF runners frequently spend less time in contact with the ground throughout the stride cycle than RF runners (Di Michele & Merni, 2013; Hayes & Caplan, 2012). Increased ground contact time has been linked to improvement in running economy (Di Michele & Merni, 2013; Williams & Cavanagh, 1987). Thus, the increased ground contact time exhibited by the participants in this study may indicate an effort to improve running economy with the onset of fatigue in order to maintain performance, coupled with a decreased ability to maintain a FF pattern and shorter contact time. This result was supported by the observation that half of the participants switched their strike pattern by the end of the run to a MF pattern.

The initial touchdown angle did not change to dorsiflexion over the course of the run according to the group mean, indicating that these runners were likely not landing with a RF pattern by run's end. However, the change in angle at touchdown indicated a trend towards landing more dorsiflexed, similar to previously reported RF ankle angles

(Pohl & Buckley, 2008). Half of the runners were determined to use a midfoot pattern based upon this data. Despite the trend towards a more posterior foot landing strategy for initial ankle angle, the opposite was true at toe-off where there was an increase in peak plantarflexion. RF runners typically have lower peak plantar flexion during the stance phase, while the runners in this study actually increased peak plantar flexion (Pohl & Buckley, 2008). The magnitude of this change was not particularly large, and may have been caused by the increase in ground contact time, as there was more time for the ankle to reach greater peak plantar flexion by the end of stance. Another possibility for the increased peak plantar flexion is that a greater range of motion at the end of the stance phase may be required to produce the force necessary to maintain speed with fatigue. These results were not significantly affected by time point in the run, as determined by the linear mixed effects regression model. The best fit line for the ankle angle across all time points of the run for all fourteen participants showed no significant change across all time points, indicating that the changes in ankle angle most likely occurred suddenly towards the end of the run, and not gradually with the onset of fatigue. This poor model fit over time could have arisen due to a washout by averaging participants across trials, as some participants showed no change in ankle angle from the beginning to end of the run.

Changes were also experienced further along the kinematic chain at the knee. The significant changes of the sagittal knee angle may indicate a few factors at play. First, as the ability of the ankle plantar flexors to produce force declined over the course of the run, the triceps surae may have lost the ability to sufficiently attenuate force, translating the impact shock further up the body. To combat this, an alternative attenuation strategy may have been adapted by allowing the knee to flex further, decreasing leg stiffness and

creating a more compliant lower limb (Derrick et al., 2002; Madigan & Pidcoe, 2003).

This may further decrease plantar flexion by reducing the contribution of the biarticular

gastrocnemius muscle, leading to a continued cycle of increased knee flexion and

reduced attenuation by the triceps surae (Madigan & Pidcoe, 2003).

Increased knee flexion during stance has been linked to improved running

economy (Williams & Cavanagh, 1987). Thus, the increased knee flexion during stance

may be both a result of a different method of shock attenuation, as well as an adaptive

strategy to increase running economy as the participants became more fatigued. Both

peak knee flexion and excursion were significantly affected by the time point in the run.

However, only the peak flexion angle showed a significant nonzero fit of the linear model

over time. This indicates that the change over time was gradual, with a steady increase of

knee flexion, rather than a sudden break point during the run at which flexion increased.

This gradual change was not observed at the ankle. However, changes were seen at the

knee in most participants, while seven participants showed little change at the angle.

Thus, in the participants who did demonstrate a change, these could have come on

gradually for both measures, which may have been skewed by washout across the means

when the non-changing participants were included.

Significant differences were found in internal-external rotation of the tibia during

stance, which are in agreement with the hypothesis that FF runners alter their strike

patterns over time. Pohl et al. (2008) found that RF runners exhibit greater internal

rotation and an overall greater range of motion throughout stance when compared to FF

runners. The increased internal rotation of the tibia found may indicate further altered

kinematic strategies for force attenuation, as it indicates that changes at the knee and

ankle were not constrained to the sagittal plane. Both of these changes showed a non-zero slope, indicating a gradual change over time similar to what was found at the knee. Changes in the tibial internal-external rotation patterns indicate further similarities between the strike patterns utilized by participants at the end of the run and a RF running pattern. The changes found between the beginning and end of the run correspond with changes shown in the literature between FF and RF patterns (Pohl & Buckley, 2008).

Forefoot-rearfoot inversion-eversion angles were also supportive of an altered strike pattern: a greater range of motion here is indicative of a RF pattern when compared to a FF pattern (Pohl & Buckley, 2008). No significance was found for any of the inversion-eversion measures when tested by the linear mixed effects regression model, indicating that time was not a significant factor on these measures. Increased peak internal rotation, as found at the end of the run in this study, further indicated the increasing similarities at the end of the run between the fatigued mechanics of these FF runners and RF strike patterns.

The kinetic results found in this study support the hypotheses of altered landing mechanics and a potential adaptive strategy to fatigue in FF runners. Active peak, though not found to be significantly reduced from beginning to end of the run, showed a trend towards reduction. With an increase in contact time, there is a longer period over which force can be applied. This longer force application allows for the propulsive impulse to be maintained while the overall active peak magnitude is reduced. A decreased active peak is common in RF runners compared to the force profile of FF runners (Dickinson et al., 1985; Laughton et al., 2003). Thus, the trend towards a decreased active peak

suggests a trend towards altered foot strike patterns and attenuation strategy as it approaches significance.

Along with a reduce active peak, an increase in both the impact peak and vertical loading rate is characteristic of a RF strike pattern (Cavanagh & Lafortune, 1980; Dickinson et al., 1985; Novacheck, 1998). The presence of an impact peak, or a more pronounced peak, is common in the RF strike pattern vertical ground reaction force profile; thus, the emergence of this peak may indicate that mechanical changes are occurring in the lower extremity.   The impact peak represents the passive impact transient of ground contact, and is an indicator of the type of shock attenuation strategy used (Novacheck, 1998). Thus, an increased prevalence and magnitude of impact peak may indicate that the runners were utilizing more passive methods of shock attenuation at the end of the run, more common in RF runners. This contrasts with the usual active attenuation strategies of FF runners, which were evident earlier in the run. These kinetic results likely resulted from changes occurring in the force-producing capability of the ankle plantar flexor muscles. The decline in torque output by run's end may indicate a decreased ability for these muscles to attenuate impact, causing the shift towards more passive mechanisms. This deficit at the ankle would likely cause the increased impact peak prominence as well as an overall increased loading rate. An increase in loading rate indicated that force was applied more rapidly to the lower extremity. With an increase in loading rate, or any change in force application to the lower extremity, other methods of force attenuation are necessary to maintain system integrity, such as the increased knee flexion throughout stance found in this study.

The significant decline in leg stiffness was expected given the kinematic results, such as increased knee flexion and ankle dorsiflexion, which led to a more compliant lower extremity. However, the fact that vertical stiffness did not change was somewhat unexpected based upon stiffness findings of other work (Dutto & Smith, 2002). Vertical stiffness is reliant upon peak vertical force; thus, the lack of significant change is not surprising when the peak vertical force results are taken into consideration. Despite the altered joint angles and kinetic results already described, vertical oscillations of the center of mass were kept relatively constant in the fatigued runners, further causing the vertical stiffness to remain unchanged. Decreased leg stiffness may indicate an increased reliance on knee flexion to attenuate force further up the kinematic chain due to the reduced impact shock absorption by the plantar flexor muscles at the ankle.

Possibly the most telling results for this study revolve around the final hypothesis: changes in ankle plantar flexion strength and the effects of the prolonged run on pre- and post-run EMG measures. Plantar flexion strength from pre- and post-run decreased significantly, indicating that fatigue may be at play, limiting the force-producing capability of the muscles crossing the ankle. However, mean frequency of the EMG signal, a primary measure of peripheral muscular fatigue, did not show any significant differences. This result was surprising, though not unheard of in the literature. Mizrahi *et al.* (1997) found no difference in both the time and frequency demands of EMG during a thirty minute run at anaerobic threshold (Mizrahi et al., 1997). Given that this study was performed to voluntary exhaustion, it was expected that these muscle groups would exhibit signs of peripheral fatigue by the end of the run. However, it is possible that repetitive submaximal loading of the triceps surae and other ankle plantar flexors was not

enough to induce fatigue of these muscles. It has been proposed that mean frequency

may not be an appropriate fatigue measure in activity involving low, intermittent load

levels (Mizrahi et al., 1997). Thus, force reduction may have been a result of central

fatigue, or general fatigue experienced throughout the body. Despite the lack of evidence

of peripheral muscle fatigue given by the mean frequency measure, there were significant

declines in iEMG in the lateral and medial heads of the gastrocnemius. These results are

may indicate decreased central activation during contraction, caused by either a decline in

motor neuron firing rates or a decrease in motor unit recruitment (Bigland-Ritchie,

Johansson, Lippold, & Woods, 1983; Stephens & Taylor, 1972).

The purpose of this study was to investigate the effects of a prolonged, intensive

run on the kinematics, kinetics, ankle strength, and EMG of FF runners. This study was

designed to expand upon existing research regarding FF and RF foot strike patterns, add

further evidence to the current debate on running form, and to add empirical kinetic and

kinematic evidence to support the findings of studies on competitive long distance races.

Much of the current literature has indicated that a RF running style is more economical

than, or at least metabolically equivalent to, a FF running style (Ardigo et al., 1995;

Gruber et al., 2013; Ogueta-Alday et al., 2013). Furthermore, research has shown that

over the course of long distance races, the small percentage of runners utilizing a FF

strike pattern decreases between the beginning and end of the race (Elliot & Ackland,

1981; Hasegawa et al., 2007; Larson et al., 2011). Thus, previous work and the results

found here may indicate that strength deficits caused by eccentric loading on the triceps

surae and Achilles' tendon in FF running lead to an increased metabolic cost of running,

87

causing these runners to alter their running mechanics to more closely resemble RF running (Ogueta-Alday et al., 2013).

4.5 Conclusion

Previous studies have indicated that the percentage of FF runners in endurance races declines between the beginning and end of the race. A FF running pattern has been found to be no more metabolically efficient than a RF running style, and some studies have shown that RF running is more economical than FF. The present study found that significant kinematic, kinetic, strength, and muscle activity changes occur in FF runners running to voluntary exhaustion. Ankle plantarflexion strength was found to decrease after the run, corresponding with decreased iEMG output, indicating that force reduction may have been caused by central fatigue. These changes in force output likely corresponded with adaptive changes in gait mechanics. Both kinematic and kinetic changes occurred, causing the typical FF patterns for these measures at the beginning of the run to trend towards resembling RF patterns. Initial ankle angle did move towards decreased plantarflexion; however, despite the similarities to a RF pattern, the strike pattern of these runners did not completely switch to a RF pattern. The findings of this study support the findings of previous literature, indicating that fatigue leads to a decrease in force output of the triceps surae which in turn leads to altered impact attenuation strategies and mechanics. Over a longer run, such as a 10k or marathon length effort, it is expected that this trend would continue, leading to even further deficits of ankle plantarflexor strength and a continued trend towards a RF strike pattern. Under these conditions, a higher percentage of runners may alter their mechanics to be classified as utilizing a different footstrike pattern. It may require a longer bout of running to

induce the peripheral muscular fatigue that was not found in this study during the shorter, intensive run.

CHAPTER 5

SUMMARY AND FUTURE DIRECTIONS

5.1 Summary

Studying the forefoot (FF) strike pattern has recently become more popular in the literature, due to its association with minimal and barefoot running. Promoters of the style tout improvement in running economy and injury reduction, recommending runners who utilize a rearfoot (RF) strike pattern to change to a FF pattern (Cavanagh & Lafortune, 1980; Hasegawa et al., 2007; Lieberman et al., 2010). However, recent work has indicated that there is no improvement in running economy in a FF pattern over a RF one; in fact, some findings have indicated that economy is improved in RF running (Ardigo et al., 1995; Gruber et al., 2013; Ogueta-Alday et al., 2013). Studies have found that increased speed often leads to a greater percentage of FF running (Hasegawa et al., 2007; Hayes & Caplan, 2012; Novacheck, 1998; Payne, 1983). However, studies focusing on competitive endurance events have also found that the percentage of runners using a FF strike pattern decreases from the beginning to end of the event, regardless of skill or fitness level (Elliot & Ackland, 1981; Hasegawa et al., 2007; Kerr et al., 1983; Larson et al., 2011). Thus, there must be biomechanical factors that prevent maintenance of this FF pattern in endurance events.

Significant differences in kinematic and kinetic patterns exist between FF and RF patterns, and are already reported prominently in the literature (Cavanagh & Lafortune, 1980; McClay & Manal, 1995; Novacheck, 1998; Pohl & Buckley, 2008; Williams et al., 2000). Most notable are the differences in initial touchdown sagittal ankle, which is dorsiflexed in RF runners and plantarflexed in FF runners, as well as the increased impact

peak presence and loading rate of the vertical ground reaction force. Work has been done in RF runners running with a converted FF strike pattern, and these runners have been able to exhibit some of the same characteristics of habitual FF runners (Williams et al., 2000). There is also a large body of literature regarding the effects of fatigue on running mechanics (Derrick et al., 2002; Dierks et al., 2010; Friesenbichler et al., 2011; Mizrahi et al., 2000a; Mizrahi et al., 2000b; Mizrahi et al., 1997). However, there has been little study on the kinematic, kinetic, and muscular effects of an exhaustive, prolonged run in an habitual FF running population.

5.2 Study Results

The findings of the present study indicated that significant kinematic, kinetic, and muscular level changes occur in FF runners over the course of a prolonged, intensive run. Both kinematic and kinetic variables measured confirmed the hypothesis that FF runners, when pushed to exertion, would begin to trend towards a RF strike pattern. Sagittal ankle and knee kinematics began to resemble characteristics of known RF patterns, as well as tibial rotation and forefoot-rearfoot inversion-eversion angles. A more prominent, and greater magnitude, impact peak arose in the vertical ground reaction force component, and was accompanied by an increase in vertical loading rate. Leg stiffness decreased between the beginning and end of the run, which along with the other kinetic findings, indicated altered landing and impact attenuation strategies. Maximal torque output of the ankle plantarflexor muscles decreased between pre- and post-run measurements, and integrated EMG of the lateral and medial heads of the gastrocnemius indicated a decline in central activation.

The results found in this study were in line with the original hypotheses; however, there were no significant results found in the mean frequency content of the ankle plantarflexor muscles at run's end, which is a surrogate measure for peripheral muscular fatigue. This was believed to be the primary mechanism driving the altered coordination strategies of the lower extremity in FF runners with fatigue. Despite this lack of significance, maximal force output still decreased, indicating that some mechanism is at play causing a diminished force-producing capability of the triceps surae. This points to a decreased ability maintain running economy while appropriately attenuating impact shock at the ankle through eccentric contractions. It is likely that the repetitive submaximal eccentric loading of the triceps surae and Achilles' tendon leads to impaired stretch-shorten cycling, increasing the overall metabolic cost of this active shock attenuation mechanism and leading the body to find different passive attenuation strategies, like a RF pattern. The findings of this study are in support of this hypothesis.

5.3 Future Directions

Further investigation is necessary to determine the specific mechanisms at work leading to these kinematic, kinetic, and muscular changes of FF runners over long, intense bouts. It is still unclear as to what drives the altered mechanics of these runners with fatigue, as well as what type of fatigue is at play: general, central, or peripheral. The present study provided empirical, laboratory-based evidence to support the findings of studies of competitive endurance events, indicating that changes were occurring in mechanics during the run. Collecting metabolic data along with kinematic and kinetic

data could aid in identifying the relationship between economy, fatigue, and altered mechanics.

The length of the run used in this study ensured that each runner reached voluntary exhaustion within the selected timeframe. However, the distance covered by these runners during the prolonged run was less than 5 kilometers for most. Much of the existing endurance race data involves races from 10k up to the marathon and ultramarathon type distances. Different energetic systems are involved in these longer endurance events and thus there could be different mechanisms, or different mechanics, involved in the latter stages of these runs. Literature already exists for RF running patterns at these long distances, but a more complete picture would be established by examining the effects of such a bout on FF runners.

The present study found that despite certain habitual running patterns, changes occur over the course of a prolonged, intensive run. The implications of these changes require further investigation as well. Injury prevalence was not a factor in selecting a population for this study; however, a longitudinal study could be used to look for any relationship between individual participant performances and changes over the run and future injury history and performance in the future. It remains a mystery why runners select their natural footfall patterns, and perhaps investigating the continued performance and training of a select group of FF runners would help begin to solve this problem.

Frequently in the literature, when FF running is investigated, the population used is a converted RF group – this is due both to the difficulty in locating a population of these runners and to have a perfectly matched control group. It would be worthwhile to conduct a training study in which habitual FF runners were converted to RF runners to

examine the impact on energetics, mechanics, and injury prevalence, in addition to competitive performance. The literature suggests that long, endurance race performances may be aided by this switch; however, it is unclear what type of injury effects (if any) that there would be, as well as shorter race performances. Also, if many of these runners already show similar characteristics to RF patterns by the end of a prolonged, intensive run, it may be the case that there would not be a substantial difference if the RF pattern was adapted for all running. Habitual foot strike patterns may not exist in an absolute for every individual. Each individual runner may utilize a wide spectrum of footfall pattern in given situations to optimize their mechanics for specific environmental and physiological constraints. This could have further implications for training, equipment, and injury treatment considerations.

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INFORMED CONSENT FORM
Biomechanics Laboratory
Department of Kinesiology
University of Massachusetts
Amherst, MA 01003

TITLE: Footstrike Patterns During a Prolonged Treadmill Run.

PRINCIPAL INVESTIGATOR: Carl Jewell, Joseph Hamill, Ph.D.
PURPOSE: To examine the changes in footfall patterns during a prolonged run.

REQUIRMENTS: You have been asked to participate in this study because: 1) you are an active male or female, age 18-45; 2) you are fit enough to complete a 30 minute treadmill run; 3) you run with a forefoot footfall pattern; and 4) you have not suffered any injuries to the lower extremity in the last year.

GENERAL TESTING PROCEDURES: The study consists of two visits. The first day of data collection, you will be asked to report to the Biomechanics Laboratory (Room 023), located in the Totman building. You will be observed running for short distances of about 20 meters with complete recovery in between to assess footfall pattern. After initial verification of your foot strike pattern, you will be fitted with wireless, non-invasive skin electrodes to monitor your muscle activity running both barefoot and wearing shoes over the same 20-meter runway. This visit will last 60 minutes.

In the second visit, you will again be asked to report to the Biomechanics Laboratory (Room 023), located in Totman building. Passive reflective markers and wireless electrodes will be placed at various body landmarks on your person and you will be asked to run on a treadmill at your preferred speed with these on. The laboratory will provide footwear that you will use during the run. The prolonged run will last approximately 30 minutes. In total, the procedure should take approximately 60-90 minutes. Before and after the run, you will be asked to perform an isometric calf raise wearing a weighted pack to compare pre- and post-exercise muscle fatigue. At the end of the procedure, all equipment will be removed from your person and you will be free to go.

EXPECTED RISKS OR DISCOMFORTS: During any type of exercise, especially a prolonged exercise, there are slight health risks. These include the possibility of fatigue and muscle soreness. However any health risks are small in participants who have no prior history of cardiovascular, respiratory or musculoskeletal disease or injury. Any ordinary fatigue or muscle soreness is temporary.

Participant Initials_____

EXPECTED BENEFITS: It is expected that the results of this study will broaden the theoretical basis for understanding footfall patterns during running.

ALTERNATIVE PROCEDURES: There are no alternative procedures that can be used non-invasively to measure these parameters. These procedures are standard for this type of experiment.

QUESTIONS AND ANSWERS: Any questions concerning testing procedures, risks, benefits, or participant's rights will be answered by the investigators.

WITHDRAWAL: You are under no obligation to participate in this project. You are free to withdraw your consent and participation at any time, for any reason.

CONFIDENTIALITY: All data collected during this session will remain confidential with regard to your name and identification.

COMPENSATION: No special compensation will be awarded for participation in this study.

ADDITIONAL INFORMATION: Should you have any questions about your treatment or any other matter relative to you participation in this project or if you experience a research related injury at any time during this study you may contact Dr. Joseph Hamill via e-mail (jhamill@kin.umass.edu); by telephone (413-545-2245); or by mail (Department of Kinesiology, Totman Building, University of Massachusetts Amherst, 30 Eastman Lane, Amherst, MA 01003). If you would like to discuss your rights as a participant in a research study or with to speak with someone not directly involved with this study, you may contact the Office of Research Affairs at the University of Massachusetts via e-mail (humansubjects@ora.umass.edu); by telephone (413-545-3428); or by mail (Office of Research Affairs, Research Administration Building, University of Massachusetts Amherst, 70 Butterfield Terrace, Amherst, MA 01003).

Participant Initials_____

STATEMENT AND PARTICIPANT SIGNATURE (Investigator Copy):

The investigators have read and understood the General Guidelines for the Right and Welfare of Human Subjects (Sen. Doc. 79-012) and agree to fulfill these guidelines to the best of their ability.

Investigator Signature: _____   Date: _____

As a participant in this study, I have read and understood the Informed Consent Document and hereby give my consent to participate in this study.

Participant Name: _____

Participant Signature: _____ _____   Date: _____

Address: _____

_____
       (City)   (State)

Telephone: (___)___ _ _____

Witness Name: _____Witness Signature: _____

STATEMENT AND PARTICIPANT SIGNATURE (Participant Copy):

The investigators have read and understood the General Guidelines for the Right and Welfare of Human Subjects (Sen. Doc. 79-012) and agree to fulfill these guidelines to the best of their ability.

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Participant Name: _____

Participant Signature: _____   Date: _____

Address: _____

_____
      (City)   (State)

Telephone: (__) _____

Witnessed: _____   Signed: _____

MODIFIED PHYSICAL ACTIVITY READINESS QUESTIONNAIRE

Date (MM/DD/YY): _____/_____/_____

Participant Number: _____

Please answer the following questions to the best of your knowledge (circle YES or NO).

1. **Yes** **No** Has your doctor ever said you had heart trouble or a heart murmur?

2. **Yes** **No** Do you ever suffer pains in your chest?

3. **Yes** **No** Do you ever feel faint or have spells of severe dizziness, passed out, palpitations or rapid heartbeat?

4. **Yes** **No** Has the doctor ever told you that your blood pressure was too high? (systolic $\geq$ 160 mm Hg or diastolic $\geq$ 90 mm Hg on at least 2 separate occasions)

5. **Yes** **No** Do you smoke cigarettes?

6. **Yes** **No** Do you have diabetes?

7. **Yes** **No** Do you have a family history of coronary or other atherosclerotic disease in parents or siblings prior to age 55?

8. **Yes** **No** Has your serum cholesterol ever been elevated?

9. **Yes** **No** Is there any physical reason not mentioned here why you should not follow an activity program even program even if you wanted to?

Below please provide an explanation for any of the questions to which you answered YES.

_____

_____

_____

_____

Participant Initials_____

QUESTIONNAIRE

Date (MM/DD/YY): _____/_____/_____

Participant Number: _____

Age (in years) _____

Gender: (check one) Female _____ Male _____

Height: _____ Feet _____ inches or _____ cm

Weight: _____lbs or _____ kg

Please Check One:

Do you use specialized insoles or foot orthotics? YES _____ NO _____

Do you have any injuries that may affect the way you walk or run: YES_____ NO _____

If YES, please describe the injury, and when it happened:

_____

_____

_____

Did you injure your lower extremities in the last year: YES _____ NO _____

If YES, please describe the injury, and when it happened:

_____

_____

_____

NON-SIGNIFICANT REGRESSION SCATTERPLOTS
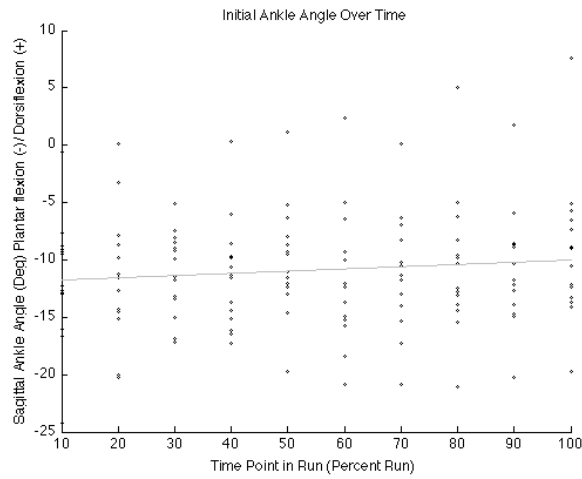
Figure B.1: Initial Sagittal Ankle Angle against Time Point in Run. Line is best fit of the data.
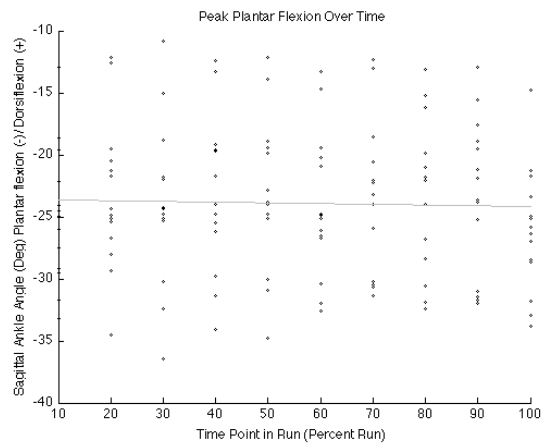

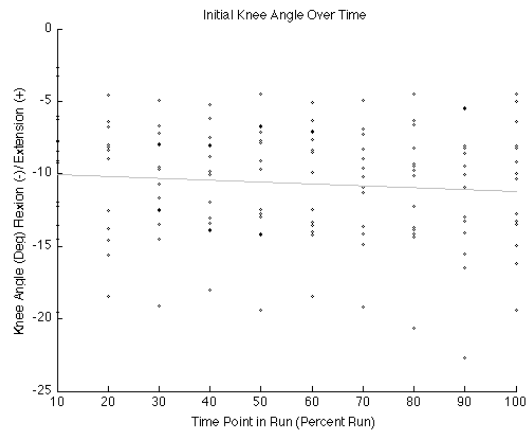**Figure B.2: Peak Plantarflexion Angle against Time Point in Run. Line is best fit of the data.**

Figure B.3: Initial Sagittal Knee Angle against Time Point in Run. Line is best fit of the data.
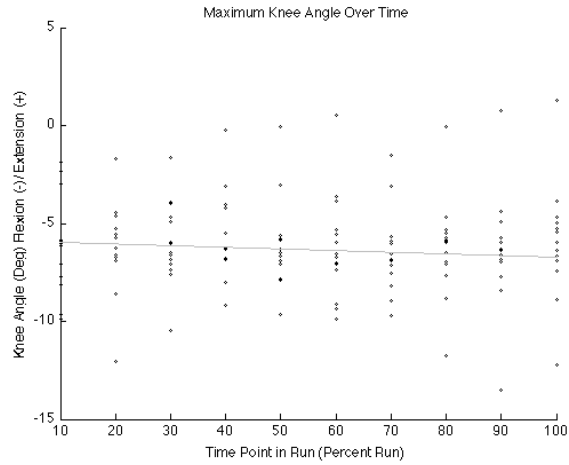
Figure B.4: Maximum Sagittal Knee Angle against Time Point in Run. Line is best fit of the data.
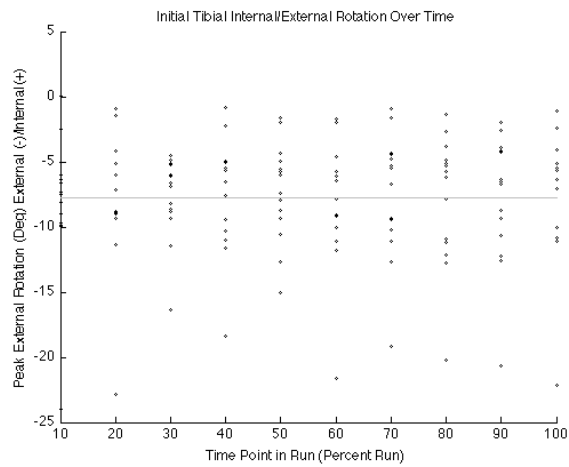

Figure B.5: Initial Tibial Int/Ext Rotation Angle against Time Point in Run. Line is best fit of the data.
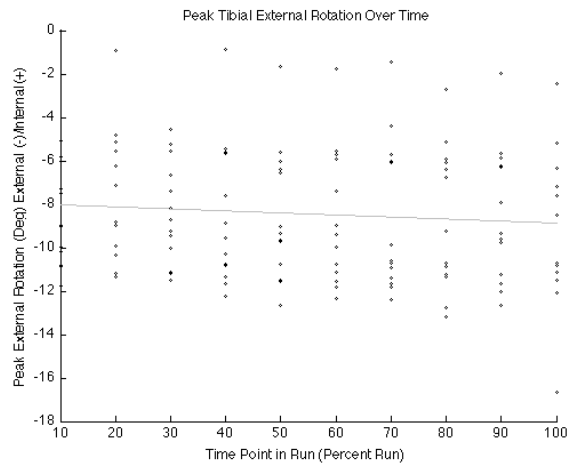

Figure B.6: Peak Tibial Rotation Angle against Time Point in Run. Line is best fit of the data.
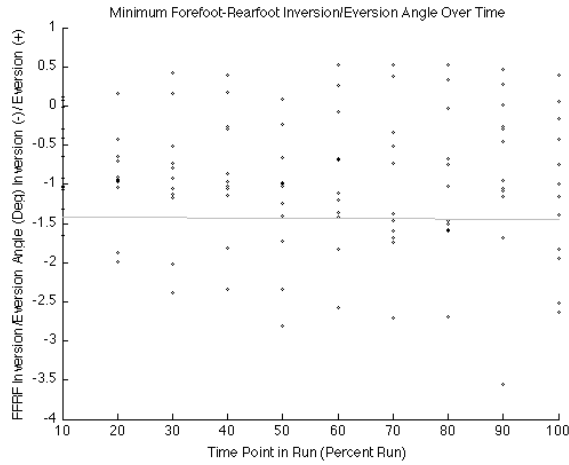
Figure B.7: Minimum FF-RF Inv/Ever Angle against Time Point in Run. Line is best fit of the data.
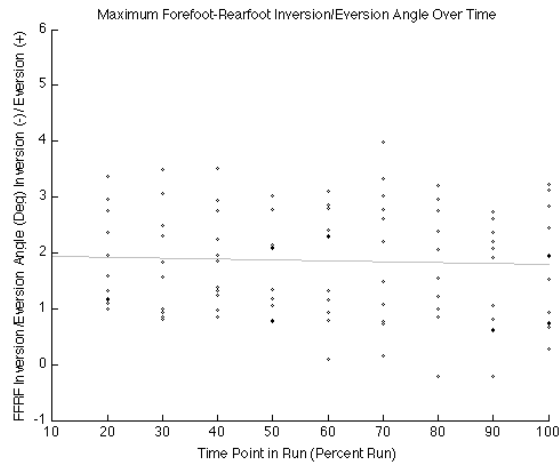

Figure B.8: Maximum FF-RF Inv/Ever Angle against Time Point in Run. Line is best fit of the data.
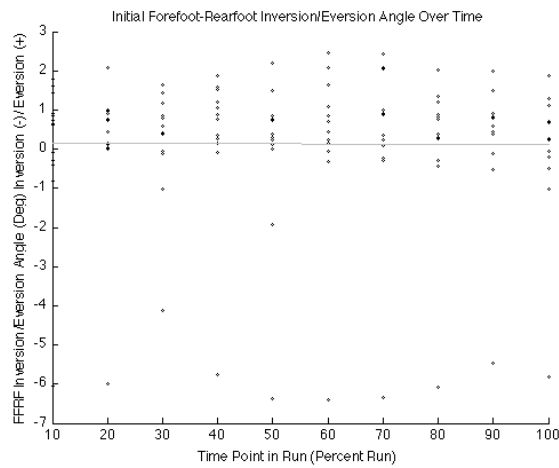

Figure B.9: Initial FF-RF Inv/Ever Angle against Time Point in Run. Line is best fit of the data.
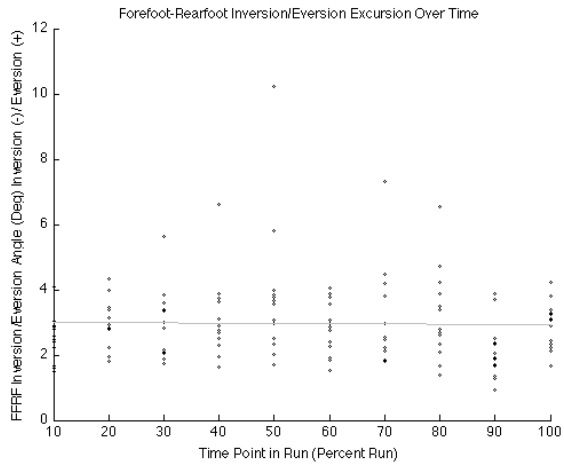
Figure B.10: FF-RF Inv/Ever Excusion against Time Point in Run. Line is best fit of the data.
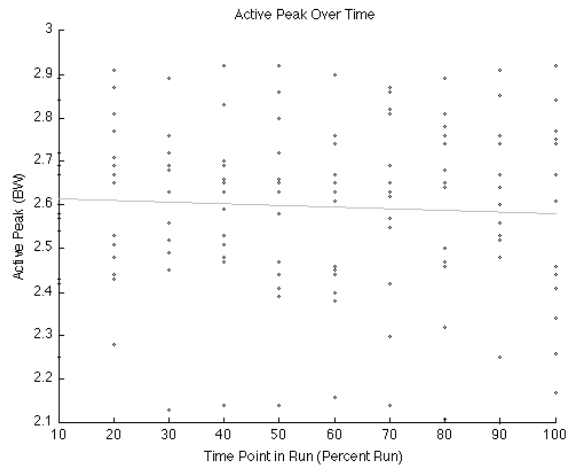

Figure B.11: Active Peak against Time Point in Run. Line is best fit of the data.

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