May 2016

Foot Kinematics and Neuromuscular Preactivation in Habitual Forefoot and Rearfoot Runners

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FOOT KINEMATICS AND NEUROMUSCULAR PREACTIVATION IN HABITUAL FOREFOOT AND REARFOOT RUNNERS

by

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A Dissertation Submitted in

Partial Fulfillment of the

Requirements for the Degree of

Doctor of Philosophy

in Health Sciences

at

The University of Wisconsin-Milwaukee

May 2016

ABSTRACT

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The University of Wisconsin-Milwaukee, 2016
Under the Supervision of Professor Stephen C. Cobb

As the rate of running related injuries has failed to decline despite advances in footwear, many researchers have begun focusing on different foot strike patterns possible contribution to injury risk. While many studies have focused on the differences between RFS and FFS running, few have investigated kinematic differences within the distal foot in habitual RFS and FFS runners and have failed to consider mechanical and neuromuscular changes due to fatigue. The purpose of this study, therefore, was to investigate foot kinematics and neuromuscular differences between RFS and FFS runners at the beginning and end of an exhaustive run. Fifteen habitual RFS and 15 habitual FFS runners (27.6 ± 5.64 years) performed a maximal 5 km treadmill run. A seven segment foot model was used with 3D motion capture methods to calculate joint kinematics of six functional articulations: rearfoot, calcaneonavicular, calcaneocuboid, medial forefoot, lateral forefoot, and first metatarsophalangeal (MTP). Four dual Ag/AgCl EMG surface electrodes were attached to the medial gastrocnemius, peroneus longus, soleus, and tibialis anterior to identify neuromuscular activity. Motion capture and EMG data were analyzed for five consecutive steps at the beginning and end of the 5 km run. Motion capture data was processed to investigate foot kinematic and joint coordination variability differences between the foot strike patterns at the beginning and end of the 5 km run. EMG data was processed to investigate neuromuscular preactivation onset and magnitude (iEMG) differences between the foot strikes at the beginning and end of the run. Mixed between-within groups statistical tests were used to compare variables between the foot strike patterns at the beginning and end of the exhaustive run. Exploration of kinematic results indicated a more supinated foot in FFS runners at initial contact and through early stance. The increased foot supination may result in a more rigid foot, but a less stable ankle joint. When the foot is moving toward greater pronation, a greater demand on soft tissues for stability is expected which may imply increased risk of soft tissue injury within the foot for RFS runners. Both groups demonstrated an increased range of motion at the end of the run during the first (0-20% of stance), 3rd (51-75% of stance), and 4th (76-100% of stance) stance subphases which may be a result of muscular fatigue and may increase injury risk to dynamic stabilizers of the foot articulations. With respect to joint coordination, rearfoot-midfoot coupling variability increased in both groups during midstance (21-50% of stance) at the end of the run. The increased variability may have been indicative of neuromuscular compensation to alter step-to-step variability in order to avoid overstressing tissues which may lead to overuse injury. Neuromuscular preactivation magnitude was increased and occurred earlier in the tibialis anterior in RFS runners and preactivation onset was earlier in the gastrocnemius in FFS runners. While RFS runners require tibialis anterior activation to maintain a dorsiflexed position at initial contact, it is likely that the earlier gastrocnemius onset in FFS runners facilitates positioning of the foot for initial contact with the forefoot. The earlier gastrocnemius onset in FFS with no significant difference in magnitude may suggest different roles of the gastrocnemius between the foot strikes and may be clinically relevant when looking

at overuse injury risks. There was no difference in neuromuscular preactivation as a result of the 5 km run, suggesting that neuromuscular fatigue did not affect how the muscles prepared for initial contact.

To my husband who supported me through this,

without him this would not have been possible,

and to my children

that they may understand the importance of

being educated and persistence in their goals.

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LIST OF ABBREVIATIONS

ABD	Abduction	MFF	Medial forefoot
ADD	Adduction	MFS	Midfoot strike
ANOVA	Analysis of variance	MSFM	Multi-segment foot model
CC	Calcaneocuboid articulation	MTP	1st metatarsophalangeal articulation
CNC	Calcaneonavicular articulation	MVC	Maximal voluntary contraction
DF	Dorsiflexion	PF	Plantarflexion
EMG	Electromyography	RC	Rearfoot complex
EV	Eversion	RFS	Rearfoot strike
FFS	Forefoot strike	RM	Repeated measures
fron	Frontal plane	RMS	Root mean square
HR_{max}	Heart rate max	RPE	Rate of perceived exertion
iEMG	Integrated electromyography	sag	Sagittal plane
INV	Inversion	SD	Standard deviation
LFF	Lateral forefoot	tran	Transverse plane
MANOVA	Multivariate analysis of variance	VGRF	Vertical ground reaction force
Met	Metatarsal		

ACKNOWLEDGEMENTS

I would like to thank first and foremost my advisor, Stephen Cobb, for being willing to take me on as a graduate student and embark on this journey with me. I have always been appreciative of his willingness to work with me and my circumstances, to always help me continue forward in my education and scholarship. As a non-traditional student, he has helped me to become the first in my family to achieve a terminal degree and set a high standard for my children in so doing. I am also very thankful to the students in the Musculoskeletal Injury Biomechanics Lab for assisting in my projects. A special thanks, especially, to Emily Gerstle for being there to bounce ideas and thoughts off of and assisting in all types of instrumentation malfunction.

Reaching this level would not have been possible without inspiring and encouraging instructors along the way. I thank to members of my dissertation committee, Kristian O'Connon, Jennifer Earl-Boehm, Kevin Keenan, and Kurt Beschorner for their support and insight throughout the dissertation process. A special thanks, to Ann Snyder, my master's advisor at UW-Milwaukee, for encouraging me to take the next step and showing her care and concern for me throughout my doctoral education as well. From my time as a student at the University of Utah, I thank Bob Toth for encouraging my investigation into research and Jim Martin for introducing me to the fun and exciting world of biomechanics.

Outside the classroom support was essential to advancing in my degree and a special thanks to my husband and children, Taylor, Audrey, Charley, and Eli, for understanding when mom had to stay late and for adding me to your prayers. I also thank my mother, Mary Schreiber, and Sarah Dominguez, for always helping with childcare needs and for being my “someone” to talk to when life got out of hand. And finally, a special thanks to my grandmother, Shirley Schreiber, for helping me always keep my sights high and for always loving and supporting not only me, but my entire family.

1. INTRODUCTION

Since the 1970's, the percentage of individuals who have taken up running has dramatically increased (Marti, Vader, Minder, & Abelin, 1988; van Mechelen, 1992; Yoder, 2013). As the amount of participants in the sport has risen, so too has the occurrences of running related injuries. Recent studies have indicated that 19.4-79.3% of runners suffer some form of running related injury each year (Bahr & Holme, 2003; van Gent et al., 2007). Much of the research to date has focused on decreasing running injuries through advancements in footwear. The underlying theory has been that impact loads are the major contributing factor to overuse running injuries (Light, McLellan, & Klenerman, 1980). However, there is little evidence that running injuries rates have decreased despite scientific advances in footwear (Richards, 2009) which has lead researchers to investigate the potential role of foot strike patterns on injury rates. Recently, Lieberman et al. (2010) investigated foot strike running patterns in habitually shod and habitually barefoot individuals and found that those who were not accustomed to wearing footwear generally ran with a midfoot or forefoot strike pattern. Habitual shod runners, on the other hand, tend to run with a rearfoot strike pattern (~75-94%) (Hasegawa, Yamauchi, & Kraemer, 2007; Kerr, Beauchamp, Fisher, & Neil, 1983; Larson et al., 2011). This has raised questions as to the foot's natural ability to attenuate impact forces. Subsequent research has focused on comparing barefoot and minimalist shoe running with that of traditional running shoes to try to identify which may better prevent running injuries (Divert et al., 2008; Shih, Lin, & Shiang, 2013; Squadrone & Gallozzi, 2009). The biomechanical differences between the habitually shod and barefoot runners identified by Lieberman, et al., (2010) however, were associated with the strike pattern, not the footwear. Only a few studies have used methods that allow biomechanical evaluation between strike patterns by controlling the shod condition (Ahn, Brayton, Bhatia, & Martin, 2014; Pohl & Buckley, 2008; Shih et al., 2013; Yong, Silder, & Delp, 2014). These studies have identified earlier plantarflexor neuromuscular activation in FFS runners (Ahn et al., 2014; Shih et al., 2013), strong forefoot-rearfoot coupling patterns (Pohl & Buckley, 2008), and better shock absorption when switching from a RFS to a FFS (Shih et al., 2013).

The differences in impact loads and ground reaction force between individuals running with different strike patterns has been well established (Cavanagh & Lafortune, 1980; Williams, McClay, Hamill, & Buchanan, 2001), and will not be the focus of this project. The ground reaction force differences between the strike patterns are, however, important to understand as they are related to the kinematics of the lower extremity. In general, rearfoot runners have an impact peak in the first subphase of running which is not evident in midfoot and forefoot runners (Lieberman et al., 2010; Williams, McClay, & Manal, 2000). The repetitive exposure of the lower extremity to large vertical ground reaction forces is theorized to result in overuse injuries (Cavanagh & Lafortune, 1980; Clarke, Frederick, & Cooper, 1983; Hreljac, Marshall, & Hume, 2000). The movement of adjacent joints during the stance phase of gait is also influenced by the magnitude and direction of the ground reaction force. As rearfoot strikers and forefoot strikers impact the ground at different areas of the foot, different lower extremity kinematics have been observed between the foot strike patterns when changing footwear conditions (Bonacci et al., 2013; Lieberman et al., 2010; Shih et al., 2013). At initial contact, rearfoot strikers have a greater dorsiflexed foot position than forefoot runners which contributes to differing knee and hip kinematics while running (Kulmala, Avela, Pasanen, &

Parkkari, 2013; Stackhouse, Davis, & Hamill, 2004). However, there is conflicting research regarding whether or not the differing rearfoot kinematics have an impact on the development of running related injuries such as tibial stress fractures and plantar fasciitis (Milner, Hamill, & Davis, 2010; Pohl, Hamill, & Davis, 2009; Pohl, Mullineaux, Milner, Hamill, & Davis, 2008; Tam, Astephen Wilson, Noakes, & Tucker, 2014; Willems et al., 2006). It has been suggested that investigation of more distal foot motion may contribute to the understanding of the relationship between foot kinematics and running related injuries (Pohl et al., 2009). To date, however, differences in distal foot motion between these foot strike patterns have not been investigated. Identification of the differences may be particularly crucial to forefoot running as the large ground reaction forces must first pass through the foot at initial contact.

When using motion capture techniques to evaluate lower extremity motion, the foot has traditionally been defined as a single, rigid body segment. Only recently has the use of multi-segment foot models been used in research. These foot models allow the foot to be segmented into multiple sections, generally ranging from 2-8 segments (MacWilliams, Cowley, & Nicholson, 2003; Morio, Lake, Gueguen, Rao, & Baly, 2009; Pohl, Messenger, & Buckley, 2006). Although a greater number of segments allow for a more detailed understanding of foot motion, limited skin surface area on the foot and current tracking technology increase the difficulty in being able to analyze motion of the multi-segmented foot. Therefore, a model with sufficient functional segments that does not inhibit data collection is important. The seven segment model selected for this project allows for reliable kinematic measurements (Bauer, Joshi, Klinkner, & Cobb, 2011) while representing functional segmentation of the rearfoot, medial and lateral midfoot, and medial and lateral forefoot. Differences in joint kinematics have been identified for both walking and running in individuals with different foot pathologies and arch height (Bauer, 2012; Cobb et al., 2009).

In addition to the motion at an individual joint, the interaction between joints, also impacts lower extremity motion and the transfer of forces along the kinetic chain. In rearfoot strike runners, abnormal joint coupling between the rearfoot complex and leg have been observed in individuals with patellofemoral pain and stress fractures (Bates, Osternig, Mason, & James, 1979; DeLeo, Dierks, Ferber, & Davis, 2004; Heiderscheit, Hamill, & Tiberio, 2001; McClay & Manal, 1997; Nigg, Cole, & Nachbauer, 1993; Powers, Chen, Reischl, & Perry, 2002). Very little, however, is known about joint coupling patterns during forefoot or midfoot strike running. By initiating ground contact with the forefoot, joint couplings within the distal foot will have a strong influence on the transfer of forces proximally along the lower extremity during the initial loading phase of stance. Those studies that have looked at foot joint coupling in rearfoot and forefoot strikers have only done so with a two segment (rearfoot, forefoot) foot model (Eslami, Begon, Farahpour, & Allard, 2007; Pohl & Buckley, 2008). Additionally, identifying a satisfactory method of quantifying joint coupling in a way that can be interpreted clinically has been a challenge (Chang et al., 2008; Miller, Chang, Baird, Van Emmerik, & Hamill, 2010). Currently the most accepted methods of quantifying joint coupling are continuous relative phase angle and vector coding. Although both are commonly used, the results of the two methods cannot be directly compared (Miller et al., 2010). Pohl and Buckley (2008) identified good coupling patterns between rearfoot frontal plane motion with transverse plane tibial rotation as well as forefoot sagittal plane motion using both cross-correlation and vector coding techniques. Using relative phase angle, Eslami, et al. (2007) identified a more in-phase relationship in forefoot transverse plane and rearfoot frontal plane coupling between the various phases of running stance for individuals running shod compared with barefoot running where out-of-phase coupling was observed (37° difference).

Joint coupling variability has further been utilized with cyclic movements, such as gait, as an indicator of injury risk. Previously, large amounts of variability in successive steps were thought to lead to injury. Though extreme amounts of variability are still thought to increase injury risk, the dynamic systems theory, however, hypothesizes that some variability may be beneficial to allow for adaptation to varying constraints between successive steps (Haken, Kelso, & Bunz, 1985; Schoner & Kelso, 1988). Because biological tissues adapt to stresses placed on them, variability while running may allow for stress to be dispersed to surrounding tissues and therefore avoid repetitive overuse. Though decreases in variability in ankle and tibial coupling has been found in injured runners, differences between forefoot and rearfoot runners have not been established.

Joint kinematics and variability of movement patterns are controlled by the neuromuscular system. Therefore, muscular activity is expected to vary between the different strike patterns. Increased activity of the triceps surae during forefoot running is expected as the heel is lowered to the ground after initial contact. Rearfoot strikers, on the other hand, require more tibialis anterior activity to maintain a dorsiflexed foot while making initial contact with the ground and then eccentrically contracting to lower the forefoot to the ground as well. Forefoot runners have demonstrated earlier preactivation of the gastrocnemius and delayed tibialis anterior preactivation when compared to rearfoot runners prior to initial contact (Shih et al., 2013). Preactivation of muscle contributes to joint stability and preparation for large impact loads which may otherwise cause injury. The different muscle requirements for joint stability, foot posture, and loading response between the foot strike patterns suggests that each foot strike pattern may have a different fatigue responses as well. Understanding how muscle preactivation patterns change after running exhaustion may aide in understanding the different roles each leg muscle contributes to the varying foot strike patterns and how to better avoid overuse musculoskeletal injury.

Most running motion capture studies have utilized a protocol that required the subjects to perform running trials after a brief warm-up period. However, many overuse injuries are thought to occur later in a run, when the lower extremity muscles become fatigued (Mizrahi, Verbitsky, & Isakov, 2000a; Verbitsky, Mizrahi, Voloshin, Treiger, & Isakov, 1998). Fatigue results in decreased force production and delayed muscle activity which may lead to abnormal lower extremity kinematics (Mizrahi et al., 2000a; Petrofsky, Guard, & Phillips, 1979; Wu, Chang, Wu, Guo, & Lin, 2007). This has been supported by research which has observed altered lower extremity kinematics following exhaustive runs (Derrick, Dereu, & McLean, 2002; Dierks, Davis, & Hamill, 2010; Donahue & Sharkey, 1999; Milgrom et al., 2007; Mizrahi, Verbitsky, Isakov, & Daily, 2000b; Mizrahi, Voloshin, Russek, Verbitski, & Isakov, 1997). Therefore, to better understand how foot and ankle motion may contribute to overuse injuries, it is necessary to consider alterations in these motions that occur as a result of fatigue. The differences in lower extremity neuromuscular activity between the foot strike patterns following an exhaustive run have not been investigated. Understanding how RFS and FFS runners respond differently to an exhaustive run may help to explain differences in the types and rate of injuries seen in runners.

As specific foot and ankle kinematics have been linked to various running overuse injuries, it is important to understand how kinematics differ between running styles. Muscle activity and tissue loading both influence the kinematics that occur during the stance phase of running. This may be especially true as lower extremity muscles become fatigued. Therefore, as overuse running injuries are closely linked to lower extremity kinematics, understanding how foot motion and muscle activity differ between strike patterns in uninjured runners will allow future research to further investigate how to prevent injuries for each foot strike pattern.

Statement of the Problem

Rearfoot kinematics are associated with overuse injury risk while running. However, most research investigating this relationship focused almost exclusively on rearfoot strike runners. Recent running trends, however, have integrated minimalist shoes and barefoot running, which results in many runners switching to a midfoot or forefoot strike pattern. To date, it is unclear how distal foot kinematics and leg musculature activity are influenced by different running strike patterns. Most previous research has treated the foot as a single rigid segment, neglecting the motion in the distal foot during the stance phase of running gait. Though a few studies have investigated foot motion with multi-segment foot models, the majority have not included models with sufficient functional segments and have only focused on rearfoot strike running. As many researchers, clinicians, and runners argue as to which foot strike pattern is superior, studies have yet to identify the foot kinematics and muscular activation associated with each foot strike pattern. In addition, most running research protocols do not consider the kinematic and muscular changes that may occur after exhaustion has set in.

Because many running injuries occur when some level of fatigue is present, consideration should be given to how an exhaustive run may change kinematics and neuromuscular profiles. Though recently converted FFS runners do not appear to have any differing kinematics from habitual FFS runners (Williams et al., 2000), asking recently converted FFS runners to perform an exhaustive run would likely have different results. Therefore, utilizing habitual RFS and FFS runners would be necessary to truly understand how these foot strike patterns differ at the beginning and end of an exhaustive run. Such information may help begin to understand what injury etiology occur with each foot strike pattern.

Purpose

The purpose of this project was to explore the differences in foot kinematics and lower extremity muscular activation between habitual rearfoot strike (RFS) and forefoot strike (FFS) runners before and after an exhaustive run. Specifically, patterns of distal foot motion, joint coupling variability, and extrinsic leg musculature preactivation were investigated.

Research Aims and Hypotheses

Specific aim 1. To determine the differences in joint kinematics between habitual RFS and FFS runners at the beginning and end of an exhaustive run.

Hypothesis 1.1. It was postulated that significant kinematic differences would be observed between RFS and FFS runners at initial contact and in distal foot range of motion during the early subphases of running stance. This was hypothesized to occur due to variation in center of pressure during early stance between rearfoot and forefoot running.

Hypothesis 1.2. No differences were expected in the second half of gait as push-off was thought to be similar between the two strike patterns. Further, similar fatigue effects on kinematics were expected in both strike patterns due to similar muscular usage during push-off in both the foot strikes.

Hypothesis 1.3. Following the exhaustive run, range of motion in all joints was hypothesized to increase for both foot strike patterns because of muscle fatigue. Because of large eccentric contractions of the ankle plantarflexors, FFS runners were hypothesized to have a larger increase in joint range of motion due to exhaustion.

Specific aim 2. To identify the differences in foot and ankle joint coupling variability between habitual RFS and FFS runners at the beginning and end of an exhaustive run.

Hypothesis 2.1. Because of a more distal vertical ground reaction force at initial contact in FFS running, increased joint coupling variability was expected in the distal foot during the first half of the stance phase. This variability was hypothesized to decrease with fatigue as a result of the neuromuscular system decreasing its ability to adapt and increasing overuse injury risk, based on the dynamic systems theory.

Specific aim 3. To identify the muscular preactivation differences between habitual RFS and FFS runners at the beginning and end of an exhaustive run.

Hypothesis 3.1. It was hypothesized that FFS runners would exhibit an earlier onset and larger iEMG value of gastrocnemius and peroneus longus activity. This was hypothesized due to the immediate and large eccentric motion the plantarflexors undergo during the loading phase of stance. Smaller values for the tibialis anterior were expected when compared to RFS runners because of the need for RFS runners to maintain a dorsiflexed foot posture at initial contact.

No differences were expected for the soleus based on previous research suggesting the soleus contracts similarly for each strike pattern (Jacobs, Bobbert, & van Ingen Schenau, 1993).

Hypothesis 3.2. Preactivation for both foot strike patterns was hypothesized to be delayed and iEMG was thought to decrease with fatigue as a result of decreased neural drive.

Delimitations

Purposeful decisions made in the design of this study include the following:

1. Fifteen habitual rearfoot strike (RFS) and forefoot strike (FFS) runners (minimum 10 miles running per week) were recruited from college campuses and running clubs/groups in the Milwaukee area to participate in this study.
2. The percentage of FFS runners has been estimated at 1-2% of runners based on various classification techniques. Recently, however, the percentage is thought to be underestimated based on the misclassification of midfoot strike runners (Graf, Rainbow, Samaan, & Davis, 2013). Therefore, since the two strike patterns do share similar vertical ground reaction force (VGRF) patterns, this study did not differentiate between midfoot strike runners and FFS runners. The methods for determining foot strike pattern in this study were by vertical ground reaction force while running.
3. The seven segment foot model chosen for this study was selected because it included medial and lateral segmentation of the midfoot and forefoot. Medial and lateral foot motion has been observed to contribute significantly to lower extremity kinematics (Rouhani, Favre, Crevoisier, Jolles, & Aminian, 2011; P. Wolf et al., 2008).

4. The sandals used for the study were chosen following pilot testing of several different 'minimalist' sandals in both RFS and FFS runners. All subjects used the same footwear to eliminate kinematic differences due to footwear. The purpose of the pilot testing was to identify a sole thickness that would allow barefoot or minimalist runners to maintain a FFS and that would allow traditional shod RFS runners to maintain a RFS. The 12 mm Mono sandals (Luna Sandals, Seattle, WA) allowed runners to run with their preferred foot strike pattern. A running speed of 7.5 mph was chosen for EMG and kinematic data collection because it was assumed that habitual runners that ran a minimum of 10 miles per week would be able to run for at least 30 s at that speed even at the conclusion of an exhaustive run.
5. A maximal 5 kilometer run was used as the exhaustive run protocol based on the time it generally takes to complete (under 40 minutes). As only a 10-day accommodation period is given the subjects, 5 km is a reasonable distance to build up to in the sandals in that time frame. Additionally, pilot testing had proven that exhaustive criteria based on heart rate max (HR_{max}) and rate of perceived exertion (RPE) can be achieved during the 5 km run. Furthermore, time trials and run-to-exhaustion methods have been shown to produce similar results of fatigue (Laursen, Francis, Abbiss, Newton, & Nosaka, 2007).
6. EMG of the tibialis anterior, peroneus longus, medial gastrocnemius, and soleus were chosen for observation. Tibialis anterior and gastrocnemius/soleus EMG have different roles in the FFS and RFS running pattern during the loading phase and therefore deemed necessary to include. The peroneus longus has yet to be investigated between the foot strike patterns, but is of interest because of its role in avoiding excessive inversion and

lateral ankle sprains. Delayed onset of muscle activity has been linked to individuals with lateral ankle sprains and therefore an understanding of onset timing while running is important to understand from a dynamic stability standpoint (Konradsen & Ravn, 1990).

7. Data collection for EMG and kinematic data were performed while participants ran on a treadmill, and not overground. This was decided because of the difficulty in some subjects to consistently perform running trials at a specified speed, especially after an exhaustive bout of exercise. In order to assure that the exhaustive effects do not wear off due to inconsistency in overground trials, all data was captured while running the 5 km run on a treadmill. Perhaps even more importantly, when looking for variability characteristics in joint coupling, consecutive steps are generally a better indicator of the step-to-step variability that occurs.
8. Vector coding methods according to Heiderscheit, et al. (2002) were selected over those of Ferber, et al. (2005) because of the range of angles. The Heiderscheit method allows for calculations of joint angles to range from 0-360^{o}, where the Ferber method allows for only 0-90^{o} range. The larger range allows for determination of in-phase, anti-phase characteristics of the joint coupling.

Limitations

The following are limitations of this study:

1. The retroreflective surface markers were assumed to represent the underlying bone of the various segments. Although, error from soft tissue artifact is inevitable

(Reinschmidt, van den Bogert, Nigg, Lundberg, & Murphy, 1997b), the effects were minimized by placing markers over skin where minimal soft tissue movement occurs and through the use of optimization procedures.

2. EMG noise occurs as a result of various factors including movement of wires and crosstalk (Farina, Merletti, & Enoka, 2004). Additionally, some wire leads were not working optimally and resulted in partial data for some subjects.
3. Various methods have been used for determining muscular activity onset, but none have been determined to be the most accurate. This study used a threshold of 2 standard deviations above baseline, but visual inspection of this was performed to assure accurate determination of muscular activity onset.
4. Testing was conducted in minimalist sandals and, therefore, is not to be directly inferable of motion in other types of footwear or to running barefoot.
5. Shoe uppers may add restriction to foot motion. To allow access for marker placement, sandals were chosen as opposed to shoes with cut out windows. Therefore, foot kinematic motion that occurs within a traditional running shoe may be different.
6. Because of space limitation on the surface of the foot and inaccessibility of some of the foot bones to surface markers, not all bony segments of the foot may be represented by surface markers. Functional segments were identified instead.
7. Three-dimensional motion capture requires that a minimum of two cameras have view of a retroreflective marker in order to acquire 3D coordinates. At times this does not occur and the marker appears to be missing. For estimating correct position of missing

markers, interpolation by cubic spline or by virtually joining based on the surrounding positional data were used.

Assumptions

The following assumptions were made when considering the results:

1. All subjects were injury free, healthy, and run at least 10 miles per week.
2. All runners participating in the study consistently ran with the foot strike pattern seen during the screening and were a representative sample of the population. Additionally, it was assumed that despite using a different type of footwear than what they habitually use, their habitual foot strike pattern remained the same.
3. The 10-day accommodation period given for the shoes was sufficient for shoe adaptation. Any changes in foot kinematics when comparing pre- and post- exhaustive run were assumed to be a result of the exhaustion and not caused by the insufficient footwear accommodation. This accommodation period length was pilot tested and used by other researchers (Bonacci et al., 2013; Nyska, McCabe, Linge, & Klenerman, 1996).
4. Maximal voluntary contractions (MVC) performed by the subjects were truly their maximal effort. Practice with maximal contractions during an earlier visit and strong verbal encouragement during the MVC were used to assure this takes place.
5. The 5 kilometer run was performed at the fastest pace possible by the individual. By allowing each subject to adjust speed accordingly and tracking heart rate and RPE, confirmation that the run was truly a maximal effort was monitored.

6. The cohesive bandage used to secure marker clusters on the foot was assumed to not affect kinematics of the foot as it is very flexible and even stiffer athletic tape has shown to not effect kinematics while running (Lindley & Kernozek, 1995; Verhagen, van der Beek, & van Mechelen, 2001).
7. Marker placement was accurate over the appropriate bony landmarks based on sufficient practice of the investigator applying the markers and by having the same investigator apply markers to all subjects.
8. Motion capture, EMG, and all other equipment used were accurate and sensitive enough to determine differences between and within groups.

2. REVIEW OF LITERATURE

Introduction

The primary purpose of this study was to compare the foot kinematics and leg muscular preactivation in habitual rearfoot strike and forefoot strike runners. This review of literature first gives an overview of the different foot strike patterns while running and their general kinematic characteristics. Because running shod and barefoot have different kinematic effects on the various foot strike patterns, this review includes an overview of how each foot strike pattern is effected by footwear. The foot model used in this study was a multi-segment foot model which partitions the foot into six segments. Therefore, this review also includes an overview of the different types of multi-segment foot models that have been used in research and functional segmentation of the foot which helps determine what segments are important to track for understanding foot motion. From the multi-segment foot model kinematics, joint coupling was determined for various adjacent foot articulations prior to determining joint coupling variability. An overview of joint coupling and its quantification methods are therefore included in this review of literature as well as details to the theories behind variability in biological systems. As all the kinematics and joint coupling variability are a result of muscle activation, which were observed in this study, this review of literature also includes an synopsis of leg muscular activation while running and particulars of electromyography (EMG) characteristics, specifically onset of muscle activity and integral EMG (iEMG) activity prior to the running stance phase.

In conjunction with the primary purpose of this study, the study will also incorporate an exhaustive run protocol to better understand how kinematics, joint coupling variability, and muscular preactivation may change as a result of fatigue. The final piece of this review of literature will explain what is known about fatigue and different fatigue protocols that have been used in research for mimicking training sessions and running performances. This is important to understand as altered kinematics may occur as a result of fatigue and lead to injury or decreased performance in running.

Running Foot Strike Patterns

Runners can be classified into three or four different strike patterns based on the area of the foot that first makes contact with the ground while running. The three main strike patterns are rearfoot, midfoot, and forefoot, with some researchers distinguishing toe striking as a fourth category.

***Rearfoot strike pattern*.** The rearfoot strike (RFS) pattern, also referred to as heel strike or heel-toe running, occurs when a runner makes initial ground contact with the heel of the foot. This method of running is the most common strike pattern among novice to elite runners with researchers identifying RFS in approximately 75-99% of runners (Bertelsen, Jensen, Nielsen, Nielsen, & Rasmussen, 2013; Hasegawa et al., 2007; Kasmer, Liu, Roberts, & Valadao, 2013; Larson et al., 2011). It should be noted, however, that this prevalence rate is lower (only 33%) when strike pattern is self-reported, suggesting that some runners may be unaware of the type of strike pattern they habitually perform (Goss & Gross, 2012). The exact location where the foot initially contacts the ground within the posterior portion of the foot in RFS runners can greatly vary, possibly being the cause of individuals misreporting their actual foot strike pattern (Cavanagh & Lafortune, 1980).

***Midfoot strike pattern*.** Midfoot strikers (MFS) make up approximately 0-23.7% of runners identified by researchers, and 43% of runners as identified by self-reporting methods (Bertelsen et al., 2013; Goss & Gross, 2012; Hasegawa et al., 2007; Kasmer et al., 2013; Larson et al., 2011). This strike pattern occurs when the forefoot and rearfoot make simultaneous contact with the ground (Cavanagh & Ae, 1980).

***Forefoot strike pattern*.** Runners with a forefoot strike pattern (FFS) only comprise 0-2% of runners as observed in research studies, though 20% of runners self-report having a FFS (Bertelsen et al., 2013; Goss & Gross, 2012; Kasmer et al., 2013; Larson et al., 2011). This strike pattern is defined as making initial contact with the forefoot, or distal 1/3 of the foot (Cavanagh & Ae, 1980). Sprinting, or fast-paced running are most often associated a FFS (Doherty, 1971).

***Toe strike pattern*.** The additional foot strike pattern that has been identified as toe striking occurs when initial contact is made with the forefoot, but the rearfoot never makes contact with the ground. Because the foot strikes the ground in the forefoot region, despite not making heel contact, this foot strike pattern is often not differentiated as a separate foot strike pattern (Lieberman, 2012) and, likewise, was not differentiated from the FFS pattern presently.

***Foot Strike Classification Methods*.** Discrepancies in the prevalence of runners that use each strike pattern are further complicated by the various methods that exist for classification of foot strike pattern. The least technical method reported in the literature is by visual methods, most often by examining footage taken during a race with a video camera (Hasegawa et al., 2007; Kerr et al., 1983). Visual methods are also still well excepted techniques for determining foot strike pattern. However, error can occur due to digital camera limitations such as frame rate and inadequate focus which may lead to improper foot strike classification. This method also seems to be less accurate for identifying MFS (Altman & Davis, 2012b).

Cavanagh and Ae (1980) introduced a foot strike determining method in which the total length of the foot or shoe is divided into equal thirds (proximal, middle, and distal). The position of the foot's center of pressure at initial contact with respect to the dividing lines determines if the runner is a FFS (distal third), MFS (middle third), or RFS (proximal third) runner (Cavanagh & Ae, 1980). It is thought that perhaps the number of FFS runners which has previously been reported in the literature using this method may be incorrect because of how FFS runners contact the ground (Graf et al., 2013). When barefoot, FFS runners appear to contact the ground on the lateral aspect of their 5th metatarsal head. In some individuals, this occurs on the border of the partition between the distal and middle sections of the foot. Therefore, many FFS runners may be incorrectly identified as MFS. Recent methods have been developed to avoid this problem by partitioning the foot into thirds between the most posterior aspect of the calcaneus and the 1st metatarsal head rather than the tip of the second toe or shoe (Graf et al., 2013). This method appears to be a valid method of identifying FFS patterns during barefoot running. The method has not, however, been utilized to compare foot strike pattern during shod running (Graf et al., 2013).

Another method of foot strike pattern identification has been to examine the dorsiflexion angle of the ankle at initial contact. Altman and Davis (2012a) found this method to be 60-83% reliable when comparing it to visual methods (Hasegawa et al., 2007; Kerr et al., 1983) and the center of pressure method (Cavanagh & Ae, 1980). This method is least accurate with FFS runners as they sometims make initial contact in a dorsiflexed ankle position. This is made possible by increasing the knee flexion angle at initial contact which allows for minimal change in the ankle angle in order to make contact with the forefoot at initial contact (Squadrone & Gallozzi, 2009).

Ground reaction force characteristics of the different running strike patterns is also different between RFS and FFS running and can be used to distinguish between them (Bramble & Lieberman, 2004; Lieberman et al., 2010; Williams et al., 2000). The most pronounced difference is a sharp initial impact peak, or a vertical ground reaction force (VGRF) curve with a double peak (Figure 2.1), in RFS running and is discussed in further detail later in this review. When comparing MFS and FFS running, however, both patterns are similar in that they do not have the characteristic initial impact peak seen in RFS running. The VGRF curve was used for the present study to distinguish between the foot strike patterns. Distinction was only made between FFS and RFS running patterns. Individuals with a MFS pattern were included in the FFS group since their VGRF patterns are similar and difficulty is found in distinguishing between the two strike patterns (Altman & Davis, 2012a; Williams et al., 2000).

***Foot strike pattern prevalence*.** Beyond differences in classification methods, prevalence of each foot strike pattern may be influenced by the length of endurance running and different levels of running experience. Hasegawa et al. (2007) reported foot strike patterns (n = 283) based on visual methods (filming speed = 120 Hz) at the 15 km point of an international half marathon to be 74.9% RFS, 23.7% MFS, and 1.4% FFS. Kerr et al. (1983) used similar methods as Hasegawa et al. (2007) but observed (filming speed = 60 Hz) recreational runners in a 10 km race and more competitive runners in a marathon (n = 753) and found 81% RFS, 19% MFS, and 0.2 % FFS. In this study, only 45% of faster runners (running speed of 332 m/min) demonstrated a RFS running pattern compared to 88% in slower runners (running speed 207 m/min). Similarly, Larson et al. (2011) reported 94.4% RFS, 3.6% MFS strikers, and 1.9% FFS at the 10 km point for recreational half-marathon, relay, and full-marathon runners using visual methods (filming speed = 300 Hz, n = 936). In this study, marathon runners were also video recorded at the 32 km point where 92.3% of FFS runners had converted to a MFS or RFS and 59.5% of MFS runners converted to RFS. Only 1.8% of RFS converted to either a MFS or FFS. Differences in foot strike prevalence may be partially attributed to the experience level of runners, one study being elite to sub-elite runners (Hasegawa et al., 2007) and the other being sub-elite to recreational runners (Larson et al., 2011) as well as a difference in filming speed (120 Hz and 300 Hz, respectively) in which the initial contact moment may have been miss identified. As reported in Kerr et al. (1983), there is a much lower prevalence of RFS runners in those who complete distance running at a faster pace (88% compared to 45%). In a laboratory setting, Nunns, House, Fallowfield, Allsopp, and Dixon (2013) reported foot strike patterns in male military recruits (n = 120) to be 77% RFS, 8.3% MFS, and 10.8% FFS when running barefoot by using a plantar pressure mat and dividing the foot into thirds. This study is the only known study to use a large cohort with more technical measures than those of visual methods and demonstrated a much larger portion of FFS runners than the previous mentioned studies. However, most of the observed difference is likely due to the recruits running barefoot rather than the methods chosen.

***Summary*.** In summary, it is well accepted that the majority of runners make initial ground contact with a RFS, though MFS and FFS are not uncommon. There is some discrepancy as to the best method for determining one's foot strike, with visual, kinematic, and kinetic data being used for determining foot placement at initial contact. Though joint angles and plantar pressure locations have been used, visual methods continue to be well accepted for determining foot strike patterns, though utilizing cameras with faster filming speeds may produce more accurate results. Characteristic impact peaks in the VGRF curve further aid in discriminating between the various foot strike patterns. This study used VGRF data to determine if an individual is a RFS or FFS (which will include FFS and MFS) runner.

Running Kinematics

***Running gait cycle*.** The running gait cycle is defined between the instant a foot first contacts the ground (initial contact) to the instant the same foot contacts the ground again (Novacheck, 1998). Within this cycle, the stance phase is identified between the initial contact and toe off events of the same foot. The stance phase during running is generally around 30-40% of the total gait cycle with less contact time associated with faster running speeds (Novacheck, 1998).

The running stance phase can be further separated into four subphases, with each subphase exhibiting key factors of stance. Subphase 1, or loading phase (~0-20% stance phase), is defined between initial ground contact to initial loading. In RFS running, this is where the initial impact peak is observed. Subphase 2, or midstance (~21-50% stance phase), occurs between the initial loading to full body weight acceptance. It marks the end of absorption, when the leg is no longer working to absorb VGRF, but prepares to push against the ground for forward propulsion. Subphase 3, or terminal stance (~51-75%), begins with full body weight acceptance and ends half the distance to toe-off. During subphase 3, the center of gravity of the body moves over the midpoint of the center of pressure and the leg begins to push the body forward. Subphase 4 is from subphase 3 to the final toe-off (~76-100%), and often is referred to as pre-swing (Ferber et al., 2005). It is during this phase that final propulsion occurs to move the body forward and the foot leaves the ground.

***Effect of Foot Strike Pattern on Running Kinematics*.** Much of running research has investigated the effect of footwear on running kinematics, comparing various shod conditions to barefoot. Some researchers, however, have neglected to consider the shift in foot strike pattern prevalence for each of these conditions.   Barefoot runners often utilize a MFS or FFS pattern and the majority of shod runners present with a RFS (Bramble & Lieberman, 2004; Lieberman, 2012; Lieberman et al., 2010). Exceptions have been reported especially when habitually shod runners initially transition to barefoot running or vice versa (Lieberman et al., 2010). However, many RFS runners who habitually run shod convert to a MFS or FFS when barefoot (Divert et al., 2008; Lieberman et al., 2010). Most research investigating the difference between barefoot and shod running kinematics have only used habitually shod runners who more often have a RFS. Further, some researchers have required subjects to perform a specific foot strike pattern although the subject may have preferred a different one (e.g. a habitual shod RFS runner asked to run barefoot RFS, but naturally shifts to a FFS when barefoot). It is difficult to conclude, therefore, that the kinematics in these studies mimics what habitual runners of each foot strike pattern would actually display. As habitually shod runners tend to acquire a MFS or FFS while barefoot, it is unknown at what point this occurs and at what point it becomes habit. It was suggested by Divert, Mornieux, Baur, Mayer, and Belli (2005b) that conversion from a RFS habitual shod runner to a FFS while running barefoot may occur after a specific running length. Methods that require participants to only take a few steps down a runway may allow participants to maintain a RFS while longer runs, such as the three minute run used by Divert et al. (2005b) may require FFS conversion to avoid discomfort. There are also, however, runners who habitually run shod that run with a FFS or MFS (Cavanagh & Lafortune, 1980; Hasegawa et al., 2007; Larson et al., 2011). Therefore, it is important to consider both how the strike patterns influence running kinematics and how footwear contributes to the differences between the strike patterns.

***Rearfoot strike running*.**

Shod running. Arguments have been made that the modern-day running shoe with cushioning, arch support, and a heel-to-toe drop promote RFS running (Lieberman et al., 2010; Robbins & Hanna, 1987). Due to the increased sole thickness under the heel, the heel is more likely to make ground contact first because cushioning prevents ground contact from being uncomfortable. When running with a RFS pattern there are two distinct peaks visible in the VGRF during stance (Figure 2.1-A) (Cavanagh & Lafortune, 1980; Nigg, Bahlsen, Luethi, & Stokes, 1987). The initial peak is referred to as the passive peak as passive structures such as shoe cushioning have been thought to control its magnitude. This impact peak is an indicator of the loading rate and magnitude of force on the foot which is then transferred up the kinetic chain. The loading rate of the VGRF has been theorized to be related to overuse running injuries (Light et al., 1980; Robbins & Hanna, 1987). When a RFS occurs during shod running, the heel cushion causes an increase in the time it takes to lower the center of mass compared to when RFS occurs during barefoot conditions. The result is a reduction in the magnitude of the passive VGRF peak and loading rate at initial contact in shod versus barefoot conditions (De Clercq, Aerts, & Kunnen, 1994; De Wit, De Clercq, & Aerts, 2000; Divert et al., 2005b; Komi, Gollhofer, Schmidtbleicher, & Frick, 1987). This requires different kinematics and muscle control than with a FFS.

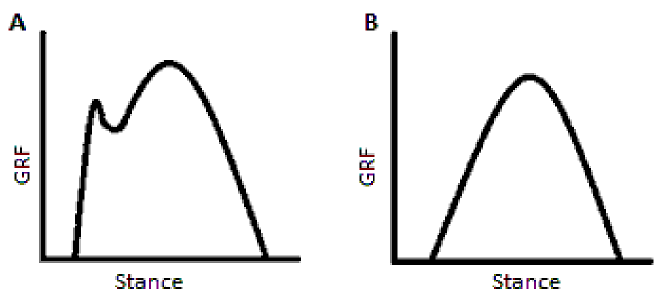

Figure 2.1. Typical vertical ground reaction force (VGRF) of a rearfoot strike (RFS) runner (A) and forefoot (FFS) or midfoot strike runner. In A, the first peak, also known as the passive peak, is not usually visible when running with a FFS or midfoot strike (B).

Kinematic analyses of running have customarily identified the foot as a rigid segment, analyzing joint motion of the ankle, or rearfoot complex. Generally, at initial contact when running, the rearfoot complex, defined as the combined motion of the talocrural, subtalar, and talofibular joints (Nester, 1997), is in a dorsiflexed position and slightly inverted to allow for initial contact to occur on the lateral border of the calcaneus (Cavanagh & Lafortune, 1980; Mann, Baxter, & Lutter, 1981; Reinschmidt, van Den Bogert, Murphy, Lundberg, & Nigg, 1997a). The rearfoot complex plantarflexes slightly during subphase 1 to allow for the forefoot to contact the ground. The rearfoot complex then begins to dorsiflex until reaching a peak during subphase 2. Sagittal plane motion then reverses, reaching peak plantarflexion at take-off (Figure 2.2-A) (Reinschmidt et al., 1997a; Stacoff, Kaelin, Stuessi, & Segesser, 1989; Williams et al., 2000). The initially inverted rearfoot complex everts through subphases 1 and 2 of stance. Like dorsiflexion, peak eversion occurs during subphase 2 of stance. Inversion then completes the stance phase through subphases 3 and 4 (Figure 2.2-B) (Reinschmidt et al., 1997a; Stacoff, Nigg, Reinschmidt, van den Bogert, & Lundberg, 2000a; Williams et al., 2000). In the transverse plane, the rearfoot complex is approximately in a neutral position at initial contact. The rearfoot complex then abducts as the body transitions from subphase 1 to subphase 2 and peaks around 20-25% stance. Following peak adduction, the rearfoot complex steadily adducts throughout subphases 3 and 4 (Figure 2.2-C) reaching peak adduction at take-off (Reinschmidt et al., 1997a).

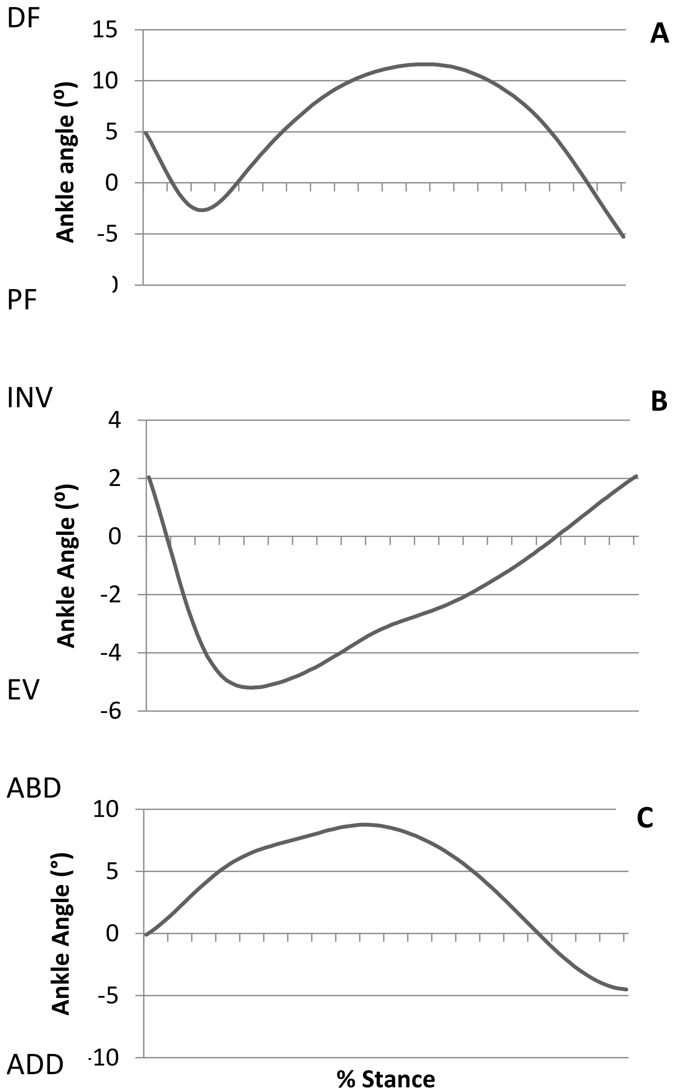

Figure 2.2. Graphs depicting motion of the rearfoot complex in the sagittal plane (A), frontal plane (B), and transverse plane (C) while running shod with a rearfoot strike pattern.
DF= dorsiflexion, PF=plantarflexion, INV=inversion, EV=eversion, ABD=abduction, ADD=Adduction

Barefoot running. When running with a RFS pattern while barefoot, the passive VGRF peak and its rate of loading are often increased when compared to RFS shod running (De Clercq et al., 1994; De Wit et al., 2000; Dickinson, Cook, & Leinhardt, 1985; Komi et al., 1987; Oakley &

Pratt, 1988). Interestingly, heel striking while barefoot also tends to have a double impact peak, or two separate peaks during initial loading (De Clercq et al., 1994; De Wit et al., 2000). This has led many to dispute the safety of running barefoot as the loading peak is associated with injury (Light et al., 1980; Robbins & Hanna, 1987). As a result of the changes in VGRF characteristics, RFS runners that continue in their habitual RFS pattern tend to alter kinematics and increase cadence to help attenuate shock while running (Shih et al., 2013).

The most noted difference in rearfoot kinematics while running barefoot with a RFS versus running shod with a RFS is a decreased angle of dorsiflexion at initial contact. Traditional shod runners that maintain a RFS while running barefoot do so at a 4-11^{o} decreased dorsiflexion angle at initial contact (Bonacci et al., 2013; De Wit et al., 2000; Lieberman et al., 2010; Shih et al., 2013). Shih et al. (2013) observed a 4.52^{o} decrease in ankle angle at initial contact and a 3.64^{o} increase in ankle range of motion throughout stance when running barefoot in 12 habitually shod RFS male runners. Marker placement, however, was on the shoes' surface for the shod condition and on the skin for the barefoot, which could account for some of the reported differences. De Wit et al. (2000) compared rearfoot dorsiflexion and eversion in seven long distance male runners during both barefoot and shod conditions (markers were placed on the shoe during the shod condition) using a two-dimensional technique. Participants were instructed to not alter their running pattern between the conditions; they were to maintain a RFS. Despite the instruction, kinematic differences were observed with decreases in rearfoot eversion during impact (3.3 to 4.0^{o} difference) and dorsiflexion position at touchdown (8.2 to 10.7^{o} difference) while running barefoot. Bonacci et al. (2013) found similar results between barefoot and shod running (4.48 ± 1.6 $m \cdot s^{-1}$) in highly trained runners (n=22, mean 10 km time = 33.7 ±3.7 min) with smaller differences also observed between traditional running shoes and a minimalist shoes (Nike Free 3.0) or racing flats (Nike LunaRacer2). Rearfoot complex angle at initial contact ranged from 0.78^o of dorsiflexion in the barefoot condition to 4.52^o in the minimalist shoe, 4.25^o in the racing flat, and 5.31^o in the traditional running shoe. Peak dorsiflexion angle was also lower in the barefoot condition (24.94 ± 2.6^o) than in the traditional shoe (27.51 ± 2.7^o) and racing flat (26.33 ± 2.9^o) conditions. The participants in the barefoot condition also had greater plantarflexion at toe-off (10.91 ± 9.6^o) than all of the shod conditions (4.77 ± 9.5 to 6.01± 8.4^o). Peak ankle eversion was also observed to be decreased in barefoot running (9.7 ± 2.5^o) when compared with traditional shoes (12.55 ± 3.0^o) or racing flats (11.03 ± 3.3^o) (Bonacci et al., 2013). It is difficult to determine by these studies, however, how accurate the differences seen between barefoot running and shod running are as markers for tracking motion were generally placed on the shoe. It has been shown that there is a significant difference between shoe motion and foot motion within the shoe (Morio et al., 2009; Stacoff et al., 2000a; Stacoff et al., 2001). In contrast to the previous studies, Stacoff et al. (2000a) tracked foot motion as opposed to shoe motion and found no difference in rearfoot complex inversion when comparing shod and barefoot runners. However, intracortical bone pins were used for this study, limiting the number of participants to only five males, and participants were instructed to not change their foot strike pattern. Although individual differences were observed between the conditions for some participants, the differences never varied more than 3 or 4 degrees. The eversion velocity, however, was slower when the runners ran barefoot (barefoot = 116.64 ± 28.40 deg·s^{-1}, shod ranged from 129.92 ± 29.94 to 144.30 ± 56.52 deg·s^{-1}). In summary, even within habitual RFS runners, there are large amounts of kinematic variability due to the shoe and especially between barefoot and shod conditions where the strike pattern may change completely. Despite the variability, however, studies that allowed for foot strike pattern changes in the different types of footwear saw similar patterns in kinematic changes between barefoot and shod conditions.

***Forefoot strike running*.**

Shod running. Unlike with RFS running, FFS running often does not demonstrate a passive VGRF impact peak. Researchers have repeatedly observed a smooth parabola shaped VGRF curve in FFS runners (Figure 2.1-B) (Cavanagh & Lafortune, 1980; Oakley & Pratt, 1988; Williams et al., 2000). Marked differences in VGRF between the strike patterns are mostly limited to the absence of the passive impact peak during initial loading (subphase 1). Oakley and Pratt (1988) directed a subject pool of mostly non-runners to run (running speed = 3.3 to 3.6 $m \cdot s^{-1}$) with a heel strike and then a toe strike pattern. Loading rates while RFS running were over 7.5 times higher than those forefoot striking (Oakley & Pratt, 1988). Habitual FFS runners have also been compared to recently converted FFS runners and both groups were found to have similar loading rates and peak VGRF (Williams et al., 2000). Shih et al. (2013) investigated shod RFS and FFS running in 12 habitual shod RFS male runners and found an increase in loading rate of 17.46 $BW \cdot s^{-1}$ when running with a RFS compared to FFS while shod. The decrease in loading rate during FFS running was attributed to the attenuation of the VGRF as the heel was lowered to the ground. During the lowering of the heel, the VGRF can be transferred to angular motion of the ankle and further dissipated by the ankle plantarflexors as they eccentrically contract.

Altered rearfoot complex sagittal plane motion between FFS and RFS runners is most evident in subphase 1 of running. A plantarflexed rearfoot complex position (10.05 ± 3.66^{o} at initial contact) is generally present in FFS running for the entire loading phase (subphase 1) as it progresses towards a dorsiflexed position in subphase 2 of stance (Shih et al., 2013; Stackhouse et al., 2004). Shih et al. (2013) observed almost a 20^{o} difference in sagittal plane ankle angle initial contact position between RFS and FFS shod running of 12 male habitual RFS runners. In addition to the apparent difference in sagittal plane motion at the beginning of the stance phase of running, Stackhouse et al. (2004) observed decreased peak eversion, increased eversion excursion (FFS = 16.38^{o}, RFS = 13.65^{o}) and velocity (FFS = $270.6\ ^{o}\cdot s^{-1}$, RFS = $190.91\ ^{o}\cdot s^{-1}$) and increased dorsiflexion excursion (FFS = 31.57^{o}, RFS = 19.24^{o}) and velocity (FFS = $426.73\ ^{o}\cdot s^{-1}$, RFS = $317.36\ ^{o}\cdot s^{-1}$) in 15 habitual RFS runners when they ran with a FFS. In this study, although the subjects conducted the testing in running shoes, holes were cut in the area around the calcaneus to allow for a marker cluster to be attached to the foot, unlike the previously mentioned articles where the markers were placed on the shoe (Shih et al., 2013; Stackhouse et al., 2004).

Barefoot running. Where marked differences occur in VGRF loading rate between shod and barefoot conditions while running with a RFS, such dramatic differences are not seen in FFS running (Oakley & Pratt, 1988). In a study of 14 non-experienced runners and four experienced runners, Oakley and Pratt (1988) found no significant difference in loading rate during FFS running when comparing barefoot to shod. Shih et al. (2013) instructed 12 habitual RFS runners to run with a RFS in both shod and barefoot conditions and then with a FFS in both shod and barefoot conditions. They found loading rate differences between all conditions except between the FFS barefoot and shod conditions. Based on VGRF characteristics, barefoot and shod FFS running does not have the differences seen when comparing VGRF of barefoot and shod RFS running.

Only small kinematic differences have been reported between barefoot and shod FFS running. Shih et al. (2013) observed a 2.93^{o} increased plantarflexion ankle angle at initial contact while running with a FFS in a shoes compared to barefoot. This difference is most likely due the shoe's sole which presents with a thicker portion under the heel and thinner sole under the forefoot. As markers in the shod condition were placed on the shoes' uppers and on the skin in the barefoot condition, it is unclear if the differences reported were a result of actual kinematic changes or shoe motion compared to foot motion. Squadrone and Gallozzi (2009) observed eight habitual barefoot runners and found a 7^{o} difference in ankle sagittal plane angle at initial contact with barefoot running when compared to shod, but no difference in any kinematic variables between barefoot running and running in minimalist shoes (Vibram Fivefingers). A shift of initial contact plantar pressure to the posterior portion of the foot while running shod indicated a flatter foot placement at initial contact, but for all three conditions (barefoot, minimalist shoe, running shoe) the mean strike index (Cavanagh & Ae, 1980) was consistent with that of a MFS runner. It was not reported if all runners were MFS runners or if some were FFS and others RFS runners. Beyond sagittal plane motion, research is lacking in comparing kinematic characteristics between barefoot and shod FFS running.

Despite only small differences in kinematic and kinetic characteristics observed, spatiotemporal parameters between FFS barefoot and shod running are evident. FFS running requires a higher frequency of step rate, smaller step length, and shorter contact time (Bonacci et al., 2013; De Wit et al., 2000; Divert, Baur, Mornieux, Mayer, & Belli, 2005a; Divert et al., 2005b; Hamill, Russell, Gruber, & Miller, 2011; Nunns et al., 2013; Squadrone & Gallozzi, 2009). Shih et al. (2013) identified an increased cadence (2.80 ±0.19 vs. 2.72 ± 0.18 steps·s^{-1}) and decreased time of flight phase (13.89 ± 2.03 vs 15.25 ± 2.29 ms) in barefoot FFS running when compared to shod FFS running. Squadrone and Gallozzi (2009) found a decreased stride length and increase in stride frequency in barefoot running (2.19 ± 0.2m and 91.2 ± 0.9 stride/min) compared to running in minimalist shoes (Vibram Fivefingers, 2.29 ± 0.16m and 88.3 ± 0.9 stride/min) and traditional running shoes (2.34 ± 0.15m and 86.0 ± 1.1 stride/min) in habitual barefoot runners. A change in any of these spatio-temporal variables alone may contribute to small alterations in VGRF and kinematic patterns in runners. For this reason, it is important to monitor other variables that may also alter spatio-temporal variables, such as running speed and footwear.

In comparing shod and barefoot running kinematics, few researchers have controlled for strike pattern, making it difficult to determine if differences are due to strike pattern changes or the footwear condition. Studies that have controlled for strike pattern have often used habitual RFS runners and request them to also perform the FFS trials. Shih et al. (2013) has been the only known study to compare the FFS pattern and footwear conditions while also looking at ankle angle. At initial contact, they observed a significant increase in plantarflexion of nearly 3° when running shod compared to barefoot with the FFS pattern.

Summary. When running, kinematic differences of the rearfoot complex are regularly reported between RFS and FFS running regardless of footwear type. Most of these differences are reported in sagittal plane motion during subphase 1 of stance. It is during subphase 1 that

RFS runners have a dorsiflexed rearfoot position at initial contact which quickly plantarflexes while FFS runners contact the ground with a plantarflexed rearfoot position and proceed to dorsiflex the ankle. These kinematic differences may undergo further changes based on the type of footwear that is worn while running. Traditional running shoes will decrease the dorsiflexion angle of the foot because of the shoe drop which elevates the calcaneus, but make it more difficult to run with a MFS or FFS. Most extreme differences due to footwear appear between traditional running shoes and barefoot running conditions. In addition, most research comparing RFS and FFS running use habitual RFS runners for both running conditions, making it unclear if habitual FFS runners would exhibit the same kinematics. It is still unclear how kinematics of individuals with a habitual RFS compare to those who habitually FFS, irrespective of what type of footwear they are running in.

Multi-Segment Foot Modeling

The foot has historically been modeled as a single rigid segment in gait analysis despite the foot's mobility created by its 26 bones and 30 articulations. Many researchers have found evidence to support the notion that alterations in rearfoot motion can lead to further alterations along the kinetic chain (McClay & Manal, 1997; Messier & Pittala, 1988; Shambaugh, Klein, & Herbert, 1991; Stergiou, Bates, & James, 1999), but have ignored the contribution of movement that occurs within the foot itself and how it effects lower extremity motion. Until recently, motion capture technology has made it difficult to identify specific segments within the foot because of the size of the foot and inability to properly identify segments with surface markers. The talus, which sits between the tibia and calcaneus, making up the talocrural and subtalar joints, is completely surrounded by other bones. Capturing talus motion is therefore only accessible through imaging such as with magnetic resonance imaging or computed tomography, cadaveric studies, or invasive intracortical bone pin methods (Beimers et al., 2008; Hirsch, Udupa, & Samarasekera, 1996; C. J. Nester et al., 2007; Whittaker, Aubin, & Ledoux, 2011). Each of these methods has its limitations. To date, medical imaging techniques are unable to capture motion while moving through a range of motion and cadaveric data cannot adequately take into account the role of musculature and the individuals' habitual motion pattern. Because of the invasive nature of bone pins, few studies are able to directly track bone motion, and those that do have had very small sample sizes (Arndt et al., 2007; Lafortune, Cavanagh, Sommer, & Kalenak, 1992; Lundgren et al., 2008; C. Nester et al., 2007; Reinschmidt et al., 1997a; Stacoff et al., 2000a). Additionally, high inter-subject variability observed in studies using bone pins, make such studies difficult to generalize to the population (Nester, 2009; Stacoff et al., 2000a). Only one study has used bone pins to quantify motion of the individual foot bones while running. Arndt et al. (2007) tracked the foot motion of 4 males during slow barefoot running (1.9-2.3 $m \cdot s^{-1}$) using intracortical bone pins. Though variability between subjects was high for some joints, general motion patterns were evident. There was a larger range of motion in subtalar joint plantarflexion and dorsiflexion (24.7 ± 3.9^o) than in the talocrural joint (8.9 ± 3.2^o). Large talonavicular rotations were also evident (frontal = 13.5 ± 4.1^o, sagittal = 6.5 ± 2.9^o, transverse = 8.7 ± 1.4^o). Additionally, motion of other joints ranged from 1.6 – 11.4^o in all planes for the following articulations: cuboid-fifth metatarsal, cuboid-calcaneus, cuboid-navicular, cuneiform-navicular, and cuneiform-first metatarsal.

Surface markers are a non-invasive method used for motion capture that has commonly been used to track dynamic movement of the underlying bony structure. However, soft tissue motion becomes an issue with the use of surface markers (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996; Reinschmidt et al., 1997b). Despite this error, surface markers are readily used and accepted as an adequate form of analyzing gait kinematics. They also provide a safe, non-invasive alternative to intracortical pins and therefore larger subject populations are able to be assessed. Because of the lack of motion description provided by modeling the foot as a single rigid body, foot and ankle researchers identified the need for adequately producing models that accurately describe foot motion (Davis, 2004). A variety of models using surface markers have been introduced in the literature varying from 2 to 8 segment models (Table 2.1 and 2.2) (Buczek, Walker, Rainbow, Cooney, & Sanders, 2006; Cobb et al., 2009; Hwang, Choi, & Kim, 2004; Jenkyn, Anas, & Nichol, 2009; Leardini et al., 2007; MacWilliams et al., 2003; Morio et al., 2009; Rouhani et al., 2011). Stacoff et al. (1989) were among the first to evaluate foot motion during running by comparing frontal plane motion of the forefoot (identified as torsion) and rearfoot (identified as pronation). They found that by decreasing forefoot torsion through the use of shoes, rearfoot pronation was increased in nine subjects with varying levels of fitness (sedentary to 10 hrs of exercise per week). Where much of the running injury focus had been on excessive rearfoot complex/foot pronation, Stacoff et al. (1989) successfully introduced the importance of understanding the motion occurring within the foot. As a result, researchers have begun to partition the foot into various segments beyond the traditional single rigid segment. Because of the required minimum of three markers per identified segment for 3D motion capture, identifying each bone within the foot via surface markers is not currently possible due to space restriction. For this reason, functional segmentation is necessary. Wolf et al. (2008) stated that "a functional unit is present when its bones rotate either (i) in the same direction, (ii) in the opposite direction, or (iii) when one or more bones show no rotation." Based on intracortical pin data while walking, Wolf et al. (2008) were able to identify functional units of the talonavicular and navicular-medial cuneiform in the frontal and transverse plane, the navicular-cuboid in all three planes, and the medial cuneiform-first metatarsal. It was recommended that minimally the calcaneus, navicular-cuboid, medial cuneiform-first metatarsal, and fifth metatarsal should be identified as separate segments. Rouhani et al. (2011) attempted to find functional segments as well using reflective, surface markers and suggested segmentation to occur in the rearfoot, medial and lateral forefoot, and toes. These recommendations did not include the midfoot as a segment despite their conclusion that the leg-rearfoot and rearfoot-midfoot were the dominant movement joints during walking gait.

***Multi-segment foot model running kinematics*.** Of all the multi-segment foot models (MSFM) that have been introduced, few have been used to identify motion patterns while running. Pohl et al. (2006) used a two segment foot model, identifying the forefoot and rearfoot, on 12 recreational runners (running a minimum 2 hours per week). They observed the forefoot in dorsiflexion and abduction during the first half of stance and then plantarflexion and adduction during the second half of stance under three different step widths while running. Specific discrete variables for forefoot motion were identified and included forefoot dorsiflexion excursion (10.2 to 12.5^{o}), abduction excursion (-5.5 to -6.6^{o}), time to peak forefoot dorsiflexion (52.4 to 59.3% of stance), and time to peak forefoot abduction (46.0 to 50.5% of stance). It was noticed, additionally, that during a narrow, cross-over step pattern, some subjects transitioned to a FFS pattern which may have contributed to high variability recorded in the forefoot frontal plane kinematics. In a later study, Pohl and Buckley (2008) observed the same discrete variables of forefoot and rearfoot motion during RFS, FFS, and toe running for the same set of subjects. Significant differences were observed for the FFS and toe running strike patterns when compared to RFS running in forefoot dorsiflexion excursion (13.0 ± 2.4^{o} FFS, 13.9 ± 2.1^{o} toe running, 7.3 ± 1.8^{o} RFS) and forefoot abduction excursion (-6.4 ± 1.7^{o} FFS, -6.5 ± 1.4^{o} toe running, -3.7 ± 1.2^{o} RFS). Time to peak forefoot dorsiflexion and abduction were only different while toe running (52.1 ± 6.8% and 45.9 ± 8.4% of stance) when compared to RFS running (58.2 ± 4.9% and 53.8 ± 11.5% of stance).

Using a similar foot model as Pohl et al. (2006), Morio et al. (2009) compared foot motion to shoe motion by looking at discrete variables of running in barefoot and during two shod (sandal) conditions. Differences were observed between barefoot and shod conditions for forefoot eversion amplitude (12.0 ± 3.9^{o} barefoot, 10.7 ± 4.6^{o} soft sole, 10.0 ± 4.9^{o} hard sole), eversion slope (-371.9 ± 98.5^{o}/s barefoot, -232.3 ± 126.9^{o}/s soft sole, -142.8 ± 138.4^{o}/s hard sole), eversion/inversion maximum slope (-10.9 ± 179.0^{o}/s barefoot, 116.8 ± 115.9^{o}/s soft sole, 136.0 ± 114.8^{o}/s hard sole), abduction excursion (-3.4 ± 2.2^{o} barefoot, -4.5 ± 2.2^{o} soft sole), and adduction amplitude (9.6 ± 2.6^{o} barefoot, 8.2 ± 2.8^{o} soft sole, 6.9 ± 2.6^{o} hard sole). No significant difference was seen between the footwear conditions in the sagittal plane.

The Leardini MSFM (Leardini, Benedetti, Catani, Simoncini, & Giannini, 1999) includes a midfoot segment, something not included in the previously mentioned studies, and has been used to investigate running kinematics of female athletes with high and low arches (Powell, Long, Milner, & Zhang, 2011; Powell, Williams, & Butler, 2013). Combined (n=10 low-arched, n=10 high arched), peak eversion at the rearfoot-midfoot and midfoot-forefoot joints ranged from 5.3 ± 2.6^{o} to 5.5 ±2.5^{o} and 6.5 ±2.5^{o} to 11.6 ± 5.0^{o} respectively. Eversion excursion was between 3.7 ± 3.2° to 3.9 ± 1.8° for rearfoot-midfoot joint and 5.6 ± 1.7° to 9.2 ± 3.7° for midfoot-forefoot eversion. The Leardini model (Leardini et al., 1999) has also been compared to the commonly used Oxford model (Carson, Harrington, Thompson, O'Connor, & Theologis, 2001) in running trials (Powell et al., 2013). The main difference between these two models, again, is the inclusion of a midfoot segment in the Leardini model which is excluded in the Oxford model. The Leardini model was shown to be more sensitive in detecting peak eversion angles of the midfoot and forefoot while the Oxford model was more sensitive to detecting eversion excursion values between the forefoot and rearfoot, most likely due to the combining of two segments (forefoot and midfoot) which would increase the overall motion by summing the joint excursion of the two joints. This comparison is an example of how simplification of a foot model to not include functional segments may not adequately describe the motion that is occurring in the foot. In addition to the inclusion of a midfoot segment, some researchers have recognized the different movement characteristics of the medial and lateral forefoot, but few MSFM have included this aspect despite its ability to be reliably measured (Bauer et al., 2011; Hwang et al., 2004; Jenkyn & Nicol, 2007; MacWilliams et al., 2003; Rouhani et al., 2011; Simon et al., 2006; Tome, Nawoczenski, Flemister, & Houck, 2006). Only one foot model containing both medial and lateral midfoot and medial and lateral forefoot segments has been found reliable during running protocols (Bauer, 2012; Bauer et al., 2011; Bauer, Joshi, Klinkner, & Cobb, 2012; Seneli, Joshi, Bauer, & Cobb, 2013). This model is a six segment model (hallux, medial forefoot, lateral forefoot, navicular, cuboid, and calcaneus) and has been tested on healthy adults (Bauer et al., 2011; Bauer et al., 2012; Cobb, Bauer, & Joshi, 2014), adults with low arch structure (Seneli et al., 2013), and adults with plantar fasciitis (Bauer, 2012). From this foot model it is evident that during barefoot running the medial forefoot moves differently than the lateral forefoot (sagittal plane range of motion = 20.90 ± 4.68^o and 7.68 ± 3.19^o, respectively) and the calcaneonavicular complex functions differently compared to the calcaneocuboid joint (sagittal plane range of motion = 10.85 ± 2.57^o and 18.66 ± 1.86^o) (Bauer et al., 2012).

***Limitations of foot models*.** While there are advantages to understanding variables that effect foot motion and therefore advantages to MSFM, there are also many limitations to this type of motion capture. One such limitation is adequate access to the foot segments. Some researchers evaluating MSFM while running have required their participants to run barefoot (Pohl & Buckley, 2008; Pohl et al., 2006; Powell et al., 2011). This is relevant since previous studies have established that foot and ankle mechanics differ between running barefoot and running shod (Bonacci et al., 2013; Lieberman et al., 2010; Shih et al., 2013; Squadrone & Gallozzi, 2009; Zhang, Paquette, & Zhang, 2013). Studies that have used sandals as a shod condition have done so with the assumption that increased motion may occur within the foot compared to a traditional running shoe with restricting uppers, though stiffer sandals do still limit the motion over barefoot conditions (Eslami et al., 2007; Morio et al., 2009). Simply tracking the shoe is not an accurate description of foot motion (Morio et al., 2009). Other researchers have cut holes into shoe uppers to avoid losing the constriction of the shoe upper and still allow for skin placement of markers and marker clusters (Reinschmidt et al., 1997a; Shultz & Jenkyn, 2012; Stacoff et al., 2000b; S. Wolf et al., 2008). Increasing the number of markers and marker clusters, however, would require extreme modification of a shoe's upper.

Therefore, if striving for evaluating kinematics with a shod condition, the complexity of the MSFM may dictate the type of footwear that is needed for the study.

***Summary*.** Based on the information from previous studies, it is evident that the foot is not a single rigid segment, but rather has an intricate movement pattern dependent on the speed, style of footwear, and other parameters while running. These movement patterns within the foot itself help dictate kinematics and kinetics along the lower extremity which may affect performance and injury status of runners. Choosing proper segmentation that is representative of foot motion is imperative to adequately understand how the foot contributes to lower extremity motion. In addition to the rearfoot complex which is regularly distinguished in kinematic running studies, it is now known that the midfoot and forefoot should be separately distinguished as should medial and lateral section because of the motion that occurs in these segments independent of the other segments. It is unknown to what extent varying kinematics occurs between foot strike patterns. With proper segmentation to best represent foot motion, further understanding of foot strike pattern kinematics can be investigated to further understanding into kinematics that prevent injury and improve performance.

Article	Population tested	Mode	Segments	Rearfoot/Calcaneus	Talus	Midfoot	Cuboid	Medial foot	Lateral foot	Medial Rearfoot	Navicular	Forefoot	Medial Forefoot	Lateral Forefoot	1st Met	1st Met- Hallux	Hallux	Lateral Phalanx	Medial Phalanx	Phalanx 1-5
Carson et al. (2001)	adults: club feet	walking	3	X								X					X			
Jenkyn & Nicol (2007)	adults	walking	5	X	X	X							X	X						
Bruening et al. (2012)	children	walking	3	X								X					X			
Cobb et al. (2009)	adults: low arches	walking	4	X							X		X			X				
Dixon (2007)	children	walking	3	X								X					X			
Stebbins et al. (2006)	children	walking	3	X								X					X			
Kidder et al. (1996)	adult	walking	3	X		X											X			
Hwang et al. (2004)	adults: males	walking	8	X	X		X						X	X			X	X	X	
Rattanaprasert et al. (1999)	adults: tibialis posterior dysfunction	walking	3	X								X					X			
Cornwall & McPoil (1999)	adults	walking	3	X							X				X					
Leardini et al. (2007)	adults	walking	3	X		X						X								
Arampatzis et al. (2002)	gymnasts	landing	6	X	X			X	X									X	X	
Tome et al. (2006)	adults: tibialis posterior dysfunction	walking	4	X									X	X			X			
MacWilliams et al. (2003)	children	walking	8	X			X			X			X	X				X	X	

Table 2.1. Summary of the different multi-segment foot models that have been developed and modified in research. Under population tested, all studies with a pathological sample also had a healthy control sample tested. Met = metatarsal

Article	Population tested	Mode	Segments	Rearfoot/Calcaneus	Talus	Midfoot	Cuboid	Medial foot	Lateral foot	Medial Rearfoot	Navicular	Forefoot	Medial Forefoot	Lateral Forefoot	1st Met	1st Met- Hallux	Hallux	Lateral Phalanx	Medial Phalanx	Phalanx 1-5
Leardini et al. (1999)	adults	walking	4	X		X									X		X			
Rao et al. (2009)	adults: midfoot arthritis	Walking, step down	4	X									X	X			X			
Houck et al. (2009)	adults: tibialis posterior dysfunction	walking	3	X											X		X			
Sawacha et al. (2009)	adults: diabetic foot	walking	3	X	X	X														
Rouhani et al. (2011)	adults	walking	9	X		X							X	X						X
Saraswat et al. (2012)	children: planovalgus feet	walking	3	X								X					X			
Dubbeldam et al. (2013)	adults: rheumatoid arthritis	walking	4	X		X						X					X			
Myers et al. (2004)	children	walking	4	X								X					X			
Powell et al. (2011)	adult: female athletes	walking, running, step down	3	X		X									X					
Morio et al. (2009)	adults: males	running	2	X								X								
Pohl et al. (2006)	adults: recreational runners	running	2	X								X								
Bauer et al. (2011)	adults: plantar fasciitis	running	6	X			X				X		X	X		X				

Table 2.2. Continuation of the summary of the different multi-segment foot models that have been developed and modified in research. Under population tested, all studies with a pathological sample also had a healthy control sample tested. Met = metatarsal

Joint Coupling

Joint coupling has been determined to be an important factor to consider when looking at both walking and running gait for performance and injury purposes. It is theorized that perhaps the movement pattern of one joint by itself is not the cause of dysfunction, but rather how that joint is functioning with other joints along the kinetic chain. Extensive research has been conducted on the joint coupling of ankle pronation (eversion) with tibial internal rotation during closed chain activities. The coupling relationship has been theorized to contribute to pain and disability associated with pathologies such as patellofemoral pain and stress fractures (Bates et al., 1979; DeLeo et al., 2004; Heiderscheit et al., 2002; McClay & Manal, 1997; Nigg et al., 1993; Powers et al., 2002). Quantifying this relationship in a clinically relevant method, however, has been a challenge. Kinematics of a single joints are most often compared using angle-time graphs, but such graphs do not adequately characterize or measure coupling between two joints even when the two joints are illustrated on the same graph. The difficulty in adequately describing joint couples has led to various ways of quantifying the coupling.

***Quantification methods of joint coupling*.**

Discrete variables. Some studies have quantified joint coupling by examining joint position at discrete points along the stance phase (Pohl & Buckley, 2008; Pohl et al., 2006; Reinschmidt et al., 1997a; Stacoff et al., 2000b), while others take into account the dynamic movement of the joints throughout the stance phase (Hamill et al., 1999; Heiderscheit et al., 2001). Arguments have been made that the discrete quantification methods do not take into account the dynamic movement of the joints throughout the stance phase of gait and may miss critical moments when the coupling is out-of-phase (Hamill et al., 1999).

Joint ratios. Joint ratios have been used to quantify the amount of movement occurring at one joint or in one plane of motion with respect to that of another joint (DeLeo et al., 2004; McClay & Manal, 1997; Nigg et al., 1993). When calculating joint ratios, the angular displacement of the distal joint over a given portion of gait, such as initial contact to a peak angle or a discrete point along the stance phase, is divided by that of the proximal joint. The ratio is said to represent the amount of motion between the two joints. For example, a ratio of 2.0 would indicate 2^{o} of motion at the distal joint for every 1^{o} of motion at the proximal. However, the motion may not be occurring at the same time during stance and therefore may not actually represent coupled motion.

Cross-correlation. Another method of analysis involves the use of cross-correlation to identify the relationship between angular displacement curves of the joint motions being investigated (Li & Caldwell, 1999; Pohl & Buckley, 2008; Pohl et al., 2006; Pohl, Messenger, & Buckley, 2007). However, this method only addresses the correlation of the movement and not magnitude, therefore one joint may be moving at twice the velocity of the other and the cross-correlation would not indicate such activity because they are moving at the same time.

Dynamic variables.

Continuous relative phase. An advanced method of dynamic assessment is continuous relative phase analysis, which provides temporal and spatial information (Hamill et al., 1999; van Emmerik & Wagenaar, 1996). This method plots the normalized angular position against the normalized angular velocity for each of two joints. Drawing a line from the origin to a given point along the stance phase plot produces an angle with respect to the horizontal. This is known as a phase angle (0-90^{o}) (see Figure 2.3). The difference between the phase angle of the two joints is the relative phase. With sinusoidal motions, this method describes coordination patterns well, but this is not the case more complex motion patterns including transitioning from walking to running (or running to walking) where relative phase patterns can vary largely (Diedrich & Warren, 1995; Heiderscheit et al., 2002). A continuous change in a control parameter (such as speed) may present discontinuity in the relative phase (Diedrich & Warren, 1995). Continuous relative phase analysis of non-sinusoidal motions can include a phase shift or low frequency oscillations in the signal (Peters, Haddad, Heiderscheit, Van Emmerik, & Hamill, 2003). Normalizing the data can remove this problem, but then temporal parameters between the two joints will not be reliable (Peters et al., 2003).   It is difficult to interpret continuous relative phase clinically because of the added velocity component (Chang et al., 2008; Miller et al., 2010).

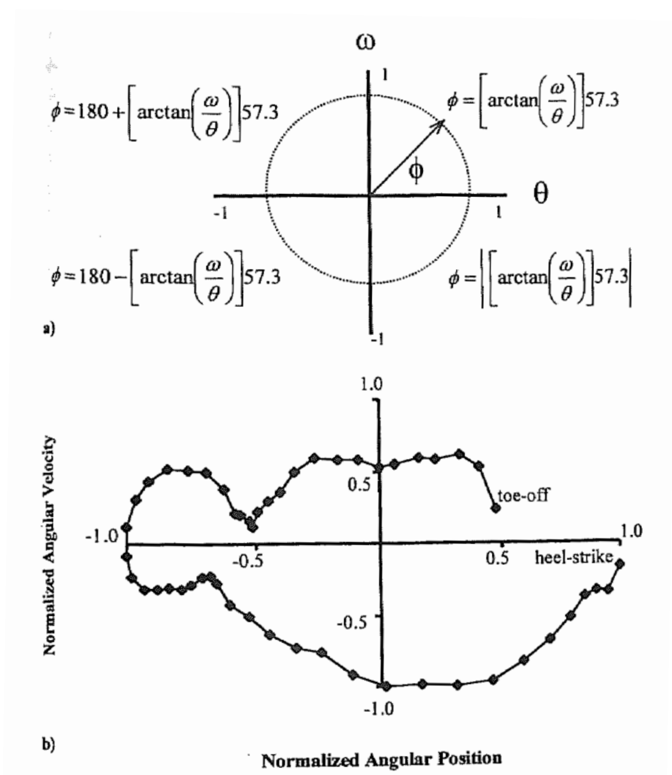

Figure 2.3. Phase plot describing phase angle from normalized angular position and velocity data (top) and a typical lower extremity joint coupling example (bottom). Reprinted with permission from Hamill et al.(1999).

Vector coding. Recent research involving complex kinematics of the lower extremity has also used vector coding methods (Ferber et al., 2005; Heiderscheit et al., 2002; Pohl & Buckley, 2008; Sparrow, Donovan, van Emmerik, & Barry, 1987). Angle-angle diagrams, or relative motion plots, plot the two joint motions in question against each other. The distal joint motion is often plotted on the y-axis and the proximal joint motion on the x-axis. By creating a vector between two adjacent points along the angle-angle diagram and orienting it with respect to the right horizontal, an angular orientation is obtained (see Figure 2.4). Vector coding angles of 45^o in each quadrant (namely 45^o, 135^o, 225^o, and 315^o) signifies a one-to-one ratio of movement between the two joint motions. It is not clear if a one-to-one ratio is optimal for all foot joint coupling relationship, though it has been theorized to be optimal for the couple between subtalar eversion and tibial internal rotation (McClay & Manal, 1997).

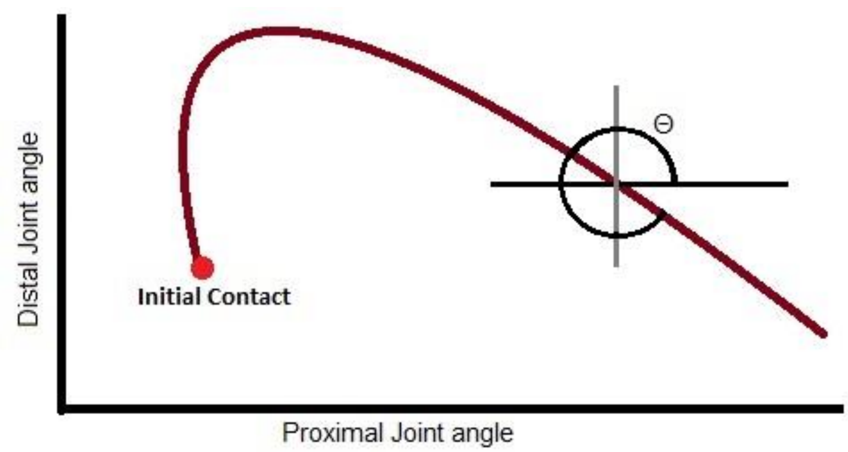

Figure 2.4. Example of an angle-angle plot with the vector coding angle (Θ) illustrated at a discrete point during the stance phase.

Vector coding angles can be evaluated at discrete points during the stance phase, or as a mean of the angles during a period of time, such as during the subphases of stance (Ferber et al., 2005). The terms anti-phase, in-phase, proximal phase, and distal phase have been used to characterize the location of the resulting vector angle from the horizontal (Chang et al., 2008). Anti-phase describes when the proximal and distal segment are moving opposite direction to one another (vector coding angle = 135^{o} or 315^{o}) while in-phase is when they are moving in the same direction (vector coding angle = 45^{o} or 225^{o}). When a vector coding angle is along the horizontal axis, the coupling pattern is in proximal phase, where only the proximal joint is moving, while an angle along the vertical axis would quantify a distal phase where the opposite is true. Chang et al. (2008) identified vector coding ranges to identify the classification of joint coupling vector coding angles when the angle does not lie directly over one of the axes or along a 45^{o} diagonal (Table 2.3). One disadvantage of vector coding over continuous relative phase is the exclusion of angular velocity. However, unlike continuous relative phase, it is able to compare joint angles to the original position signal and therefore more easily interpreted clinically (Miller et al., 2010; Peters et al., 2003). A recent comparison of the two methods determined that their results are not always similar and warns against comparing results from one method to that of the other (Miller et al., 2010).

Table 2.3. Joint coupling pattern catagorization based on vector coding angle (Θ) from the right horizontal.(Chang, Van Emmerik, & Hamill, 2008; Ferber, Davis, & Williams, 2005)

Coupling Pattern	Angle ranges
Anti-phase	$112.5^{o} \le \Theta < 157.5^{o}$, $292.5^{o} \le \Theta < 337.5^{o}$
In-phase	$22.5^{o} \le \Theta < 67.5^{o}$, $202.5^{o} \le \Theta < 247.5^{o}$
Proximal phase	$0^{o} \le \Theta < 22.5^{o}$, $157.5^{o} \le \Theta < 202.5^{o}$, $337.5^{o} \le \Theta < 360^{o}$
Forefoot phase	$67.5^{o} \le \Theta < 112.5^{o}$, $247.5^{o} \le \Theta < 292.5^{o}$

Joint coupling in the foot. Though joint coupling has been researched in running in both healthy and various pathological conditions (DeLeo et al., 2004; Ferber et al., 2005; Fisher, Bauer, Joshi, & Cobb, 2013; Hamill et al., 1999; Heiderscheit et al., 2002; McClay & Manal, 1997; Miller et al., 2010), few researchers have examined the joint coupling of the foot using MSFMs. Pohl et al. (2006) divided the foot into forefoot and rearfoot segments to track coupling characteristics of the midfoot joint with the rearfoot complex motion during running with different step widths in 12 active adults (≥ 2 hrs per week of exercise that involved running). In general for all of the step widths, the rearfoot dorsiflexed and everted during the first half of stance while the forefoot dorsiflexed and abducted. During the second half of the stance phase the rearfoot inverted and plantarflexed while the forefoot plantarflexed and adducted. In effort to evaluate the coupling characteristics of rearfoot and forefoot throughout the entire stance phase, cross-correlations between rearfoot eversion/inversion with forefoot plantarflexion/dorsiflexion, forefoot eversion/inversion, and forefoot abduction/adduction were performed. The results revealed high cross correlations (r) for rearfoot eversion/inversion with forefoot plantarflexion/dorsiflexion ($r < -0.85$) and forefoot abduction/adduction ($r > 0.94$), but not with forefoot eversion/inversion ($r < 0.32$). These coupling patterns were slightly out of phase as rearfoot inversion (40.4 to 47.7% of stance) began prior to forefoot plantarflexion (52.4 to 59.3% of stance) and adduction (46.0 to 50.5% of stance).

In a similar study, Pohl et al. (2007) compared rearfoot-forefoot coupling patterns over different speeds and modes of gait (walking and running) in 12 active adults who were all RFS runners. Besides the coupling between rearfoot eversion/inversion with forefoot eversion/inversion (walking, $r = 0.41$; running, $r = 0.15$), running always had higher cross correlations than walking (rearfoot eversion/inversion with forefoot plantarflexion/dorsiflexion, walking $r = -0.80$; running $r = -0.96$; rearfoot eversion/inversion with forefoot abduction/adduction, walking $r = 0.91$; running $r = 0.97$). Vector coding revealed high coupling during running between rearfoot eversion/inversion and forefoot plantarflexion/dorsiflexion (medium run = 40.3 ± 5.0^{o}; fast run = 40.6 ± 5.2^{o}), but increased forefoot abduction/adduction motion when coupled with rearfoot eversion/inversion (medium run = 24.4 ± 2.6^o; fast run = 24.5 ± 3.1^o).

Joint coupling with different foot strike patterns. In a study comparing forefoot and rearfoot coupling patterns of 12 active runners while running with RFS, FFS, and toe running patterns, Pohl and Buckley (2008) found similar cross-correlation coefficients (*r*) between rearfoot eversion/inversion coupled with forefoot motion in all three planes of motion (forefoot plantarflexion/dorsiflexion, *r* between -0.96 to -0.86; forefoot abduction/adduction, *r* between 0.92 to 0.96; forefoot eversion/inversion, *r* between -0.20 to -0.06). Interestingly, the FFS running pattern cross-correlations for all of the rearfoot and forefoot motions were slightly higher than those for RFS running. In addition to the cross-correlations, vector coding methods were used to identify the magnitude of changes between the joints. Mean vector coding angles were calculated between rearfoot eversion/inversion and forefoot plantarflexion/dorsiflexion (44.8 ± 5.1^o FFS, 39.9 ± 4.0^o RFS), forefoot abduction/adduction (24.5 ± 4.7^o FFS, 21.7 ± 4.0^o RFS), and forefoot eversion/inversion (27.4 ± 5.6^o FFS, 28.4 ± 5.2^o RFS). Coupling angle curves revealed a decreased joint coupling angle during the first subphase of stance in the RFS condition (as evaluated by visual inspection of coupling angle-percent stance plots), indicating a greater movement at the proximal joint relative to the distal joint. Because of the strong coupling between the forefoot and rearfoot motion, Pohl and Buckley (2008) suggested midfoot coupling may be important in transferring movement along the foot and up the kinetic chain. They also postulated the idea that the dorsiflexed forefoot at initial contact when running with a RFS may alter the talonavicular articulation. Such an alteration would influence the transfer of motion to the rearfoot and ultimately to the tibia which would result in decreased internal rotation excursion (Hintermann, Nigg, Sommer, & Cole, 1994). None of the vector coding angles, however, were compared for significant differences. Eslami et al. (2007) compared forefoot and rearfoot joint coupling in 16 male RFS runners during barefoot and shod conditions and found only at heel-strike was there a difference in mean phase angle (stance subphases examined were heel-strike, 0%; foot-flat, 5-25%; heel-rise, 25-50%; push-off, 50-75%; toe-off, 75-95%) between rearfoot eversion/inversion and forefoot abduction/adduction (71.5 ± 45.4^{o} barefoot, 34.5 ± 28.2^{o} shod). No phase angle differences in any other subphases or with the sagittal or frontal planes of the forefoot were seen. Additionally, during barefoot running subjects demonstrated a greater coupling relationship during the first 5% of stance than during any other subphase of stance. Shod running, however, had a greater coupling relationship for push-off (50-75% of stance) and toe-off (75-95% of stance) than for heel-rise (25-50% of stance).

One limitation of these studies is the lack of quantifying midfoot motion as well as medial and lateral motions of the forefoot. Only recently has research begun to look at joint coupling within the foot using more sophisticated foot modeling (Fisher et al., 2013; Seneli et al., 2013). Seneli et al. (2013) compared foot coupling from barefoot running to that of running in a sandal without arch support in 10 active adults and observed strong coupling using vector coding methods between rearfoot eversion/inversion with calcaneocuboid eversion/inversion during subphase 1 and 4 of stance while barefoot (44.40 ± 14.16^{o}, 49.61 ± 14.50^{o}, respectively) and between subphase 3 and 4 (45.21 ± 13.77^{o}, 47.27 ± 16.07^{o}, respectively) while shod. Coupling between calcaneonavicular inversion/eversion and medial forefoot plantarflexion/dorsiflexion also showed strong coupling during subphases 1 and 3 while shod

(41.48 ± 12.15°, 42.76 ± 15.15°, respectively). Calcaneocuboid eversion/inversion with lateral forefoot plantarflexion/dorsiflexion also showed strong coupling during subphase 3 while shod (40.77 ± 13.58°). These results illustrate the midfoot-forefoot coupling patterns that contribute to the kinematics of the running stance phase as well as differences in coupling patterns based on footwear. It is still unknown, however, to what extent the midfoot and medial/lateral forefoot contribute to different foot strike patterns. Additionally, it is unknown what coupling characteristics are ideal within the foot to help prevent injury and improve performance..

***Variability in joint coupling*.** Some researchers have begun focusing more on the variability of joint motion and coupling rather than the coupling direction angles. In engineered systems, variability is usually undesirable and treated as noise, however, variability in biological systems may be a demonstration of health. Hamill et al. (1999) introduced the application of the Dynamical Systems Theory to lower extremity running injuries based on previous work done in movement coordination and stability (Haken et al., 1985; Schoner & Kelso, 1988). It is based on the idea that regular movement patterns result in a loss of adaptability, which may increase the risk of injury by not allowing the body to adapt to changes in the environment or different tasks. Supporting this theory, less joint coupling variability has been demonstrated in individuals with patellofemoral pain and anterior cruciate ligament deficient knees compared to healthy controls (Hamill et al., 1999; Heiderscheit et al., 2002; Moraiti, Stergiou, Vasiliadis, Motsis, & Georgoulis, 2010). The decreased variability in movement patterns may also lead to overloading of a specific area of tissue instead of distributing the stress to different areas, resulting in injuries associated with overload such as osteoarthritis. Though variability in movement patterns have been quantified in the larger joints of the lower extremity, variability within the segments of the foot has not been investigated.

***Summary*.** It has been established that joint coupling characteristics are important for running kinematics and may contribute to musculoskeletal injury. Most research in this area has focused on coupling of rearfoot eversion/inversion with tibial rotation, negating the effect the joints of the foot have on the rearfoot (Heiderscheit et al., 2001; McClay & Manal, 1997). However, it has been established that the joint couplings within the foot may affect rearfoot motion, specifically eversion/inversion, and that the foot strike pattern further affects this motion (Pohl & Buckley, 2008; Pohl et al., 2006, 2007). As identified by Pohl and Buckley (2008), FFS running resulted in larger cross-correlations of joint coupling than RFS running which may indicate a more synchronous movement between forefoot and rearfoot while FFS running. To further understand the extent of this relationship, a more sophisticated foot model is needed to further understand how the foot contributes to rearfoot motion. Lacking in MSFM joint coupling and foot strike literature is the variability in consecutive steps while running as it may indicate how each foot strike pattern functions to prevent overuse injuries.

Leg Muscle Activation

***Electromyography*.** Peripheral neuromuscular activity has been researched using various methods, among which electromyography (EMG) is the most common. Surface electrodes properly placed over muscle are able to identify small changes in voltage caused by action potentials of the underlying motor units. The voltage changes are used as an indicator of the muscle’s activity (Gerdle, Karlsson, Day, & Djupsjöbacka, 1999). There are various EMG signal analyses researchers have used to further understand muscle activity level. Motor pattern recruitment is often evaluated by identifying events when the muscle is turned on. This can be done by identifying the onset and offset of the EMG signal. Because of the amount of noise that is produced while capturing an EMG signal, adequate methods for identifying the onset of a muscle are necessary to avoid error. Visual methods of EMG onset have been used as well as computer algorithms in order to determine when a muscle becomes active, however, there is currently no consensus regarding the best method (Ebig, Lephart, Burdett, Miller, & Pincivero, 1997; Hodges & Bui, 1996). For computer-based methods, procedures vary from setting a standard threshold (Tomberg, Levarlet-Joye, & Desmedt, 1991; Zhou, Lawson, Morrison, & Fairweather, 1995) to using standard deviations or a percentage of the EMG signal to identify a threshold (Johnson & Johnson, 1993; McKinley & Pedotti, 1992; Morey-Klapsing, Arampatzis, & Bruggemann, 2004).

In addition to identifying when a muscle is active, the intensity of the muscle activity, or contraction, can be identified by the amplitude and frequency of the EMG signal. After removing all negative EMG values through rectification of the signal, amplitude can be evaluated in many different ways including maximum values, means, medians, integrated values, or root-means-square values (Gerdle et al., 1999; Ishikawa, Pakaslahti, & Komi, 2007; Kyrolainen, Avela, & Komi, 2005; Shih et al., 2013). The root-mean-square analysis has been recommended by Basmajian and De Luca (1985) and commonly used in research (Gerdle et al., 1999; Mizrahi et al., 1997; Shih et al., 2013). It also has reduced amplitude cancellation compared to other EMG averaging methods (Keenan, Farina, Maluf, Merletti, & Enoka, 2005). Morey-Klapsing et al (2004) compared various onset detecting algorithms and integrated EMG signals and concluded that the detection of muscle onset alone does not give adequate information as to whether or not the muscle is contracting sufficiently to produce adequate force, but rather onset and integrated EMG should be used in combination to better understand the muscle preactivation characteristics.

***EMG while running*.** Because of the variation in foot position at initial contact associated with the different strike patterns, leg musculature activation may also be different. In a FFS running pattern, the gastrocnemius and soleus complex undergo a much larger eccentric load as they attempt to gradually control the lowering of the calcaneus down to make ground contact. In a RFS running pattern, on the other hand there is a need to support the foot in a dorsiflexed position and lower the forefoot to the ground during the first subphase of stance, requiring more activity from the tibialis anterior.

*Preactivation***.** Preactivation of a muscle during gait is often evaluated as an indicator of how the muscle "prepares" for the stance phase and is identified as the amount of EMG activity during the swing phase immediately prior to initial contact. Because of electromechanical delay, the musculature of the lower extremity must preactivate in preparation for stabilization during landing and to avoid injury (Konradsen & Ravn, 1990; Zhou, 1996). In RFS running, the tibialis anterior needs a higher level of preactivation in order to maintain the ankle in dorsiflexion and then eccentrically lower the forefoot to the ground. The main EMG activity of the tibialis anterior appears to be during the 100 ms leading up to and following initial ground contact, peaking at heel contact (Dietz, Schmidtbleicher, & Noth, 1979; Elliott & Blanksby, 1979; von Tscharner, Goepfert, & Nigg, 2003). In FFS running, the gastrocnemius acts to assist in lowering the heel and shock absorption during subphase 1 of stance and would therefore benefit from adequate muscle stiffness. Shih et al. (2013) compared EMG of the tibialis anterior and gastrocnemius in 12 habitually shod runners while running with a RFS and FFS in both barefoot and shod conditions. While shod, the tibialis anterior had a higher preactivation (assessed 50 ms before initial contact to initial contact) while running with a RFS (48.59 ± 11.5 % maximal voluntary contraction (MVC) than with a FFS (16.79 ±7.3 % MVC). The opposite was true for the gastrocnemius, with increased preactivation in the FFS (61.96 ± 13.5 % MVC) compared to the RFS (13.83 ± 7.9 % MVC). Similar differences were also seen when barefoot running with a FFS and RFS (Shih et al., 2013). Jacobs et al. (1993) compared mono- and biarticular muscle activation in seven elite male FFS runners and found high levels of gastrocnemius activation prior to initial contact which continued during lengthening of the gastrocnemius, allowing for maximal utilization of the stretch-shortening cycle. The mono-articular soleus did not have high levels of activation prior to initial contact.

Footwear. There is also variation in how footwear alters neuromuscular activity in running. While running with a RFS, barefoot running has been reported to have higher levels of preactivation in the gastrocnemius, and soleus muscles than RFS running while shod (Divert et al., 2005b; Shih et al., 2013). When normalized to percent of maximal voluntary contraction (MVC), the medial and lateral gastrocnemius have a 1-3% increased activation in barefoot running when measured using surface EMG. Olin and Gutierrez (2013) observed increased average medial gastrocnemius activity when 18 habitual shod RFS runners (run per week = 20.9 ± 6.0 km) ran for the first time barefoot with their normal RFS pattern and a further increase in average activity when running barefoot with a FFS (Divert et al., 2005b; von Tscharner et al., 2003). Komi et al. (1987) observed preactivation EMG patterns of four active males in various footwear conditions and identified a strong correlation between the lateral gastrocnemius preactivation amplitude and VGRF impact peak magnitude. This correlation was attributed to the need to stiffen the leg extensor muscles in order to buffer the initial impact. The study did not specify foot strike pattern, but barefoot running was associated with an increased impact peak versus shod running, therefore, a RFS pattern may be assumed. Wakeling et al. (2001) attempted to identify muscle activity that corresponded to impact on the foot during running with repetitive striking of a pendulum on the calcaneus while supine, reproducing running-like forces, but minimizing the role of the muscles in joint motion. In a study of 20 active males, they found similar results in that muscle activity increased as a result of increased rate of loading in both the tibialis anterior and medial gastrocnemius. EMG of the tibialis anterior muscle has evidence of increased pre-activation intensity with shod running compared to running barefoot. After initial contact there is more of a delayed response in tibialis anterior activity when running shod compared to barefoot as found in 40 male runners (>25 $km \cdot wk^{-1}$) (von Tscharner et al., 2003). O'Connor and Hamill (2004) analyzed the effect of varus and valgus shoe wedges in tibialis anterior, peroneus longus, medial and lateral gastrocnemius, and soleus EMG in 10 recreationally active males (all RFS) and failed to observe any differences in EMG activity based on the different calcaneal varus and valgus positioning at initial contact. Nigg et al. (2003) tested 20 male runners in two different shoe conditions, one of medium hardness and elastic and the other softer and more viscous, and likewise found no difference in the onset timing of the muscles or preactivation intensity of the medial gastrocnemius or tibialis anterior.

Spatiotemporal parameter changes. In addition to the difference in muscle activation due to change in foot contact posture, alterations in muscle activity have been shown to exist with changes in spatiotemporal parameters. Chumanov et al. (2012), in a study of 45 recreational runners, found that an increase in cadence of 10% over preferred cadence increased medial gastrocnemius muscle activity during late swing/preactivation, but attributed the difference to a change in foot posture, though ankle angle was not reported. Likewise, increases in speed result in increased EMG activity of the gastrocnemius and tibialis anterior, likely due to an increase in motor unit recruitment (Chumanov et al., 2012). Because of potential EMG changes due to changes in speed and cadence, it is important to control for these parameters or to acknowledge that differences between conditions or subjects may be due to differences in spatiotemporal parameters. When specifically looking at RFS and FFS running, it is already known that FFS running tends to be performed with a higher step cadence (Bonacci et al., 2013; Divert et al., 2008; Nunns et al., 2013; Shih et al., 2013; Squadrone & Gallozzi, 2009). Therefore, even at a similar speed of running, the step rate must be taken into consideration when looking at differences in EMG profiles. In the present study, though step rate will not be controlled, it will be measured and compared for differences. It is still unknown if similar EMG differences in step rate will be seen between FFS and RFS runners.

Fatigue on EMG. Fatigue has also been shown to both have an effect and no effect on EMG activity. Some of the discrepancy may be associated with defining fatigue and the type of fatigue elicited in the protocol (Gandevia, 2001). With prolonged submaximal activity, an increase in EMG activity is most likely due to recruitment of more, larger motor units (Bigland-Ritchie, Furbush, & Woods, 1986; Petrofsky et al., 1979). EMG activity has also, however, been shown to decrease during running activities, likely a result of decreased conduction velocity (Mizrahi et al., 2000a; Petrofsky et al., 1979; Wu et al., 2007). Wu et al. (2007) tested 29 male

Army Infantry students while running 20 minutes at a 12 km·hr^{-1} pace and found an increase in EMG amplitude of the medial gastrocnemius and tibialis anterior during minutes 10 and 15 of the run over activity at minute five. At minute 20, however, the EMG amplitude decreased to be similar to that observed at the fifth minute. The researchers attributed the initial increase in EMG amplitude to recruitment of larger motor units and the final decrease to fatigue of the larger muscle fibers in those muscles. No change was seen in the peroneus longus muscle for the duration of the 20 minutes. Mizrahi et al. (2000a) had 14 male, recreational runners (8-10 km·wk^{-1}) run for 30 minutes at their anaerobic threshold and found no significant change in gastrocnemius normalized integrated EMG, but a decrease in tibialis anterior EMG activity after 20 minutes. Reaching metabolic exhaustive levels, such as a run based on anaerobic threshold or oxygen consumption level will not necessarily result in fatigue characteristics at the muscle level. It is therefore evident that the fatigue status of an individual can affect the EMG profile while running both during and towards the end of a run. There are no known studies comparing the effect of an exhaustive run on EMG of leg muscles between FFS and RFS runners. It is therefore unknown if the preactivation of the leg muscles will see similar changes between the foot strike patterns after an exhaustive run.

Summary. Muscular control influences joint motion and variability while running. For performance and prevention of injury, activation prior to initial contact for each stance phase while running is necessary for leg musculature. The onset and magnitude of a muscle's preactivation are dependent on running strike pattern, spatiotemporal parameters, and footwear. Because of the kinematic differences of FFS and RFS runners at initial contact and during subphase 1 of running, their activation needs are most likely to be different, but it is still unknown what differences for preactivation are present and how fatigue may change these neuromuscular characteristics.

Fatigue

Most running research comparing kinematics and joint coupling are performed when subjects have completed little more than a warm-up. However, musculoskeletal injury rarely occurs at the beginning of a run. Most running related injuries occur after an individual has experienced some level of fatigue (Mizrahi et al., 2000a; Verbitsky et al., 1998). When muscles are functioning appropriately, they control motion of joints and attenuate dynamic loads (Radin, 1986). Overuse injury may result from repetitive motion or altered kinematics caused by altered muscular contraction which is often observed with fatigue. Many researchers have disagreed on the specific definition of fatigue (Ament & Verkerke, 2009; Gandevia, 2001; Gandevia, Allen, & McKenzie, 1995) but they tend to agree on the general idea that it is “any exercise-induced reduction in the force generating capacity” of muscle (Gandevia, 2001). The type and mode of exercise performed as well as environmental factors (e.g. temperature, altitude), individual characteristics (e.g. level of expertise, genetics), muscle group used, type of muscle action , and intensity level of exercise all contribute to the onset and location of fatigue (Bazett-Jones, 2012). Beyond alterations in physiological systems such as with neuromuscular control and chemical changes at the muscle level, an individual’s perception and sensation of fatigue also affects their overall ability to continue an exercise task (Gandevia, 2001). The commonly used Borg scales (Borg, 1970) are a subjective measure of one’s fatigue level which incorporates both the physiological and psychological components of fatigue.

***Peripheral and central fatigue*.** Fatigue itself occurs from both peripheral (distal to the neuromuscular junction) and central (proximal to the neuromuscular junction) factors. In healthy individuals, peripheral forms of fatigue may occur as a result of metabolite accumulation (Holloszy & Coyle, 1984; Lamb & Stephenson, 1994; Tesch & Karlsson, 1984), depletion of glycogen stores (Coyle, Coggan, Hemmert, & Ivy, 1986; Gollnick, Armstrong, Sembrowich, Shepherd, & Saltin, 1973), and hypoxia (Richardson, Noyszewski, Kendrick, Leigh, & Wagner, 1995; Romer, Haverkamp, Lovering, Pegelow, & Dempsey, 2006). These peripheral mechanisms may also have an effect on central fatigue as their influence can occur at both the muscle level and throughout the body. Because of the many levels within the nervous system, exercise may affect any number of factors causing central fatigue. Many different physiological components such as core temperature (Nybo & Nielsen, 2001) and pain (Lund, Donga, Widmer, & Stohler, 1991) provide sensory signals to the brain to alter neuromuscular control. Likewise, psychological factors can equally effect CNS function (Ament & Verkerke, 2009).

***Theories of fatigue*.** Various theories have been developed to help understand how the body fatigues during exercise. Fatigue is very task dependent, and therefore, different fatigue models may better represent how and why fatigue is occurring during a specified activity (Gandevia, 2001; Weir, Beck, Cramer, & Housh, 2006). Two theories that have been described to be best associated with endurance tasks are the teleoanticipatory (Ulmer, 1996) and central governing theories (Noakes, St Clair Gibson, & Lambert, 2004). The teleoanticipatory theory suggests that the body uses a feedback mechanism to constantly monitor the metabolic rate and the estimated metabolic cost to complete the task and alters output accordingly. The central governing model proposes that the central nervous system is able to construct a feedforward mechanism using sensory information from the periphery to create an exercise "plan", using the physiological abilities of the individual, to execute a specific task. However, activities lasting less than one minute, such as a Wingate power test on a cycle ergometer, see a steady decrease in performance output and no "strategy" of pacing is used for the highest power output. In addition to the two theories above, the catastrophic failure model recognizes that if one or more physiological systems are stressed beyond their capacity, exercise is reduced or stopped (Edwards, 1983; Noakes et al., 2004). From another view point, Ament and Verkerke (2009) have suggested that fatigue be separated into tasks that mostly require either type I or type II muscle fibers because of the complexity that different muscle fiber types have on a specific task or their ability to resist fatigue. However, though some tasks tend to preference one muscle fiber type over another, seldom does a task use only one muscle fiber type.

***Fatigue and running*.** Despite the cause or level of fatigue, research has succeeded in showing that kinematic changes occur when running in an exhausted or exerted state (Derrick et al., 2002; Dierks et al., 2010; Donahue & Sharkey, 1999; Milgrom et al., 2007; Mizrahi et al., 2000b; Mizrahi et al., 1997). Though changes seen in an exerted state are within normal limits, the repetitive nature of running may allow the accumulation of minor differences to overload tissue (Hreljac et al., 2000). As muscles contribute greatly to the dissipation of forces while running (Bobbert, Yeadon, & Nigg, 1992; Radin, 1986), the inability of muscles to optimally perform their task will result in increased forces on other tissues. For example, research has identified imbalances in leg muscle EMG patterns and increased tibial acceleration following an exhaustive run which may increase the risk of stress fractures (Mizrahi et al., 2000a). Furthermore, exhaustive exercise bouts have also shown to change the pre-activation timing of leg muscles, which may also contribute to alterations in joint kinematics and coupling, especially at initial contact and during initial loading (subphase 1) of stance (Nyland, Caborn, Shapiro, & Johnson, 1997).

***Fatigue Protocols*.** Various methods of achieving an exhaustive or fatigue state have been used in running fatigue research. To investigate peripheral fatigue, protocols have fatigued muscle groups with concentric and eccentric resisted motions prior to collecting running data (Brüggeman, 1996; Christina, White, & Gilchrist, 2001; Ferber & Pohl, 2011). Central fatigue protocols have focused more on fatigue from exhaustive running bouts. Davis et al. (1999) used a high-intensity running protocol consisting of shuttle runs to fatigue, to mimic activity of soccer, basketball, and hockey players. Other running studies looking at endurance running, have used techniques based on an individual's anaerobic threshold (Verbitsky et al., 1998), time trials (Derrick et al., 2002), and incremental exercise to volitional exhaustion (Nyland et al., 1997). Verbitsky et al. (1998) conducted a 30 minute run at anaerobic threshold of 22 adult male recreational athletes and divided them into groups of those fatigued and those not fatigued based on end-tidal carbon dioxide pressure which decreases when fatigued. To assess level of fatigue that is more representative of a typical run, many researchers have used protocols requiring subjects to perform a run of specified length such as a 10 kilometer run (Elliot & Ackland, 1981; Paavolainen, Nummela, Rusko, & Häkkinen, 1999), marathon (Kyrolainen et al., 2000; Nagel, Fernholz, Kibele, & Rosenbaum, 2008; Nicol, Komi, & Marconnet, 1991a, 1991b; Ross, Middleton, Shave, George, & Nowicky, 2007), or ultra-marathon (Morin, Samozino, & Millet, 2011). Some of the specified length runs are completed at a predetermined speed, or constant velocity, while others are performed at the subjects' self-selected pace. In studies that utilize a predetermined speed, participants may not acquire the level of fatigue that the individual would typically experience during a regular run or they may fatigue too quickly. Bruggeman (1996) conducted three consecutive experiments comparing various methods of fatigue in untrained runners (< 15 $km \cdot wk^{-1}$). They investigated the various ways of inducing fatigue at both the central and peripheral levels. The first experiment required two separate 30 minute runs by the subjects, one below anaerobic threshold (< 5 mmol lactate) and one above (> 5 mmol lactate). The second experiment involved a resistive dynamic ankle motion of concentric inversion/adduction with eccentric eversion/abduction repeated as much as possible in a 2 minute time period. The final protocol required a run (2.6-3.0 $m \cdot s^{-1}$) to volitional exhaustion. Local muscle strength as measured dynamic leg press (concentric) and ankle inversion and adduction (concentric/eccentric) protocols decreased for all the protocols. Additionally, rearfoot inversion at initial contact was decreased during the resistive dynamic motion protocol and the exhaustive run. When comparing the rearfoot kinematics over the course of the exhaustive run (approx. 45 min), the rearfoot sagittal plane kinematics and VGRF appeared to be continually changing for the first 7 minutes of the run (about 3^o) while subjects adjusted to the treadmill and then after 12-15 minutes significant changes (about 3-5^o) due to fatigue were observed. As many running studies do not have a lengthy running protocol, it is important to note that their kinematics may not indicate normal kinematics mid- and late- run.

Neuromuscular characteristics also changed as a result of fatigue from the different protocols. Mean power frequencies shifted in both leg and thigh musculature by increasing initially, indicating the recruitment of fast-twitch muscle fibers, and decreasing, as all fibers fatigued. With the resistive dynamic motion protocol, a time shift in the EMG activity was evident after fatigue, indicating a later onset and time to maximum peak in the tibialis anterior muscle.

***The effect of fatigue on running kinematics*.**

*Ankle***.** Changes in ankle and foot kinematics as a result of an exhaustive run include increased ankle inversion at initial contact (Derrick et al., 2002) and peak eversion during stance (Clansey, Hanlon, Wallace, & Lake, 2012; Dierks et al., 2010; VanGheluwe & Madsen, 1997). Just prior to volitional exhaustion (when runner could no longer continue running at treadmill pace) while running at 4.5 $m \cdot s^{-1}$, Van Gheluwe et al. (1997) observed increased peak eversion (2.1^o difference) and eversion excursion (1.8^o difference) during stance in 20 physical education students (20-29 years old), all of which were RFS. Similar results were found after a 3200 m time trial of 10 recreational runners with small, but significantly different peak eversion (1.1^o difference) and eversion excursion (1.4^o difference) during the stance phase at the end of the run (Derrick et al., 2002).   The foot strike pattern used by the runners, however, was not identified. In a study of 20 recreational runners (RFS) that ran to exhaustion (85% HR_{max}, or RPE ≥ 17), Dierks et al. (2010) had observed the greatest kinematic changes in rearfoot eversion, compared to knee and hip kinematics with an increased excursion of 1.2^o and peak eversion of 1.5^o during the stance phase. After a local muscular fatigue protocol of the ankle invertors and dorsiflexors, Christina et al. (2001) observed decreased ankle inversion (2.2^o difference) and dorsiflexion (3.2^o difference) respectively at initial contact in RFS runners . Likewise, Kellis and Liassou (2009) fatigued the ankle dorsiflexors/plantarflexors, using an isokinetic protocol until only 30% of peak torque could be produced by the ankle, and also found decreased dorsiflexion at initial contact (5^{o} difference).

*Foot***.** As previously stated, few researchers have investigated kinematic and joint coupling changes within the foot via a MSFM while running. Even fewer have considered changes in distal foot function that occur following an exhaustive run. Research investigating changes in plantar pressure from running suggests alterations in foot kinematics due to exhaustion. Nagel et al. (2008) identified a shift in plantar pressure from the toes to the forefoot in 200 marathon runners while walking barefoot immediately after completing a marathon, suggesting changes in stance phase foot motion while walking after an exhaustive run. The study did not, however, evaluate the changes while running. While running, Willems et al. (2012) monitored foot pressures after a 20 km race in 52 racers and also found differences in mean and peak forces under the medial heel, metatarsals, and toes. No other research has examined the motion occurring within the foot after an exhaustive run despite the fact that the plantar pressure studies suggest that changes in kinematics may occur. Elliot and Ackland (1981) recognized that foot mechanics have the greatest influence on running mechanics, yet these mechanics have yet to be analyzed after an exhaustive run. This may be especially important due to the fact that this is time when injuries commonly occur. Additionally, the research that has compared kinematics before and after an exhaustive run has failed to include FFS runners.

***Summary*.** From previous research it is known that kinematics, kinetics, and neuromuscular characteristics change as a result of both central and peripheral fatigue. Rearfoot inversion at initial contact (Derrick et al., 2002), eversion excursion, and peak eversion increase as a result of running fatigue (Clansey et al., 2012; Dierks et al., 2010; VanGheluwe & Madsen, 1997) and the initial contact angle of ankle dorsiflexion decreases in RFS runners (Brüggeman, 1996). As suggested by foot plantar pressure data (Nagel et al., 2008; Willems et al., 2012), kinematic changes may also occur distal to the rearfoot complex and may possibly be the cause of rearfoot changes. It is still unknown what kinematic changes occur in the distal foot and how foot and ankle kinematics change in FFS runners. Fatigue also affects leg musculature by decreasing the time of preactivation of the tibialis anterior (Brüggeman, 1996). These affects have yet to be determined in FFS runners and further investigation into preactivation patterns of the plantarflexors and everters of the foot have not been done during fatiguing protocols. These patterns may be crucial in foot and ankle stability during stance especially during initial loading (subphase 1) when the inverted foot is particularly vulnerable to sprains.

3. MULTI-SEGMENT FOOT KINEMATICS IN HABITUAL FOREFOOT AND REARFOOT RUNNERS AT THE BEGINNING AND END OF AN EXHAUSTIVE RUN

Introduction

An estimated 19.4-79.3% of runners experience some type of overuse injury each year (Bahr & Holme, 2003; van Gent et al., 2007). Despite changes in footwear to combat these injuries, there is little evidence that overuse injury rates have decreased. As a result, some researchers have begun to focus on the role of foot strike patterns in running injuries (Lieberman et al., 2010; Williams et al., 2000).

Most runners (75-99%) utilize a rearfoot strike (RFS) pattern (Hasegawa et al., 2007; Larson et al., 2011). While fewer runners utilize a midfoot strike (MFS) (0-23.7% of runners) or forefoot strike (FFS) (0-2% of runners) pattern (Hasegawa et al., 2007; Larson et al., 2011), some researchers have hypothesized that these styles are a more “natural” form of running. These researchers have also argued that RFS running, which emerged because of modern footwear, may increase injury risk. Midfoot and FFS running are associated with greater knee flexion and ankle plantarflexion at initial contact compared to RFS running (Shih et al., 2013; Stackhouse et al., 2004). Although overall range of motion at the knee and ankle during the entire stance phase does not appear to differ significantly between the foot strikes, differences during early stance have been observed (Pohl & Buckley, 2008). These kinematic differences contribute to decreased vertical ground reaction force loading rates reported with MFS and FFS versus RFS patterns (Lieberman et al., 2010; Williams et al., 2000) and may be significant given the common clinical belief that higher loading rates are associated with increased risk of injury.

Although many motion analysis studies have investigated ankle and knee kinematic differences between the foot strike patterns, the effect on foot kinematics has received little attention (Pohl & Buckley, 2008; Pohl et al., 2006). Differences in weight distribution and centers of pressure location between the foot strike patterns (De Wit et al., 2000) create an increased demand on the distal foot to provide stability and dissipate force following initial contact during mid/forefoot running. Pohl and Buckley (2008) compared FFS and RFS foot kinematics using a two-segment foot model (forefoot and rearfoot) and observed differences in forefoot dorsiflexion and abduction excursion during stance. However, important differences between the foot strike patterns may have been masked in this study due to the use of an over-simplified foot model. Wolf et al. (2008) and Rouhani et al. (2011), have recommended partitioning of the foot into rearfoot, midfoot, and medial and lateral forefoot segments. To date, no study has used these recommended segments to evaluate foot motion between the foot strikes.

Furthermore, the majority of studies investigating foot strike patterns have only investigated runners in a non-fatigued state. However, most running injuries do not occur at the beginning of a run, rather after an individual has experienced some level of fatigue (Mizrahi et al., 2000a). The limited number of studies that have used exhaustive protocols have reported altered ankle kinematics not only between differing foot strikes, but also as the result of fatigue (Dierks et al., 2010). The effect of a prolonged run on foot kinematics in runners with different foot strike patterns, however, has not been investigated.

The purpose of this investigation was to identify the differences in foot kinematics between habitual RFS and FFS runners at the beginning and end of an exhaustive run. It was hypothesized that significant kinematic differences would be seen between the groups at initial contact and in distal foot range of motion during early stance based on different initial contact points. Additionally, it was hypothesized that the angular displacements of the functional articulations in the foot and ankle would increase as a result of muscular fatigue in both groups throughout stance following the exhaustive run. The increase was hypothesized to be greater in FFS vs RFS runners due to the increased use of the plantarflexors associated with making initial contact with the mid/forefoot (Ahn et al., 2014).

Methods

Participants

Fifteen habitual RFS runners were age and gender matched with 15 habitual midfoot/FFS runners (Table 3.1). All subjects ran a minimum of 10 miles per week and used their current foot strike pattern for a minimum of one year at the time of participation. Participants were excluded if they had a lower extremity injury within 6 weeks, history of lower extremity surgery, currently used custom molded foot orthotics, or any known cardiovascular problems or uncontrolled asthma. Participation in the study included an initial phone screen and two in-person visits consisting of a foot strike screen and a running gait analysis. All procedures and risks associated with the study were explained to the participants. Participants read and signed an informed consent form approved by the institution's Institutional Review Board.

Procedure

Foot strike screen

Foot strike patterns were confirmed using visual analysis of each participant's vertical ground reaction force profile during overground running (Altman & Davis, 2012a; Hasegawa et al., 2007). The ground reaction force data was captured using an AMTI force plate (Advanced Medical Technologies, Inc, Waterford, MA) sampling at 1000 Hz embedded in a 25 m runway. A RFS was defined as a vertical ground reaction force time-series with two distinct peaks during stance. A FFS was defined as a time-series that did not have two distinct peaks. Midfoot and FFS were combined into the FFS group since both have similar vertical ground reaction force patterns (Lieberman et al., 2010). Subjects completed five running trials along the carpeted walkway in their own running shoes. At least three trials had to have a consistent vertical ground reaction force time-series to be placed in a footstrike group.

Participants were then provided with a pair of 12 mm neutral running sandals (Mono, Luna Sandals, Seattle, WA) and a 10-day accommodation program (Appendix M) (Bonacci et al., 2013; Squadrone & Gallozzi, 2009). The accommodation program required participants to slowly incorporate sandal usage into their habitual training without decreasing their training mileage.

Multi-segment foot model

Clusters of four retroreflective technical markers (6.4 mm diameter) were placed on the right leg and foot to identify seven segments (Bauer, 2012) and six functional articulations (Figure 3.1). Technical markers were either directly adhered to the skin or as wand clusters using liquid adhesive (Mastisol Liquid Adhesive, Ferndale Laboratories, Inc., Ferndale, MI) and double-sided toupee tape. Elastic tape (Elastikon, Johnson & Johnson, New Brunswick, NJ) and a cohesive bandage (Powerflex, Andover Healthcare, Inc., Salisbury, MA) were used to further secure wands to skin. Three-dimensional positional marker data were collected using a 10-camera Eagle Motion Analysis System (Motion Analysis Corp., Santa Rosa, CA) sampling at 200 Hz. An anatomical calibration procedure was performed to identify the three-dimensional position of additional anatomical landmarks in order to define local coordinate systems for each segment (Appendices A and C) (Grood & Suntay, 1983).

Exhaustive run and gait assessment

Following the anatomical calibration procedure subjects warmed up for 5 minutes on a treadmill. After the warm up, subjects ran at a speed of 3.4 $m \cdot s^{-1}$ (7.5 mph) until a consistent gait pattern was observed. Motion capture was then collected for 10 s. Each subject then performed a maximal effort 5 km run (Laursen et al., 2007). During the maximal effort run subjects were able to alter the treadmill speed as necessary. The exhaustion criteria of a heart rate equal to or greater than 80% of age-predicted heart rate max (220-age) and a rate of perceived exertion at or above 17 out of 20 (Borg, 1970) were both met by all subjects. While running, one leg of the treadmill was placed on the force plate and vertical ground reaction force data was again used to verify subjects maintained their habitual foot strike. Following the run, the treadmill was once again set to 3.4 $m \cdot s^{-1}$ and a second 10 s motion capture was performed.

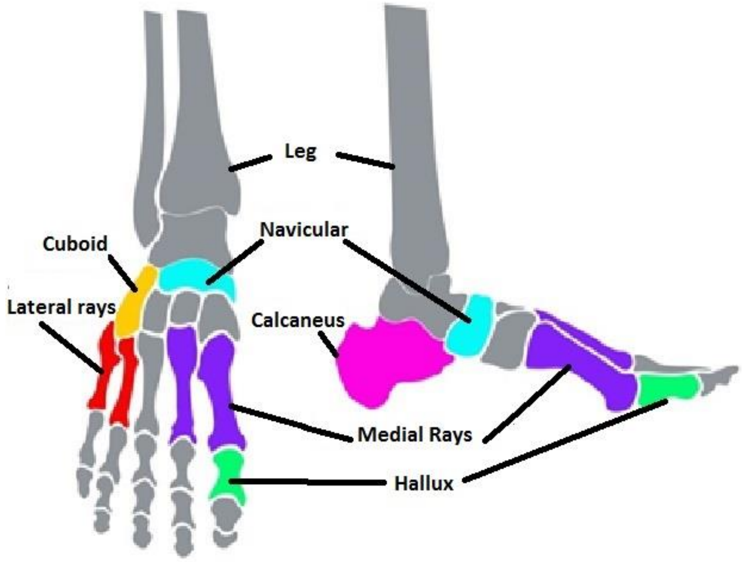

Figure 3.1 -The seven segments that were identified to form the following functional articulations: rearfoot (leg and calcaneus), calcaneonavicular (calcaneus and navicular), calcaneocuboid (calcaneus and cuboid), lateral forefoot (cuboid and lateral rays), medial forefoot (navicular and medial rays), and 1st MTP (medial rays and hallux).

Data Processing

Cortex software (Cortex v 1.1, Motion Analysis Inc., Santa Rosa, CA) was used to track 3D marker position data and a custom Matlab program (Matlab v. R2013b, Mathworks, Natick, MA) was created to filter the data with a zero-lag, low-pass fourth order Butterworth filter and 12 Hz cut-off frequency. Reconstruction of the 3D position of each rigid body segment was performed using the Calibrated Anatomical System Technique (Cappozzo, Catani, Croce, & Leardini, 1995). Finally, joint angles for each of the functional articulations were calculated using the joint coordinate systems technique (Grood & Suntay, 1983). All motions were defined as the distal segment moving on the proximal segment with positive joint angles reflecting dorsiflexion (extension), inversion, and internal rotation (adduction).

Five consecutive stance phases were processed from each 10 s motion capture. All kinematic data were time normalized to 100 percent stance. Initial contact was identified when the toe horizontal velocity was 0 m/s (Zeni, Richards, & Higginson, 2008) and toe-off was identified when the first metatarsophalangeal joint (MTP) reached maximum extension

(Appendix L) (Seneli, Pomeroy, & Cobb, 2015). The stance phase was further divided into four subphases: subphase 1 (0-20%), subphase 2 (21-50%), subphase 3 (51-75%), and subphase 4 (76-100%) for subsequent data analysis (Ferber et al., 2005).

Statistical Analyses

Independent *t*-tests were used to compare average weekly mileage between the RFS and FFS groups and the time to completion of the 3.1 mile exhaustive run.

Initial contact and peak joint angles

Mixed between-within subjects MANOVAs were performed for each functional articulation to investigate initial contact and peak stance phase joint angle differences between the foot strike groups prior to and following the exhaustive run. The between-subject factor in each MANOVA was foot strike (RFS, FFS), and the within-subject factor was time (pre-, post-run). The dependent variables in each MANOVA were initial contact angle, maximum stance phase angle, and minimum stance phase angle (Table 3.1). For the rearfoot complex, calcaneonavicular, and calcaneocuboid articulations separate mixed between-within subjects MANOVAs were performed for the sagittal, frontal, and transverse plane data. For the medial forefoot, MTP, and lateral forefoot only the sagittal plane data was investigated. Follow-up RM ANOVA testing was done to investigate statistically significant mixed between-within subjects MANOVA results. Follow-up dependent *t*-tests with a Bonferroni adjustment were performed to investigate significant mixed between-within subjects ANOVA time-by-foot strike interactions.

Table 3.1. Description of the dependent variables utilized for the mixed between-within subjects MANOVAs. Each articulation had three MANOVAs (one for each plane of motion) to process significant findings.

Articulations	Planes Analyzed	Dependent Variables	Statistical Test
Rearfoot complex Calcaneonavicular Calcaneocuboid Medial forefoot Lateral forefoot 1st MTP	Sagittal Frontal Transverse	Initial Contact Minimum Angle Maximum Angle	MANOVA for each plane of each articulation (total 18 MANOVA)

Angular displacement

Mixed between-within subjects MANOVAs with one between-subjects (foot strike) and one within-subjects factor (time) were performed to investigate differences in rearfoot complex, calcaneocuboid, and calcaneocuboid angular displacement within each of the four stance subphases. The dependent variables in each MANOVA were the sagittal, frontal, and transverse plane angular displacements within a given subphase (Table 3.2). For the medial forefoot, MTP, and lateral forefoot, RM ANOVAs were performed to identify within- and between-subjects sagittal plane differences. Follow-up testing for significant mixed between-within subjects MANOVAs were the same as indicated above.

Table 3.2. Description of the dependent variables utilized for the mixed between-within subjects MANOVAs and ANOVAs when comparing angular displacement. Each articulation had four MANOVAs or ANOVAs (depending on the joint) processed for each of the four subphases of stance.

Articulation	Subphases Analyzed	Dependent Variables	Statistical Test
Rearfoot complex Calcaneonavicular Calcaneocoboid	Subphase 1 (0-20%) Subphase 2 (21-50%) Subphase 3 (51-75%) Subphase 4 (76-100%)	Mean sagittal displacement Mean frontal displacement Mean transverse displacement	MANOVA for each subphase of stance for each articulation (total 12 MANOVA)
Medial forefoot Lateral forefoot 1st MTP	Subphase 1 (0-20%) Subphase 2 (21-50%) Subphase 3 (51-75%) Subphase 4 (76-100%)	Mean sagittal displacement	ANOVA for each subphase of stance for each articulation (total 4 ANOVA)

Significance was set at α = 0.05. For significant interaction effects that required multiple tests for follow-up analyses, a Bonferroni adjusted alpha level of 0.003 was used.

Results and Discussion

This study investigated the differences in foot kinematics between habitual RFS and FFS runners prior to and following an exhaustive 5 km run. Because of the number of articulations examined and the different measurements observed, only significantly different results will be discussed, but all means are reported in Tables 3.4 and 3.5.

Table 3.3- Participant descriptive data

Variable	RFS (n=15)	FFS (n=15)	*P*-value
Age (years)	27.7 ± 5.05	27.4 ± 6.34	
Height (cm)	179 ± 7.69	179 ± 7.02	
Mass (kg)	77.9 ± 10.3	61.2 ± 33.6	
Weekly distance (mi)	20.2 ± 11.5	22.4 ± 15.8	.64
Experience running over 10 mi (years)	4.82 ± 4.47	6.40 ± 5.59	
5K completion time (min)	23.2 ± 4.03	23.2 ± 4.03	.67

Foot strike differences

With respect to the differences between the foot strike patterns, we hypothesized significant kinematic differences at initial contact and during early stance. This hypothesis was supported at initial contact with greater rearfoot complex plantarflexion (6.59 ± 1.25° difference, $P < 0.001$) and inversion angle (2.74 ± 0.96° difference, $P = 0.01$) in FFS runners (Table 3.2). Greater plantarflexion in the calcaneocuboid articulation at initial contact (5.62 ± 1.70°, $P = 0.003$) was also observed in FFS runners (Table 3.4). The FFS group also demonstrated increased peak rearfoot inversion (2.92 ± 0.92°, $P < 0.001$) (Table 3.4), which occurred near initial contact. Forefoot running requires increased rearfoot plantarflexion to make initial contact with the forefoot. In conjunction with the greater rearfoot complex plantarflexion, the increased plantarflexion at the calcaneocuboid articulation and increased rearfoot complex inversion may function to put FFS runners in a more supinated position during early stance. The increased rearfoot plantarflexion and inversion positions in FFS runners may result in decreased ankle joint stability and put the ankle in a more susceptible position to acute injury such as ankle sprain. Although the ankle joint is in a less stable position in plantarflexion and inversion, the position contributes to supination and increased rigidity of the foot which would be important for impact in FFS runners.

In addition to the initial contact and peak angle differences, there were also significant angular displacement differences between the foot strikes in the midfoot and forefoot articulations during early stance (subphase 1). In the sagittal plane, the calcaneonavicular and calcaneocuboid articulations plantarflexed in FFS runners (1.00 ± 0.27° and 0.57 ± 0.38° plantarflexion, respectively), but dorsiflexed in the RFS runners (0.73 ± 0.28° and 0.82 ± 0.38° dorsiflexion, $P < 0.001$ and $P = 0.02$, respectively) (Table 3.5). In the frontal plane, the calcaneonavicular complex everted in FFS runners (0.93 ± 0.33°) but inverted in RFS runners (0.06 ± 0.34°, $P = 0.05$) (Table 3.3). Contrary to the initial hypothesis, RFS runners went through significantly greater medial forefoot dorsiflexion ROM during the first subphase compared to the FFS runners (0.92 ± 0.39° difference, $P = 0.03$) (Table 3.5).

The midfoot and forefoot plantarflexion observed in FFS runners may be a strategy to increase/maintain foot rigidity while the heel is lowered to the ground. The opposite may be true for RFS runners with midfoot and forefoot dorsiflexion functioning to increase foot mobility as the forefoot is lowered to the ground. Because plantarflexion and inversion are triplanar motions of supination, it is interesting to note that while FFS runners plantarflexed the calcaneonavicular articulation during early stance, the complex also everted. If FFS runners made initial contact with the center of pressure more lateral, as previously reported (Cavanagh & Lafortune, 1980), the ground reaction force may have caused this eversion motion. A similar observation was also made in RFS runners with calcaneonavicular dorsiflexion and inversion. It should also be noted that, although a significant interaction was not observed, inspection of the pre- and post-run frontal plane calcaneonavicular motion indicates that the direction of motion actually changed from an inversion motion pre-run to eversion post-run in RFS runners. The different foot kinematics between RFS and FFS running during early stance suggest loading of different foot and ankle structures. The differences may impact the location and type of injury to which each foot strike pattern is more susceptible.

In addition to the early stance subphase differences, several unanticipated range of motion differences between the foot strike patterns were also observed in the second and third subphases. In subphase 2, FFS runners had increased calcaneocuboid dorsiflexion compared to RFS runners (7.20 ± 1.30° difference, $P < 0.001$) (Table 3.5). The FFS runners' conversion from calcaneocuboid plantarflexion in subphase 1 to dorsiflexion in subphase 2 may be the result of the foot becoming more mobile to adapt to running terrain. The increased dorsiflexion that FFS runners had over RFS runners may aide them in reaching full pronation at the same time during stance since they had to come from a plantarflexed position. Additionally, in the transverse plane FFS runners had decreased calcaneocuboid abduction range of motion compared to RFS runners (2.55 ± 1.16° difference, $P = 0.031$) (Table 3.5). It is during subphase 2 when RFS runners reach their peak calcaneocuboid abduction while FFS runners reached peak near initial contact. The timing difference may be the result of a more distal center of pressure in FFS runners during the first half of stance. Also, although no interaction was observed, inspection of the pre- and post-run transverse plane calcaneocuboid motion indicates that FFS runners switched from an adduction motion pre-run to abduction post-run. In subphase 3, MTP extension was increased in FFS runners over RFS runners (2.60 ± 1.02°, *P* = 0.02) (Table 3.5). This may also be the result of a more anterior center of pressure in FFS running (Becker, Pisciotta, James, Osternig, & Chou, 2014) which allows for a quicker conversion to prepare the foot for take-off, where maximum MTP extension occurs. Inspection of the time series supports this theory as well as MTP extension appears to begin sooner in FFS runners (Figure 3.2).

Previous research has identified similar differences in rearfoot complex sagittal and frontal plane kinematics with RFS runners exhibiting a dorsiflexed position and FFS runners demonstrating a plantarflexed and more inverted position at initial contact (Bonacci et al., 2013; Cavanagh & Lafortune, 1980; De Wit et al., 2000; Lieberman et al., 2010; Reinschmidt et al., 1997a; Shih et al., 2013; Squadrone & Gallozzi, 2009; Stackhouse et al., 2004; Williams et al., 2000). Shih, et al. (2013) observed nearly a 20° sagittal plane difference while Squadrone and Gallozzi (2009) observed a 7° difference between FFS and RFS running when switching from running barefoot to shod, both of which are larger than what was observed presently. These studies, however, placed anatomical markers on the shoe in shod conditions and evaluated the rearfoot using a single foot segment model. Additionally, the same runners performed both foot strike patterns rather than evaluate habitual FFS and RFS runners which may account for some of the difference. In the frontal plane, previous research has also identified an increased inversion angle (~5°) in FFS runners compared to RFS runners while tracking shoe motion

(Stackhouse et al., 2004; Williams et al., 2000) which is comparable to the 2.75° increase observed in the present study. Because no other studies have compared functional articulations of the foot with different running strike patterns, the present findings are novel and help to better understand the patterns of rigidity and mobility within the foot for RFS and FFS runners. This exploration of how foot pronation and supination occur in healthy runners may be used in future studies to investigate differing kinematics in injured populations. Other studies using a single-segmented foot model have been inconclusive as to whether or not increased foot pronation contributes to injuries such as plantar fasciitis and stress fractures (Tam et al., 2014). Though other studies debated whether or not there are significant differences in injury rate between RFS and FFS running (Daoud et al., 2012; Goss & Gross, 2012), types of injuries and injury locations appear to be different. Daoud et al. (2012) identified much higher rates of RFS predicted injuries (hip pain, knee pain, lower back pain, tibial stress injuries, plantar fasciitis, and lower limb stress fractures) in RFS runners while FFS runners were predicted to have higher rates of metatarsal fractures, Achilles tendinopathies, and foot pain, but were not found to be different between the foot strikes. Similar results were also reported in barefoot runners, majority of which run with a FFS or midfoot strike pattern, when compared to shod runners, majority of which use a RFS running pattern (Altman & Davis, 2015). The increased time RFS runners are in a more pronated position may contribute to these types of injuries. It has been suggested by Pohl, Hamill, and Davis (2009) that understanding of midfoot pronation may help to better understand the relationship between foot pronation and running related injuries.

Exhaustive run

With respect to the exhaustive run, we hypothesized that foot and ankle angular displacements would increase throughout the stance phase in both groups following the run. At the end of the exhaustive run, both foot strike groups made initial contact with the rearfoot complex in greater internal rotation (1.55 ± 0.71° difference, $P = 0.04$) and both the calcaneonavicular (1.28 ± 0.39° difference, P = 0.003) and calcaneocuboid (2.00 ± 0.54°, P = 0.001) functional articulations in greater plantarflexion (Table 3.2). Although no interaction was found, it should be noted that further investigation of the rearfoot complex shows that the increased initial contact internal rotation angle only increased in RFS runners (2.54 ± 2.24°), but actually decreased in FFS runners (0.39 ± 1.99°). This interaction likely was masked because of the MANOVA model used. The increased rearfoot internal rotation observed in RFS runners may be the result of exhaustion to the anterior tibialis. The increased midfoot plantarflexion position at initial contact may function to contribute to the rearfoot complex plantarflexion in FFS runners (Table 3.4), which would allow FFS runners to assure forefoot contact. Additionally, the midfoot plantarflexed angle may assist in providing more bony rigidity for the foot as dynamic stabilizers become exhausted. In RFS runners, the midfoot plantarflexion may also have been a result of fatigue in the tibialis anterior muscle which is more active in RFS running (Shih et al., 2013) and therefore more susceptible to fatigue.

In addition to the changes in initial contact position, both foot strike groups also demonstrated a number of pre- to post-run changes in peak position and range of motion during the four stance subphases. During subphase 1, peak rearfoot complex internal rotation, which occurred near initial contact, increased post-run for RFS and FFS runners (1.68 ± 0.73°

increase, P = .03) (Table 3.5). Rearfoot eversion (1.02 ± 0.29° increase, P = 0.002) and external rotation (0.92 ± 0.32°, P = 0.01) range of motion during subphase 1 also increased in both groups (Table 3.5). The increased rearfoot complex external rotation displacement is perhaps necessary to allow appropriate foot position/posture needed for dissipation of force after initial contact is made with a greater internally rotated rearfoot. Increased calcaneonavicular eversion (0.56 ± 0.24° increase, P = 0.02) and adduction (0.78 ± 0.21° increase, P = 0.001) were also observed in subphase 1. The increased eversion in both joints will contribute to foot pronation and increase foot mobility going into subphase 2. The increased pronation and foot mobility may be the result of fatigue of dynamic foot stabilizers. The increase in pronation and foot mobility would rely strongly on dynamic stability and, as a result, these dynamic stabilizers which may already be exhausted from the run may become overstressed. These soft tissue structures, which would include structures such as the plantar fascia and spring ligament, may be at a higher risk for injury with the increased pronation.

During subphase 2, peak rearfoot complex eversion increased as a result of the exhaustive run (1.53 ± 0.34°, $P < 0.001$) (Table 3.5). This change contributes to increased maximal pronation while running. It has strong clinical relevance as over pronation has been linked to multiple overuse injuries (Ferber, Hreljac, & Kendall, 2009). As discussed previously, as dynamic stabilizers become fatigued, control of pronation would become more difficult.

During the third stance subphase, both groups increased rearfoot plantarflexion and inversion range of motion as a result of the exhaustive run (0.98 ± 0.38° and 0.41 ± 0.18° increase, P = 0.02 and P = 0.03, respectively) (Table 3.5). Peak calcaneonavicular and calcaneocuboid dorsiflexion, which occurred in subphase 3, decreased in both the calcaneonavicular (1.37 ± 0.45° decrease, $P = 0.006$) and in the calcaneocuboid articulations post-run (1.91 ± 0.73° decrease, $P = 0.02$) (Table 3.5). Additional subphase 3 range of motion changes post-run included increased lateral forefoot plantarflexion (0.95 ± .34, $P = 0.009$) and MTP extension post run (1.78 ± 0.35°, $P < 0.001$) (Table 3.5). The increased range of motion for all of these joints pushes the foot towards a more supinated position. As the foot prepares for the swing phase of stance, a rigid foot is crucial for a sufficient push off force. Therefore, as pronation increased during the second subphase of stance, the increased supination motion during subphase 3 may be in response to trying to make up for the necessary foot rigidity during late stance.

Finally, during subphase 4, both RFS and FFS runners had increased range of motion in rearfoot complex plantarflexion (1.11 ± 0.32°, $P = 0.002$), inversion (0.90 ± 0.30°, $P = 0.01$), and internal rotation (0.87 ± 0.29°, $P = 0.01$) post-run (Table 3.5). This directly results in greater ankle supination during push-off. Increases were also observed in medial forefoot plantarflexion range of motion (1.87 ± 0.40°, $P < 0.001$), MTP extension range of motion (1.53 ± 0.43°, $P = 0.001$) and peak MTP extension (2.60 ± 0.75°, $P = 0.002$) (Table 3.3). Like in subphase 3, the increased supination may aid in creating a more rigid foot for push-off. The increased plantarflexion of the rearfoot and midfoot would also contribute to the need for the MTP to increase extension at toe-off. Because of the less rigid foot during subphases 1 and 2, changes in subphase 4 must allow compensation for a productive push-off to maintain the same speed.

In addition to the hypothesized increased range of motion following the exhaustive run, we further postulated that the increase would be greater in FFS vs RFS runners due to the increased use of the plantarflexors associated with making initial contact with the forefoot.

This was not the case as only one interaction was identified. In subphase 2, medial forefoot dorsiflexion range of motion was increased in RFS runners (2.47 ± 1.23° increase, $P < 0.001$) but not FFS runners ($P = 0.15$) (Table 3.5). It is worth mentioning that in this subphase, medial forefoot dorsiflexion in FFS runners was nearly double that of RFS runners both at the beginning and end of the run (6.00 ± 1.15° difference overall). The medial forefoot dorsiflexion is part of the motion that lowers the medial arch to aid in force absorption as peak ground reaction forces are reached. As the foot is increasing pronation during this subphase, it would appear that fatigue of dynamic stabilizers in RFS runners contributes to increased foot mobility after an exhaustive run. As FFS runners have much more medial forefoot dorsiflexion than RFS runners both at the beginning and end of the run, it would appear FFS runners have very little rigidity in this joint to begin with and may already be near their end range of medial forefoot sagittal plane motion.

Similar to the present study, previous research utilizing exhaustive running protocols have also reported decreased rearfoot dorsiflexion in RFS runners at initial contact as well as increased peak eversion during stance (Christina et al., 2001; Dierks et al., 2010; VanGheluwe & Madsen, 1997), but not all studies are consistent. Pohl, et al. (2010) found no change in peak rearfoot eversion with a local posterior tibialis fatiguing protocol, but the difference may be the result of a local muscular fatiguing protocol rather than exhaustion from running. When running to volitional exhaustion, Derrick, et al. (2002) also observed increased rearfoot inversion at initial contact which was not found to be statistically significant in the present study. However, it appeared that the data was following this trend, but again, may have been masked by the MANOVA model.

Limitations

With the utilization of a multi-segment foot model and the number of variables examined, the mixed between-within subjects MANOVA statistical model used for most variables allowed protection for type I error. Because of this conservative model, it is likely that some results may have actually been statistically significant but were not identified as such, particularly with interaction effects. An example of this is in the rearfoot complex sagittal and frontal plane angle at initial contact which had a significant time main effect from the MANOVA, but was not found to be significantly different from follow-up testing in the present study. The data did appear to follow results that have been previously reported where the rearfoot decreased dorsiflexion and increased inversion after fatigue (Derrick et al., 2002), but may have been masked by the MANOVA model. Similarly, there were various significant reactions which were investigated with follow up testing, but because of the Bonferroni adjusted *p*-value (0.003), only one was found to be significantly different (Appendix K). .

Conclusion

This exploratory study helped to identify differences in distal foot kinematics between habitual RFS and FFS runners. Though it has been suggested that RFS runners may experience more bony tissue injury and FFS runners more soft tissue (Daoud et al., 2012), the more supinated foot of the FFS runners would create more bony stability and contradict this thought when considering soft tissue structures within the foot itself. The more pronated foot in RFS runners would suggest increased stress on dynamic stabilizers and is supported by the finding that RFS runners have experienced a higher rate of plantar fasciitis (Daoud et al., 2012). The exhaustive run affected both RFS and FFS runners by increasing peak joint angles and range of motion. The increased range of motion during the first and second subphases of stance would result in increased pronation and a more mobile foot, putting more stress and possibly increasing injury risk on dynamic stabilizers and soft tissue structures such as the plantar fascia and medial arch ligaments. Increased range of motion from the run during the third and fourth subphases of stance resulted in greater supination and foot rigidity which is necessary for push-off effectiveness. Additionally, differences were observed in the medial forefoot at the end of run where dorsiflexion range of motion increased in RFS runners in subphase 2 while FFS runners did not change, suggesting a greater fatiguing effect to medial forefoot kinematics in RFS runners. Some of the significant differences observed were relatively small (less than a degree in some cases), which may question the clinical relevance. However, in considering angular displacement, the more clinically relevant piece may be the differing direction of motion, and therefore loading of different structures such as in the first stance subphase where FFS runners had greater foot rigidity and RFS runners' greater mobility. As this study was largely exploratory and helping to identify kinematic patterns between the foot strikes, future research should begin to explore the articulations where differences were identified and compare them with injured populations.

Table 3.4- Mean ± SD of joint angles in RFS and FFS runners at the beginning (Pre-run) and end (Post-run) of an exhaustive 5 km run for the rearfoot complex (RC), calcaneonavicular (CNC), medial forefoot (MFF), first metatarsophalangeal (MTP), lateral forefoot (LFF), and calcaneocuboid (CC). Positive numbers are associated with dorsiflexion/extension, inversion, and internal rotation/adduction.

		RFS						**FFS**					
		Initial Contact		**Max angle**		**Min angle**		**Initial Contact**		**Max angle**		**Min angle**	
Joint	**Plane**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**
RC	Sag	**4.28 ± 3.31[b]**	**3.95 ± 3.63[b]**	13.3 ± 3.62	13.6 ± 3.78	-4.24 ± 4.11	-5.71 ± 3.68	**-3.40 ± 3.63[b]**	**-1.55 ± 3.69[b]**	11.3 ± 3.34	12.4 ± 3.86	-6.33 ± 4.46	-6.56 ± 3.63
	Fron	**3.46 ± 3.41[b]**	**4.28 ± 2.62[b]**	**3.87 ± 3.35[b]**	**4.82 ± 2.45[b]**	**-4.45 ± 3.91[a]**	**-5.97 ± 4.06[a]**	**6.30 ± 2.58[b]**	**6.93 ± 2.23[b]**	**6.95 ± 2.53[b]**	**7.58 ± 1.89[b]**	**-3.67 ± 3.15[a]**	**-5.21 ± 3.26[a]**
	Trans	**-0.98 ± 5.02[a]**	**2.52 ± 5.06[a]**	**0.46 ± 4.79[a]**	**3.67 ± 4.83[a]**	-13.7 ± 5.18	-13.3 ± 5.41	**2.80 ± 3.03[a]**	**2.41 ± 4.89[a]**	**4.02 ± 2.63[a]**	**4.16 ± 4.42[a]**	-12.4 ± 3.24	-13.0 ± 4.15
CNC	Sag	**-3.94 ± 2.94[a]**	**-5.38 ± 3.68[a]**	**1.28 ± 3.12[a]**	**-0.04 ± 4.32[a]**	-8.03 ± 3.73	-8.63 ± 4.15	**-5.83 ± 4.04[a]**	**-6.95 ± 3.86[a]**	**0.61 ± 2.54[a]**	**-0.82 ± 3.00[a]**	-9.80 ± 3.51	-10.0 ± 3.56
	Fron	1.81 ± 3.31	2.13 ± 2.40	3.71 ± 3.09	3.57 ± 2.72	-4.19 ± 2.51	-4.72 ± 3.13	2.02 ± 4.19	2.34 ± 4.06	2.83 ± 3.90	2.70 ± 4.00	-4.04 ± 3.34	-5.32 ± 3.08
	Trans	5.99 ± 4.49	3.34 ± 4.67	9.29 ± 4.79	7.53 ± 4.62	3.03 ± 2.51	1.22 ± 4.12	0.18 ± 4.67	0.54 ± 5.23	5.74 ± 3.84	5.93 ± 5.23	-1.04 ± 3.69	-0.91 ± 4.96
MFF	Sag	**5.11 ± 3.43[c]**	**3.66 ± 3.67[c]**	14.8 ± 3.64	16.8 ± 4.73	-3.11 ± 3.04	-3.37 ± 3.47	-4.34 ± 3.41	-2.89 ± 3.29	12.0 ± 2.74	14.3 ± 3.54	-6.59 ± 3.51	-6.62 ± 4.08
MTP	Sag	11.4 ± 6.25	13.8 ± 4.33	**36.0 ± 6.85[a]**	**39.0 ± 7.16[a]**	-2.45 ± 2.38	-3.03 ± 4.09	14.56 ± 6.57	15.5 ± 4.45	**39.2 ± 6.74[a]**	**41.4 ± 6.74[a]**	-1.13 ± 3.93	-2.71 ± 4.48
LFF	Sag	2.27 ± 2.30	2.32 ± 3.99	7.64 ± 3.13	9.01 ± 3.18	-0.39 ± 2.37	-0.23 ± 3.30	-0.90 ± 2.11	0.67 ± 3.64	5.47 ± 1.81	6.29 ± 3.86	-2.73 ± 1.94	-1.96 ± 3.68
CC	Sag	**-4.45 ± 3.01[a,b]**	**-6.42 ± 4.28[a,b]**	**4.06 ± 3.97[a]**	**2.08 ± 4.16[a]**	-10.5 ± 3.19	-11.2 ± 4.91	**-10.0 ± 5.09[a,b]**	**-12.1 ± 6.53[a,b]**	**5.03 ± 2.97[a]**	**3.19 ± 6.29[a]**	-12.4 ± 4.31	-13.9 ± 6.30
	Fron	-1.81 ± 5.77	-0.11 ± 4.72	2.59 ± 5.67	2.76 ± 4.91	-5.71 ± 4.34	-5.14 ± 4.93	0.64 ± 4.07	1.12 ± 4.72	3.54 ± 4.03	3.88 ± 4.77	-4.59 ± 3.12	-4.68 ± 6.26
	Trans	2.88 ± 3.76	2.61 ± 5.19	2.76 ± 4.91	6.86 ± 4.09	0.46 ± 3.78	-0.41 ± 3.84	1.56 ± 4.86	3.25 ± 6.52	3.88 ± 4.77	6.38 ± 3.54	0.26 ± 4.75	1.62 ± 6.22

[a] Indicates time main effects difference from follow-up RM ANOVAs for the same foot strike ($p \leq .05$)
[b] indicates foot strike main effect difference from follow-up RM ANOVAs ($p \leq .05$)
[c] indicates difference from significant time-by-foot strike interaction follow-up *t*-test ($p \leq .003$)

Table 3.5- Angular displacements for subphase 1-4 of running stance for RFS (top) and FFS (bottom, shaded) runners at the following articulations: rearfoot complex (RC), calcaneonavicular (CNC), medial forefoot (MFF), first metatarsophalangeal (MTP), lateral forefoot (LFF), and calcaneocuboid (CC).

		Subphase 1		Subphase 2		Subphase 3		Subphase 4	
Joint	**Plane**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**	**Pre-run (°)**	**Post-run (°)**
RC	Sagittal	-1.90 ± 1.76	-2.00 ± 2.06	10.1 ± 2.06	11.5 ± 1.85	**-0.68 ± 4.21**[a]	**-1.71 ± 4.28**[a]	**-13.5 ± 2.77**[a]	**-14.8 ± 3.4**[a]
		-0.31 ± 1.27	-0.37 ± 2.14	12.8 ± 3.60	12.2 ± 3.72	**-2.60 ± 2.73**[a]	**-3.53 ± 2.33**[a]	**-13.5 ± 2.84**[a]	**-14.5 ± 3.12**[a]
	Frontal	**-1.86 ± 1.37**[a]	**-2.79 ± 2.19**[a]	-5.64 ± 2.19	-6.97 ± 2.80	**3.68 ± 1.44**[a]	**4.07 ± 1.46**[a]	**2.86 ± 1.75**[a]	**3.96 ± 2.16**[a]
		-0.73 ± 1.46[a]	**-1.84 ± 1.91**[a]	-8.56 ± 3.65	-9.86 ± 3.18	**3.28 ± 1.23**[a]	**3.71 ± 1.71**[a]	**2.67 ± 1.72**[a]	**3.39 ± 2.36**[a]
	Transverse	**-2.22 ± 1.32**[a]	**-2.78 ± 2.38**[a]	-9.56 ± 3.15	-11.9 ± 3.64	4.19 ± 1.55	4.08 ± 2.18	**8.73 ± 2.05**[a]	**9.75 ± 1.86**[a]
		-1.09 ± 1.96[a]	**-2.35 ± 3.15**[a]	-13.4 ± 3.94	-12.3 ± 3.51	3.59 ± 2.68	3.46 ± 3.17	**9.27 ± 2.51**[a]	**9.98 ± 2.55**[a]
CNC	Sagittal	**0.67 ± 1.15**[b]	**0.80 ± 1.11**[b]	3.29 ± 3.83	3.07 ± 3.94	-2.66 ± 1.82	-2.55 ± 2.58	**-5.95 ± 3.70**[a]	**-4.59 ± 3.41**[a]
		-1.20 ± 0.82[b]	**-0.79 ± 1.39**[b]	6.68 ± 3.29	6.10 ± 3.02	-2.89 ± 2.83	-2.89 ± 2.56	**-6.41 ± 2.52**[a]	**-5.44 ± 2.75**[a]
	Frontal	**0.39 ± 1.66**[a,b]	**-0.27 ± 1.64**[a,b]	-3.21 ± 4.63	-3.95 ± 4.68	-1.46 ± 2.75	-1.70 ± 1.87	1.11 ± 3.13	0.92 ± 3.94
		-0.69 ± 0.99[a,b]	**-1.16 ± 1.31**[a,b]	-3.40 ± 4.41	-4.40 ± 3.51	-1.26 ± 2.45	-0.88 ± 2.54	1.34 ± 3.04	1.49 ± 2.75
	Transverse	**2.01 ± 1.43**[a]	**2.52 ± 1.44**[a]	-3.67 ± 3.26	-2.13 ± 4.13	-0.16 ± 2.98	-0.29 ± 2.66	1.62 ± 2.83	1.23 ± 2.71
		1.55 ± 1.30[a]	**2.60 ± 2.09**[a]	1.33 ± 4.00	-0.37 ± 4.12	0.02 ± 2.46	0.04 ± 2.47	-0.56 ± 3.48	0.11 ± 3.38
MFF	Sagittal	**1.59 ± 1.00**[b]	**2.13 ± 1.28**[b]	**6.23 ± 2.57**[c]	**8.70 ± 3.11**[c]	-2.76 ± 3.22	-2.91 ± 3.62	**-13.6 ± 2.84**[a]	**-15.5 ± 2.73**[a]
		0.78 ± 1.21[b]	**1.10 ± 1.26**[b]	**13.1 ± 3.65**[c]	**13.9 ± 3.33**[c]	-4.23 ± 1.57	-4.95 ± 1.68	**-13.5 ± 3.31**[a]	**-15.3 ± 2.75**[a]
MTP	Sagittal	-2.01 ± 1.20	-2.37 ± 2.36	-11.6 ± 5.42	-14.2 ± 6.12	**14.4 ± 2.6**[a,b]	**16.3 ± 3.48**[a, b]	**22.3 ± 4.60**[a]	**23.8 ± 5.53**[a]
		-1.17 ± 2.53	-2.46 ± 2.69	-14.2 ± 7.56	-15.3 ± 4.80	**17.1 ± 2.75**[a,b]	**18.8 ± 2.75**[a,b]	**20.9 ± 4.37**[a]	**22.4 ± 4.43**[a]
LFF	Sagittal	0.55 ± 1.59	0.96 ± 1.48	3.58 ± 2.15	4.19 ± 2.56	**-1.02 ± 2.70**[a]	**-2.14 ± 2.68**[a]	-5.32 ± 1.59	-5.92 ± 2.10
		1.82 ± 1.17	1.62 ± 1.35	3.38 ± 1.15	2.18 ± 2.35	**-1.11 ± 2.46**[a]	**-1.88 ± 2.28**[a]	-5.21 ± 1.67	-5.21 ± 2.20
CC	Sagittal	**0.60 ± 1.95**[b]	**1.04 ± 1.61**[b]	**7.07 ± 3.21**[b]	**6.30 ± 2.39**[b]	-3.78 ± 2.80	-3.10 ± 2.81	-9.90 ± 2.93	-8.96 ± 3.49
		-1.12 ± 1.43[b]	**-0.02 ± 2.27**[b]	**14.0 ± 3.95**[b]	**13.8 ± 4.72**[b]	-4.46 ± 2.89	-4.92 ± 2.73	-9.91 ± 3.19	-10.5 ± 3.68
	Frontal	0.78 ± 1.89	0.40 ± 2.18	-2.08 ± 6.26	-3.74 ± 4.02	1.97 ± 4.27	1.98 ± 3.67	-1.53 ± 3.97	-1.98 ± 3.41
		0.17 ± 1.67	-1.00 ± 2.32	-4.90 ± 3.56	-4.01 ± 4.50	3.64 ± 2.73	4.61 ± 2.95	1.02 ± 3.47	0.07 ± 3.02
	Transverse	1.06 ± 1.05	0.91 ± 1.63	**-2.91 ± 3.18**[b]	**-2.27 ± 3.72**[b]	2.84 ± 2.57	3.05 ± 2.09	0.68 ± 2.42	-0.32 ± 3.20
		0.92 ± 1.38	1.20 ± 1.78	**0.56 ± 3.57**[b]	**-0.64 ± 2.98**[b]	2.19 ± 2.51	2.58 ± 2.29	-1.40 ± 3.36	-1.08 ± 2.91

[a] Indicates time main effects difference from follow-up RM ANOVAs ($p \leq .05$)
[b] indicates foot strike main effect difference from follow-up RM ANOVAs ($p \leq .05$)
[c] indicates difference from significant time-by-foot strike interaction RM ANOVAs ($p \leq .003$)

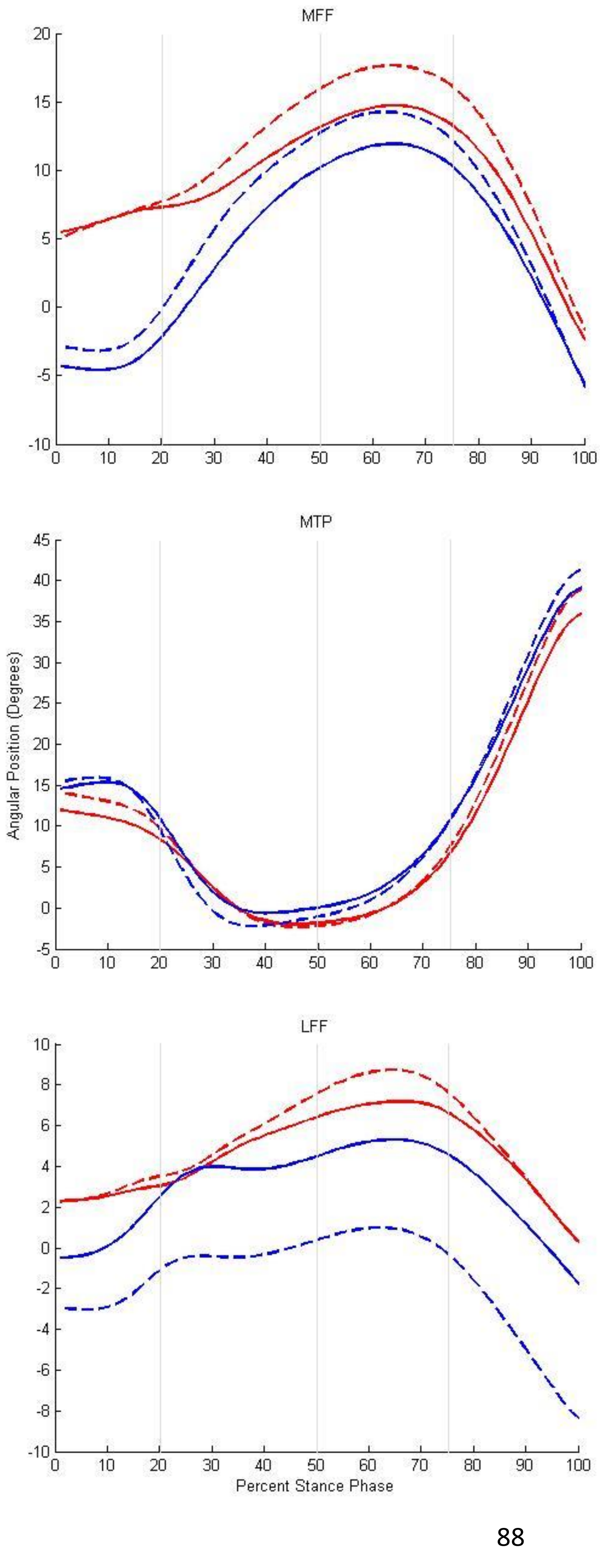

Figure 3.2- Medial forefoot (MFF), first metatarsophalangeal joint (MTP), and lateral forefoot (LFF) sagittal plane kinematics for RFS (red) and FFS (blue) runners during the stance phase of running at the beginning (solid) and end (dashed) of a 5 km treadmill run. Vertical grey lines indicate the division between the four subphases of stance.

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4. FOOT JOINT COUPLING VARIABILITY DIFFERENCES BETWEEN HABITUAL REARFOOT AND FOREFOOT RUNNERS PRIOR TO AND FOLLOWING AN EXHAUSTIVE RUN

Introduction

Much debate has erupted around the influence of foot strike pattern on injury risk in runners. Although the majority of runners (75-94%) use a rearfoot strike pattern (RFS) (Hasegawa et al., 2007; Kerr et al., 1983; Larson et al., 2011), arguments have been made that traditional running shoes encourage heel contact and that a forefoot strike pattern (FFS) or midfoot strike pattern is more "natural" and may therefore be better at reducing injury rates (Bramble & Lieberman, 2004; Lieberman, 2012; Lieberman et al., 2010). One theory is that FFS running is associated with a decreased vertical ground reaction force during initial impact that results in decreased tissue stress along the kinetic chain (Lieberman et al., 2010; Shih et al., 2013; Williams et al., 2001).

In addition to differing ground reaction force loading characteristics, the foot strikes exhibit different joint coupling patterns along the lower extremity (Pohl & Buckley, 2008). Further, joint coupling variability has been suggested to be an important factor in running related injuries. Though extreme variability can stress tissues to a point where injury occurs, the dynamical systems theory also suggests that some increased variability in coordination patterns may reduce overuse injuries (Hamill et al., 1999; Heiderscheit et al., 2002; van Emmerik & Wagenaar, 1996).

Dynamical systems theory suggests that variability is unavoidable in biological systems and by producing movement with a certain amount of variability these systems are able to adapt to varying constraints placed on them. Therefore, joint coupling patterns with little or no variability may overload musculoskeletal tissues and lead to overuse injury. This theory has been supported by observations of decreased lower extremity joint coupling variability in individuals with patellofemoral pain, chronic ankle instability, and iliotibial band syndrome (Hamill et al., 1999; Herb et al., 2014; Miller, Meardon, Derrick, & Gillette, 2008).

Although joint coupling between the rearfoot and forefoot segments have been observed to be highly correlated in FFS and RFS runners (Pohl & Buckley, 2008), the variability of these coordination patterns has not been analyzed. Additionally, more recent multi-segment foot models have begun to allow researchers to examine foot kinematics using more functional articulations (Wolf et al., 2008). Particularly when comparing different running strike patterns, motion within the foot is important to understand because of the role it plays in stability and force dissipation.

Furthermore, as running kinematics and coordination patterns often change as a result of fatigue (Dierks et al., 2010; Ferber & Pohl, 2011), joint coupling variability may also be affected. Coordination variability would affect the amount of repetitive stress tissues receive. Increased variability would disperse stress to different tissues with each subsequent step, and therefore, avoid overuse injury. However, too much variability may also stress tissues to the point where injury occurs and, therefore, may be detrimental to runners as well.

Therefore, the purpose of this study was to identify differences in foot and ankle joint coupling variability between habitual RFS and FFS runners prior to and following an exhaustive run. Use of habitual RFS and FFS runners was essential for the study to eliminate variability that may be the result of motor learning. It was hypothesized that FFS runners would have increased coupling variability within the foot early stance because of the dependence of distal foot motion with a forefoot strike. In addition, it was also hypothesized that the exhaustive run protocol would result in decreased joint coupling variability as the system decreases its ability to adapt to continual stress.

Methods

Subjects

Thirty runners (18-40 years), 15 habitual rearfoot runners and 15 habitual forefoot/midfoot runners, were recruited. Participants had to currently be running a minimum of 10 miles per week on average and to have not changed foot strike pattern within the previous year. Volunteers were excluded (Appendix E) if they had a lower extremity injury in the previous 6 weeks, a previous major lower extremity surgery, wore custom molded foot orthotics, had known cardiovascular problems, or uncontrolled asthma. The two groups were matched by age and sex. Experimental procedures were explained and participants signed an informed consent form approved by the Institutional Review Board prior to testing.

Foot Strike Screen

An in-person visual analysis of each participants' foot strike was performed to avoid error in self-reported foot strike patterns (Goss & Gross, 2012). This was done by visual analysis of the participants' vertical ground reaction force data while running on a 25 m walkway over an AMTI force plate (Advanced Medical Technologies, Inc., Waterford, MA) sampling at 1000 Hz. Presence of a double impact peak indicated a RFS, while absence of a double peak placed subjects in the FFS group, which included MFS and FFS runners (Williams et al., 2000). Subjects performed 5 running trials over the force plate both in their habitual running shoes and then in running sandals (Mono, Luna Sandals, Seattle, WA) provided by the lab. At least 3 of the 5 running trials had to exhibit the peak, or lack thereof, to be placed in the appropriate foot strike group.

After foot strike verification, subjects were given the running sandals and instructed to accommodate to them over a minimum 10 day period. They were instructed to run 1 mile in them for their first run and then increase their mileage in the sandal by 0.5 mile every training run while still maintaining their overall running mileage in their habitual running shoes. After 10 days, or when the subject was able to comfortably complete 5 km in the sandals, they returned to the lab for running analysis.

Running analysis

All subjects were equipped with clusters of four retroreflective markers (6.4 mm diameter) on the right leg and foot to identify the tibia, calcaneus, navicular, cuboid, medial rays (metatarsals 1 and 2), lateral rays (metatarsals 4 and 5), and the hallux as well as single markers to identify anatomical landmarks (Appendix A). Markers were adhered to the skin with double-sided toupee tape and liquid adhesive (Mastisol Liquid Adhesive, Ferndale Laboratories, Inc., Ferndale, MI). Wand clusters were additionally adhered using elastic tape (Elastikon, Johnson & Johnson, New Brunswick, NJ) and a cohesive bandage (Powerflex, Andover Healthcare, Inc., Salisbury, MA). Three-dimensional positional marker data was collected at 200 Hz using a 10-camera Eagle Motion Analysis System (Motion Analysis Corp., Santa Rosa, CA). A seated calibration was performed to identify anatomical landmarks from single markers and with a Davis Digitizing Pointer (C-Motion Inc., Germantown, MD) to identify joint coordinate systems for each segment.

After a 5 minute warm-up, subjects were instructed to perform a maximal effort 5 km run on a treadmill located in the center of the capture volume. All subjects began the run at a speed of 3.4 $m \cdot s^{-1}$. Once a consistent gait pattern was observed, motion capture data was collected for 10 s. Subjects were then allowed to adjust the speed to allow for maximal effort during the run. Maximal effort was reached if subjects reached 80% of their age-predicted heart rate max (220-age) and/or a rate of perceived exertion of 17 or higher on a 20 point scale (Borg, 1970), both of which were met by all participants. The treadmill speed was again set at 3.4 $m \cdot s^{-1}$ when 0.16 km remained in the 5 km run to allow for a final 10 s data collection. While running, one leg of the treadmill was positioned on a force plate to verify subjects maintained their habitual foot strike pattern by visual examination of the vertical ground reaction force profile.

Data Processing.

Five consecutive steps from each 10 s motion capture were processed. Marker data was tracked with Cortex software (Cortex v 1.1.4.368, Motion Analysis, Inc., Santa Rosa, CA) and processed with a custom Matlab program (Matlab v. R2013b, Mathworks, Natick, MA). Positional data was filtered with a zero-lag, low-pass (12 Hz) fourth order Butterworth filter. Reconstruction of 3D position of each rigid body segment was performed using the Calibrated Anatomical System Technique (Cappozzo et al., 1995) and functional articulation angles calculated using the joint coordinate systems technique (Grood & Suntay, 1983) (Appendix B). Functional articulations calculated included the rearfoot complex, calcaneonavicular, calcaneocuboid, medial forefoot, lateral forefoot, and first metatarsophalangeal articulation

(MTP). Positive angle measurements identified dorsiflexion (MTP extension), inversion, and adduction (rearfoot complex internal rotation).

All joint angle time series were normalized to 100 percent (101 data points) of stance phase. Initial contact was identified as the instant at which the horizontal velocity of the tip of the hallux was equal to zero (Zeni et al., 2008) and toe off was defined as the point at which the MTP reached maximum extension (Seneli et al., 2015).

Using another custom Matlab code, joint coupling angles (Θ) were computed using vector coding methods:

Equation 4.1. $$\theta_i = \tan^{-1}\left(\frac{y_{i+1} - y_i}{x_{i+1} - x_i}\right)$$

where Θ the joint coupling angle between 0° and 360°, and *i* is a percent of stance phase (0-100%) (Chang et al., 2008; Heiderscheit et al., 2002; MacLean, van Emmerik, & Hamill, 2010; Sparrow et al., 1987). Joint coupling angles were found for each time point. Four subphases of stance were defined as (1) 0-20%, (2) 21-50%, (3) 51-75%, and (4) 76-100% and the mean joint coupling angle was calculated for each subphase. The standard deviation of the vector magnitude was used to calculate coordination variability:

Equation 4.2. $$v^2 = 2[1 - \sqrt{(\bar{x}^2 + \bar{y}^2)}]$$

where v^2 is coordination variability and $\bar{x}$ and $\bar{y}$ are mean coordinates of the vector as determined by the cosine and sine of the joint coupling angle (MacLean et al., 2010).

Statistical Analyses

The joint coupling variability was compared for 10 different joint couples (Table 4.1). The joint couples selected were decided upon based on the constrained tarsal mechanism theory (Huson, 2000) and the concept of pronation twist (Hicks, 1953). The variability of the specified joint couples was compared using mixed between-within subjects ANOVAs to compare between and within the groups for each subphase of stance. The between group main effect was foot strike and the within group main effect was time (pre- and post-run). Significant interactions of main effects were further investigated with dependent *t*-tests for simple time main effects. The significance level was set at 0.05.

Table 4.1 -Joint couplings that were statistically compared for differences between FFS and RFS runners before and after an exhaustive run. The joints included are the rearfoot complex (RC), calcaneonavicular (CNC), calcaneocuboid (CC), medial forefoot (MFF), lateral forefoot (LFF), and 1st metatarsophalengeal (MTP) and the anatomical planes as indicated (transverse = tran, sagittal = sag, frontal = fron).

Medial foot couplings	Lateral foot couplings	Ankle couplings
RCsag-CNCsag	RCsag-CCsag	RCtran-RCfron
RCfron-CNCfron	RCfron-CCfron	
RCtran-CNCtran	RCtran-CCtran	
CNCfron-MFFsag	CCfron-LFFsag	
MFFsag-MTPsag		

Results

To be concise, only statistically significant results are reported here. Means ± SD for all of the variability measurements are located in Table 4.2.

Subphase 1

Significant between-subjects (foot strike) main effects from subphase 1 mixed between-within subjects ANOVAs indicated that FFS runners had greater joint coupling variability between the calcaneocuboid frontal and medial forefoot sagittal plane motion (CCfron_MFFsag) ($p = 0.050$) as well as between the medial forefoot sagittal and MTP sagittal plane motion (MFFsag_MTPsag) ($p < 0.001$). There was less variability between the calcaneocuboid frontal and lateral forefoot sagittal plane motion ($p = 0.048$) in FFS runners both prior to and following the exhaustive run (Table 4.2).

Subphase 2

Investigation of simple time main effects of a significant foot strike by time interaction between the calcaneonavicular frontal and medial forefoot sagittal plane motion (CNCfron_MFFsag) ($p = 0.023$) during subphase 2 indicated that variability decreased in RFS runners overtime ($p = 0.003$) while it did not change in FFS runners ($p = 0.91$) (Table 4.2). Significant foot strike main effects from subphase 2 mixed ANOVAs indicated that FFS runners had less joint coupling variability between the rearfoot sagittal and calcaneonavicular sagittal plane motion (RCsag_CNCsag) ($p < 0.001$), rearfoot sagittal and calcaneocuboid sagittal plane motion (RCsag_CCsag) ($p < 0.001$), and rearfoot frontal and calcaneocuboid frontal plane motion (RCfron_CCfron) ($p = 0.026$) when compared to RFS runners (Table 4.2). Significant time main effects indicated increased subphase 2 variability in both FFS and RFS runners end of the run for the couple between the following five adjacent joint couples: (1) rearfoot transverse and rearfoot frontal plane (RCtran_RCfron) ($p = 0.004$), (2) rearfoot transverse and calcaneonavicular transverse plane motion (RCtran_CNCtran) ($p = 0.001$), (3) rearfoot frontal and calcaneocuboid frontal plane motion (RCfron_CCfron) ($p = 0.002$), (4) rearfoot transverse and calcaneocuboid transverse plane motion (RCtran_CCtran) ($p = 0.004$), and (5) calcaneocuboid frontal and lateral forefoot sagittal plane motion (CCfron_LFFsag) ($p = 0.008$) (Table 4.2).

Subphase 3

Significant foot strike main effects from subphase 3 mixed between-within subjects ANOVAs indicated decreased coordination variability in FFS runners between calcaneocuboid frontal and lateral forefoot sagittal plane motion (CCfron_LFFsag) ($p = 0.041$) compared to RFS runners (Table 4.2).

Subphase 4

Significant foot strike main effects from subphase 4 mixed between-within subjects ANOVAs indicated increased coordination variability in FFS runners for the medial forefoot sagittal and 1st metatarsophalangeal sagittal plane motion (MFFsag_MTPsag) ($p = 0.030$) (Table 4.2).

Discussion

The purpose of this study was to investigate the differences in foot and ankle joint coupling variability between habitual RFS and FFS runners at the beginning and end of an exhaustive run. The hypothesis that FFS runners would have more joint coupling variability during early stance was partially supported with medial midfoot-forefoot couplings. The hypothesis that joint coupling variability would decrease in both groups in response to the exhaustive run was not supported as all but one coordination pattern increased variability.

Foot strike differences

The original hypothesis that FFS runners would increase variability was supported for the CNCfron_MFFsag and MFFsag_MTPsag couplings during subphase 1 of stance. However, the variability of the CCfron_LFFsag during subphase 1 and the RCsag_CNCsag, RCsag_CCsag, and RCfron_CCfron during subphase 2 were actually decreased in FFS runners when compared to RFS runners. Additionally, contrary to the original hypotheses, there were also differences during late stance in CCfron_LFFsag (increased in RFS runners) and MTPsag_MFFsag (increased in FFS runners) variability between the foot strike patterns.

According to the dynamical systems theory, joint couplings that have increased variability may be more resistant to overuse injuries by distributing stress to different tissues within the system (Hamill et al., 1999). Based on this theory, the increased joint coupling variability observed in FFS runners at the CNCfron_MFFsag and MFFsag_MTPsag couplings may suggest a decreased injury risk at these medial foot couplings during the initial loading (subphase 1) of stance. Given that FFS runners likely require greater dynamic stability in the foot because of a less stable ankle position during early stance, it may be that dynamic stabilizers create more variability than bony stability. This may help prevent overuse to medial foot structures in FFS runners. It was surprising, however, that FFS runners had decreased CCfron_LFFsag coupling than RFS in subphase 1. It is possible that because the forefoot goes from being unloaded to loaded in RFS runners that this variability is greater in RFS runners. Additionally, it is possible that the lateral arch does not have as much dynamic stability as the medial arch and therefore has different neuromuscular control needs based on the amount of stability at the ankle and other foot joints. Because both of the groups of runners used for this study were uninjured, it is assumed that these variations in variability are healthy for each indicated foot strike pattern. Likewise, when comparing variability from injured runners to these findings, the differences identified suggest the importance of comparing variability to the appropriate foot strike group.

In the second stance subphase, the increased rearfoot-midfoot coupling variability (RCsag_CNCsag, RCsag_CCsag, RCfron_CCfron) in RFS runners may have resulted from an increased muscular response to the large impact loads observed in RFS running during the first stance subphase (Lieberman et al., 2010). The increased coupling variability between the lateral midfoot and forefoot in the RFS group during subphase 3 and between the medial midfoot and MTP during subphase 4 in the FFS group were not anticipated. Like with the first subphase of stance, the increased variability may derive from the need for varying stability demands for each strike pattern during the first and second subphases of stance. The differing coordination variability from the second half of stance may be in response to earlier kinematic differences that now require different coordination patterns to assure sufficient foot rigidity for push off.

When considering the variability in rearfoot frontal plane coupling with tibial transverse plane rotation, Dierks and Davis (2007) found within-subject variability to range from 6.8 – 8.4° in uninjured recreational runners throughout the four subphases of stance which is similar to the present findings with the exception of subphase 4 where the current study observed less than 2° of variability. Perhaps some of this difference can be explained by use of a treadmill rather than overground running where propulsion requirements may be different. In addition, Dierks and Davis (2007) placed motion capture markers on the shoe and did not use consecutive steps for analysis. Previous studies investigating coupling pattern within the foot have not partitioned the foot into medial and lateral midfoot and forefoot segments, rather the studies have only investigated forefoot-rearfoot coupling (Eslami et al., 2007; Pohl & Buckley, 2008). Due to the differing foot models, the distal foot coupling pattern results in the current study cannot be compared with previous studies. The medial and lateral mid and forefoot joint coupling variability differences between the foot strike patterns observed in the current study advocate for the importance of medial-lateral segmentation when investigating foot coupling.

Exhaustive run effect

Contrary to the initial hypothesis, not only did variability increase as a result of the exhaustive run, but the differences from the beginning to the end of the run occurred during subphase 2 rather than subphase 1. There was one foot strike by time interaction during subphase 2 in the CNCfron_MFFsag joint couple where variability did decrease in RFS runners at the end of the exhaustive run, but there was no change in FFS runners. The decreased variability in this coordination pattern may be a potential problem area for RFS runners.

The remainder of the fatigue effects were associated with increased variability during subphase 2 in both groups and most of the changes were observed in the rearfoot and midfoot joint couples. During subphase 2, the body continues to accept increasing ground reaction forces, but has passed the initial loading response that is characteristically different between FFS and RFS runners (Dierks & Davis, 2007; Lieberman et al., 2010). Originally, variability was hypothesized to decrease following the run due to central and/or peripheral exhaustion that would leave the system less able to adapt to varying constraints and disperse tissue stress. The inability to disperse stress to other tissues would result in overloading of the same tissue and possible injury.

Although the results were inconsistent with the our original hypothesis, they were consistent with the findings of a previous study that investigated leg and rearfoot joint coupling after an exhaustive tibialis posterior protocol in a non-injured population (Ferber & Pohl, 2011). Ferber and Pohl (2011) suggested that perhaps the neuromuscular system works to increase variability in healthy runners in order to avoid injury when muscles become exhausted, by utilizing alternate muscles for joint stability and force dissipation. The results of the current study may support the suggestion that in non-injured runners increased variability associated with an exhaustive run may function to avoid injury and still maintain stability. Therefore, perhaps variability decreases in already injured or unhealthy individuals, but healthy, pain-free runners are able to increase variability to avoid injury. However, too much variability is also potentially damaging as the inability to control coordination may lead to overstressing tissues or not providing sufficient stability at critical moments. Based on the fact that the runners used in the current study are healthy, non-injured runners, it is assumed that the variability experienced as a result of the fatiguing run are within the normal limits of acceptable variability and help to disperse force to varying tissue without passing the yielding point.

It was surprising that changes in variability were not seen during subphase 1 as a result of the exhaustive run as this is theorized to be where impact injuries occur (Lieberman et al., 2010). Perhaps the muscular response to high loading rates during subphase 1 are only first observed after the subphase is completed, which is why variability of the couples was only observed during subphase 2.

Conclusion

Joint coupling variability during subphase 1 of stance was increased in FFS runners between the medial midfoot to forefoot while RFS runners had increased variability between the rearfoot to lateral midfoot and lateral midfoot to lateral forefoot. These variability differences are assumed to be good, having occurred in uninjured runners, and clinically suggests that distal foot coordination variability may not be interpreted identically for all foot strike patterns. Despite joint coupling variability differences between RFS and FFS runners, both groups demonstrated similar increases in coupling variability, primarily between the rearfoot and midfoot functional articulations during the second stance subphase, at the end of an exhaustive 5 km run. In uninjured RFS and FFS runners, the increases in variability following the exhaustive run may have been the result of neuromuscular system compensations to maintain stability and disperse forces in order to prevent overuse injury. Although FFS runners may use more dynamic stabilizers during early stance to aid in stabilizing the foot and force dissipation, it does not appear that they were affected by fatigue any differently than in RFS runners. More studies investigating joint coupling variability in the foot and ankle as the result of exhaustion in both healthy and injured populations are needed to better understand the relationship between coupling variability and running injuries.

Table 4.2- Joint coupling coordination variability mean ± SD for RFS (top numbers) and FFS (bottom numbers) before and after a 5 km run. Variability was found for each of the four subphases of stance: (1) 0-20%, (2) 21-50%, (3) 51-75%, and (4) 76-100%.

		Pre-run				**Post-run**			
Articulation		1	2	3	4	1	2	3	4
RCtran_RCfron	RFS	4.76 ± 2.23	**5.01 ± 1.57**[a]	6.05 ± 1.31	1.18 ± .66	4.36 ± 1.45	**5.33 ± 1.53**[a]	6.08 ± 1.43	1.33 ± .61
	FFS	6.09 ± 2.08	**4.59 ± 1.95**[a]	5.88 ± 2.09	1.20 ± .61	5.80 ± 2.46	**5.62 ± 1.75**[a]	5.86 ± 2.32	1.14 ± .72
RCsag_CNCsag	RFS	5.82 ± 1.43	**6.29 ± 2.22**[b]	8.85 ± .88	1.42 ± .71	5.71 ± 1.74	**6.16 ± 2.20**[b]	8.57 ± .73	1.47 ± .64
	FFS	5.68 ± 1.87	**2.52 ± .91**[b]	8.30 ± .94	1.74 ± .62	5.65 ± 1.65	**3.07 ± 1.28**[b]	8.47 ± 1.07	1.88 ± .74
RCfron_CNCfron	RFS	4.88 ± 2.26	6.03 ± 1.47	5.98 ± 1.05	4.26 ± 1.75	4.65 ± 1.66	6.65 ± 1.80	5.65 ± 1.14	4.11 ± 1.83
	FFS	5.91 ± 1.68	5.56 ± 1.87	6.56 ± 1.81	4.54 ± 1.84	5.30 ± 1.99	6.00 ± 1.67	6.62 ± 1.53	4.34 ± 1.63
RCtran_CNCtran	RFS	4.31 ± 2.32	**4.59 ± 1.97**[a]	6.94 ± 1.47	2.15 ± .89	4.24 ± 1.90	**5.15 ± 2.21**[a]	7.20 ± 1.58	1.96 ± .64
	FFS	5.48 ± 1.68	**3.98 ± 1.62**[a]	1.02 ± 1.85	1.87 ± 1.01	4.97 ± 2.02	**5.10 ± 1.24**[a]	7.39 ± 2.06	2.07 ± .79
RCsag_CCsag	RFS	5.37 ± 1.36	**6.00 ± 2.18**[b]	8.53 ± .861	1.45 ± .80	5.61 ± 1.42	**6.14 ± 2.07**[b]	8.45 ± .940	1.31 ± .55
	FFS	6.74 ± 2.03	**2.23 ± 1.03**[b]	8.48 ± 1.23	1.52 ± .60	6.30 ± 1.77	**2.60 ± 1.21**[b]	8.44 ± .978	1.60 ± .52
RCfron_CCfron	RFS	5.44 ± 1.84	**7.26 ± 1.32**[a,b]	5.48 ± 1.65	3.42 ± 1.89	4.55 ± 1.78	**8.17 ± .99**[a,b]	5.70 ± 1.60	3.55 ± 2.19
	FFS	5.52 ± 2.01	**6.11 ± 2.14**[a,b]	5.18 ± 2.04	4.44 ± 2.01	5.34 ± 1.77	**6.74 ± 1.77**[a,b]	4.81 ± 1.69	4.02 ± 2.04
RCtran_CCtran	RFS	4.82 ± 2.18	**5.32 ± 1.99**[a]	6.12 ± 1.81	1.88 ± .90	4.57 ± 2.14	**5.60 ± 2.23**[a]	6.16 ± 1.82	1.61 ± .81
	FFS	5.74 ± 1.61	**4.26 ± 2.06**[a]	6.48 ± 1.65	1.75 ± .98	5.92 ± 1.87	**5.28 ± 1.79**[a]	6.61 ± 1.86	1.70 ± 1.02
CNCfron_MFFsag	RFS	**4.83 ± 1.97**[b]	**5.67 ± 1.80**[c]	8.28 ± 1.18	1.32 ± .52	**4.79 ± 1.57**[b]	**4.68 ± 1.63**[c]	8.49 ± 1.03	1.33 ± .65
	FFS	**5.97 ± 1.30**[b]	**2.29 ± .80**[c]	8.72 ± .76	1.21 ± .43	**6.09 ± 2.21**	**2.32 ± .45**[c]	8.55 ± .79	1.44 ± .46
MFFsag_MTPsag	RFS	**4.60 ± 2.04**[b]	5.12 ± 1.42	4.23 ± 1.44	**1.87 ± .34**[b]	**4.54 ± 2.05**[b]	5.00 ± 1.30	4.68 ± 1.26	**1.90 ± .26**[b]
	FFS	**7.51 ± 1.46**[b]	4.21 ± 1.17	3.75 ± .99	**2.08 ± .23**[b]	**7.30 ± 1.16**[b]	4.81 ± 1.06	3.69 ± .81	**2.11 ± .26**[b]
CCfron_LFFsag	RFS	**5.96 ± 1.36**[b]	6.26 ± 1.67	**7.20 ± 1.59**[b]	2.39 ± .76	**5.96 ± 1.45**[b]	**7.24 ± 1.55**[a]	**7.73 ± 1.19**[b]	2.47 ± 1.48
	FFS	**4.92 ± 1.46**[b]	6.82 ± 1.93	**6.33 ± 1.73**[b]	2.88 ± 1.49	**5.12 ± 1.58**[b]	**7.15 ± 1.56**[a]	**6.16 ± 2.17**[b]	3.21 ± 1.80

[a] significant time main effect (pre-run to post-run)
[b] significant difference between foot strikes (FFS and RFS)
[c] significant interaction

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5. DIFFERENCES IN LEG MUSCLE PREACTIVATION BETWEEN HABITUAL REARFOOT AND FOREFOOT RUNNERS AT THE BEGINNING AND END OF AN EXHAUSTIVE RUN

Introduction

As the number of running participants has steadily increased over the last few decades, the occurrence of running-related injuries has also escalated. Recent studies suggest that 19.4-79.3% of runners experiencing some type of overuse injury each season (Bahr & Holme, 2003; van Gent et al., 2007). Despite advances in footwear technology, running related injury rates have failed to decrease; leading some researchers to focus on foot strike patterns. This has also led some clinicians and researchers to recommend that runners switch from a rearfoot strike pattern (RFS) (the pattern utilized by 75-99% of runners) (Bertelsen et al., 2013; Hasegawa et al., 2007; Kasmer et al., 2013) to a midfoot (MFS) or forefoot strike pattern (FFS) (Daoud et al., 2012).

Rearfoot strike running has been associated with a vertical ground reaction force profile that includes a sharp impact peak and higher loading rate during the load acceptance phase of running when compared to MFS and FFS running (Cavanagh & Lafortune, 1980; Lieberman et al., 2010). These forces and loading rates are thought to be related to the development of overuse running injuries. While running, leg muscles contract prior to initial contact to aid in dissipating the forces that travel proximally up the lower extremity following ground impact (Komi et al., 1987). This preactivation is thought to reduce the occurrence of impact related injuries.

The association between lower extremity muscle preactivation and vertical ground reaction force in running has been investigated. Komi et al. (1987) reported that increases in vertical ground reaction force resulted in an earlier preactivation and increased preactivation amplitude of the lateral gastrocnemius. This characteristic was also supported by Wakeling et al. (2001) using a striking pendulum apparatus, which mimicked ground reaction force on the calcaneus. As the force of the pendulum increased, amplitudes pre- and post-activation of both the tibialis anterior and medial gastrocnemius muscles increased. Running with different foot strike patterns also induces variations in the neuromuscular profile of the leg musculature that may contribute to injury risk (Ahn et al., 2014; Shih et al., 2013). In a study investigating RFS and FFS running, Shih et al. (2013) reported habitual RFS runners to have increased preactivation of the tibialis anterior while RFS running and increased gastrocnemius preactivation during FFS running. However, in a study investigating natural FFS and RFS runners, Yong, et al. (2014) did not find differences in tibialis anterior and gastrocnemius root mean square muscle activity during early stance phase. The study did not look at preactivation and therefore, information about how habitual RFS runners differ in preactivation compared to habitual FFS runners is still needed. Given the common clinical belief that the presence of a sharp vertical ground reaction force impact peak and high loading rate increases injury risk, it is important to further understand how preactivation of leg muscles may be affected by a runner's foot strike pattern.

In addition to foot strike pattern, fatigue also results in variations in lower extremity neuromuscular activity which may adversely affect a muscle's ability to dissipate impact forces (Bobbert et al., 1992; Mizrahi et al., 2000a). Previous research has identified changes in ankle muscle co-contraction and delayed preactivation after exhaustive running protocols in the tibialis anterior muscle (Mizrahi et al., 2000a; Nyland et al., 1997). Although previous research has compared neuromuscular patterns in running with RFS and FFS patterns in a non-fatigued state, it is unknown if these neuromuscular profiles are similar after an exhaustive run.

The purpose of this study was to identify the leg muscular preactivation differences between habitual RFS and FFS runners at the beginning and end of an exhaustive run. It was hypothesized that FFS runners, when compared to RFS runners, would exhibit an earlier onset and larger magnitude of gastrocnemius and peroneal neuromuscular activity. Likewise, it was postulated that the tibialis anterior would have an earlier preactivation and larger magnitude in RFS runners. No difference between the groups was expected for the soleus. It was also hypothesized that post-exhaustive run preactivation would be delayed and diminished for all muscles in both foot strike groups. Further, the delay and decreased magnitude were hypothesized to be greater in FFS runners due to the initial contact position difference compared to RFS runners.

Methods

Subjects

Thirty volunteers (15 RFS, 15 FFS; 6 female, 24 male) participated in the study (Table 5.1). All participants were currently running a minimum of 10 miles per week and had used their current strike pattern for a minimum of 1 year. Participants were excluded if they had any lower extremity pain or injury within the previous 6 weeks, a history of major lower extremity surgery, wore custom molded orthotics, or had any known cardiovascular problems or uncontrolled asthma. Rearfoot strike running participants were matched for age and sex with FFS runners. Participation in the study consisted of a phone screen to verify inclusion and exclusion criteria (Appendix E) and two separate visits to the lab. During the first visit, after IRB approved informed consent was obtained, participants completed a running questionnaire to compare running experience between the groups (Appendix N) and a foot strike screen was conducted. Eligible participants were then scheduled for a running gait analysis visit.

***Protocol*.**

Foot strike verification

Participants warmed up for 5 minutes on a treadmill and were then instructed to complete five successful running trials (3.35 ± 0.34 m/s) along a 25 m force plate instrumented walkway (Advanced Medical Technologies, Inc, Waterford, MA) sampling at 1,000 Hz. The runners performed the trials in their own running shoes and then in a laboratory supplied running sandal (Mono, Luna Sandals, Seattle, WA). Following the trials, ground reaction force data were visually inspected for an initial impact peak which is characteristic of RFS running (Cavanagh & Lafortune, 1980; Lieberman et al., 2010). Individuals exhibiting an obvious initial impact peak in at least three of the five trials for both footwear conditions were placed in the RFS group while those without an initial impact peak were placed in the FFS group. Since both MFS and FFS runners tend to lack an initial impact peak, both were placed into the FFS group (Ahn et al., 2014; Lieberman et al., 2010). This method was repeated during the second visit's running analysis by positioning one leg of the treadmill on the force plate and visually examining the vertical ground reaction force to verify foot strike pattern did not change. At completion of the foot strike verification, participants were fitted for a pair of running sandals and given a running log. They were then instructed to gradually increase running distance in the sandals over a minimum of 10 days in order to accommodate to the new footwear prior to the follow-up data collection visit. Instructions were given to increase the distance run in the sandals by ½ mile every run while maintaining total mileage in their habitual running shoes for the remainder of their planned distance.

During this visit, participants also performed maximal voluntary isometric contractions (MVC) for ankle plantarflexion, dorsiflexion, and eversion to become familiar with performing maximal contractions. For the MVC testing, the subject was positioned in a Biodex dynamometer (Biodex Medical Systems, Shirley, NY) with the knee fully extended and the ankle in a neutral position. The participant pushed against the stationary arm of the dynamometer with maximal plantarflexion force for 5 s, rested for 5 s, then pulled against the stationary arm of the dynamometer with maximal dorsiflexion force for 5 s. This sequence was repeated 3 times. The MVC for eversion was conducted with the subject placed in a manual isometric muscle testing position. Manual resistance just distal to the base of the fifth metatarsal with the ankle in neutral was then applied (Rose, Burns, Ryan, Ouvrier, & North, 2008).

Running analysis visit

Participants wore a Polar heart rate chest monitor (Polar Electro Inc., Oulu, Finland) to observe exertion level. The right leg of the participant was prepared for EMG electrode placement by shaving and then cleansing the area with alcohol. Four circular 1.0 cm diameter dual Ag/AgCl electrodes (Noraxon Dual Electrodes- 2.0 cm interelectrode distance, Noraxon USA, Inc., Scottsdale, AZ) were positioned on the skin according to SENIAM recommendations (Hermens et al., 1999) for the medial gastrocnemius, soleus, peroneus longus, and tibialis anterior. Electrodes were secured with a self-adhesive tape (PowerFlex, Andover Healthcare Inc, Salisbury, MA). An 8-channel Noraxon EMG system (Telemyo 900, Noraxon, Scottsdale, AZ) sampling at 1000 Hz was used to measure electrical activity of each muscle. The wireless EMG unit was placed in a secured pack on the back of the participant for the running protocol.

Prior to the exhaustive run, MVCs were collected for ankle plantarflexion, dorsiflexion, and eversion identical to methods used during their first visit and used to normalize EMG data. Maximal plantarflexion and dorsiflexion torque was also recorded pre- and post- exhaustive run as an indicator of muscle exhaustion.

Following MVC testing, clusters of four retroreflective markers (6.4 mm diameter) were placed on the right leg and foot to identify the leg and six foot segments (Bauer et al., 2011). Three-dimensional positional marker data were collected using a 10-camera Eagle Motion Analysis System (Motion Analysis Corp., Santa Rosa, CA) sampling at 200 Hz. Positional data of the toe tip from the multi-segment foot model was used to identify initial contact (Zeni et al., 2008).

Running protocol

Subjects warmed up for 5 minutes on a treadmill located in the center of the capture volume. After the warm up, the subjects performed a maximal effort 5 km run (Laursen et al., 2007). They began the run at a speed of 7.5 mph ($3.4\ m \cdot s^{-1}$). Once a consistent gait pattern was observed, EMG activity was collected for 10 seconds. Following the 10 s data collection, subjects were able to adjust the treadmill speed to control their own pace throughout the run. Subjects were considered to have reached an exhausted state if their heart rate was equal to or greater than 80% of age-predicted heart rate max (220-age) or if they had a rate of perceived exertion at or above 17 out of 20 on a Borg scale (Borg, 1970). All subjects reached both criteria. Heart rate and rate of perceived exertion were assessed every five minutes during the run and at the completion of the run. In the final 0.16 km, the treadmill was set to 7.5 mph (3.4 $m \cdot s^{-1}$). As soon as the subject adjusted to the speed, a second 10 s data collection was performed. Following the data capture, the participant continued at a self-selected speed until 5 km were completed.

Data Processing

From the 10 s data collections, the kinematic and EMG data from five subsequent running stance phases were identified and processed. Initial contact was determined as the point when the toe tip reached maximal anterior position (Zeni et al., 2008).

EMG processing

The amplified EMG signals were processed with a custom Matlab program (Matlab R2013a, Mathworks, Natick, MA). The signal was trimmed from 250 ms prior to initial contact until initial contact. The signal was then processed using a bandwidth filter of 10-500 Hz with a fourth order zero lag Butterworth filter followed by full wave rectification. Finally, the signal was smoothed using a Root Mean Square (RMS) algorithm with a 0.05 s running window (Eq. 5.1):

Equation 5.1. $$RMS(t) = \sqrt{{}^{1}/_{T} \int_{t}^{t+T} EMG^{2}(\sigma)d\sigma}$$

where, *T* is the period of time that the smoothing is being performed (0.05 s), *t* is the exact time point in the trial that is being considered, and *σ* represents the data points within the smoothing window (De Luca, 2006). Onset of preactivation was identified when the EMG signal exceeded a threshold of 2 standard deviations (SD) above the baseline muscle activity prior to initial contact of the stance phase. The integrated EMG (iEMG) was also calculated from onset to initial contact (preactivation period) to quantify the amount of activation occurring in the muscle prior to ground contact. The pre- and post-exhaustive run EMG signals for the five stance phases for each subject were averaged and then ensemble averaged for each foot strike group.

Statistical Analyses

Independent *t*-tests were used to investigate weekly mileage and 5 km completion time differences between FFS and RFS runners. Mixed between-within subjects analyses of variance (ANOVA) were used to compare step frequency and plantarflexion and dorsiflexion MVCs. The between subjects factor was foot strike and the within subjects factor was pre- and post-exhaustive run. Likewise, mixed between-within subjects ANOVAs were used to compare the medial gastrocnemius, soleus, peroneus longus, and tibialis anterior muscles' EMG onset and preactivation iEMG prior to and following the exhaustive run. The significance level for all of the statistical analyses was defined as $\alpha = 0.05$.

Table 5.1- Participant descriptive data (N = 30).

Variable	RFS (n=15)	FFS (n=15)
Age (years)	27.7 ± 5.05	27.4 ± 6.34
Height (cm)	179 ± 7.69	179 ± 7.02
Mass (kg)	77.9 ± 10.3	61.2 ± 33.6
Weekly distance (mi)	20.2 ± 11.5	22.4 ± 15.8
Experience running over 10 mi (years)	4.82 ± 4.47	6.40 ± 5.59
5K completion time during testing (min)	23.2 ± 4.03	23.2 ± 4.03
Mean comfort level (1-5)	3.51 ± 0.61	3.33 ± 0.64
Final comfort level (1-5)	3.20 ± 0.92	3.09 ± 0.83
Step frequency-Pre-run (steps/min)	167 ± 10.4	174 ± 9.75
Step frequency-Post-run (steps/min)	164 ± 11.3*	172 ± 12.2*
Pre-run plantarflexion MVC (Nm)	144 ± 38.0	144 ± 31.5
Post-run plantarflexion MVC (Nm)	129 ± 25.5*	136 ± 32.1*
Pre-run dorsiflexion MVC (Nm)	26.4 ± 9.55	24.6 ± 9.36
Post-run dorsiflexion MVC (Nm)	23.7 ± 6.80*	21.6 ± 6.63*

* Significant difference from pre-run measures ($p < .05$)

Results

All participants met the exhaustion criteria by the end of the 5 km exhaustive run. According to statistical tests, there were no significant differences between RFS and FFS runners in weekly mileage ($P = .67$), 5 km completion time ($P = .99$), or step frequency ($P = .06$) (Table 5.1). However, both groups exhibited a significant decrease in step frequency following the exhaustive run ($P = .02$) (Table 5.1). There were no differences between FFS and RFS runners' pre-run plantarflexion ($P = .75$) or dorsiflexion ($P = .52$) MVCs. Plantarflexion ($P = .02$) and dorsiflexion ($P = .006$) post-run MVCs were diminished for both FFS and RFS runners.

Onset timing

Due to several incidences of electrode detachment or wire lead malfunction during testing, EMG data from all of the muscles assessed were not available for all of the subjects. All EMG data that was viable was processed and sample size was adjusted accordingly (Table 5.2). Mixed between-within subjects ANOVAs revealed no significant time-by-foot strike interactions for the medial gastrocnemius ($P = .17$), soleus ($P = .43$), peroneus longus ($P = .52$), or tibialis anterior ($P = .11$). Significant between-subjects (foot strike) main effects revealed the medial gastrocnemius had an earlier onset in FFS runners ($P = .01$) and the tibialis anterior had an earlier onset in RFS runners ($P = .02$). There were no significant between-subject main effects for the soleus ($P = .10$) or the peroneus longus ($P = .44$). There were no significant time main effects for the medial gastrocnemius ($P = .67$), soleus ($P = .19$), peroneus longus ($P = .62$), or tibialis anterior ($P = .69$).

Table 5.2- Rearfoot striker (RFS) and forefoot striker (FFS) pre- and post- exhaustive run preactivation onset timing.

Muscle	Foot strike	Pre-run (ms)	Post-run (ms)	N
Medial Gastrocnemius	RFS	68 ± 60	80 ± 40	13
	FFS	124 ± 39*	101 ± 50*	14
Soleus	RFS	75 ± 63	68 ± 56	9
	FFS	121 ± 51	91 ± 45	11
Peroneus Longus	RFS	60 ± 52	57 ± 53	13
	FFS	73 ± 46	91 ± 53	9
Tibialis Anterior	RFS	147 ± 64	119 ± 96	7
	FFS	37 ± 47*	83 ± 65*	9

* Significant between-subject difference when compared to RFS runners ($p < .05$)

Integrated EMG

Mixed between-within subjects ANOVAs did not reveal any time-by-foot strike interaction effects for medial gastrocnemius (P = .13), soleus (P = .88), peroneus longus (P = .68), or tibialis anterior (P = .74). Significant between subjects main effects indicated an increased amount of tibialis anterior iEMG in RFS runners when compared to FFS runners (P = .01), but no difference between groups for the medial gastrocnemius (P = .23), soleus (P = .81), or peroneus longus (P = .50). Finally, there were no significant within-subjects (time) main effects for the medial gastrocnemius (P = .65), soleus (P = .26), peroneus longus (P = .44), or tibialis anterior (P = .28) (Table 5.3).

Table 5.3- Rearfoot striker (RFS) and forefoot striker (FFS) pre- and post- exhaustive run integrated EMG

Muscle	Foot strike	Pre-run (%MVC·s)	Post-run (%MVC·s)
Medial Gastrocnemius	RFS	2.58 ± 3.10	6.79 ± 13.05
	FFS	8.35 ± 4.32	6.03 ± 5.95
Soleus	RFS	6.29 ± 5.65	4.81 ± 4.25
	FFS	6.60 ± 5.58	5.46 ± 4.35
Peroneus Longus	RFS	3.19 ± 3.16	3.56 ± 3.80
	FFS	2.06 ± 1.71	3.29 ± 3.83
Tibialis Anterior	RFS	6.51 ± 1.98	9.38 ± 11.58
	FFS	1.28 ± 1.05*	2.83 ± 2.25*

* Significant between subject difference when compared to RFS runners ($p < .05$)

Discussion

The purpose of this study was to compare neuromuscular preactivation patterns in habitual RFS and FFS runners at the beginning and end of an exhaustive run. The hypothesis of FFS runners having an earlier onset of medial gastrocnemius activity compared to RFS runners and RFS runners having an earlier onset of tibialis activity compared to FFS runners was supported by the results. However, only the iEMG of the tibialis anterior, and not the medial gastrocnemius, differed between the groups, which is inconsistent with our initial hypothesis. It was also hypothesized that FFS runners would have an earlier onset of peroneus longus activity, but this was not supported by the present data. Likewise, the lack of significant differences in preactivation timing or iEMG values when comparing muscle activity at the beginning of the run to the end of the run were not consistent with our initial hypotheses.

Onset timing

The onset of medial gastrocnemius activity in FFS runners occurred 21-56 ms earlier than for the RFS group which is similar to previously reported results (Ahn et al., 2014; Divert et al., 2005b; Shih et al., 2013). Preactivation allows the muscle's contractile unit to create tension in the Achilles' tendon prior to initial contact in preparation for the eccentric contraction during the loading phase in FFS runners (Bobbert et al., 1992). The muscle tension is important for the dissipation of forces and joint stability following foot contact with the ground. The delayed gastrocnemius onset in the RFS runners was likely due to the fact that the RFS runners make initial contact in a dorsiflexed ankle position. This dorsiflexed position at initial contact also explains the 36-110 ms earlier onset of the tibialis anterior seen in RFS runners, which also is similar to previously reported results (Shih et al., 2013; von Tscharner et al., 2003).

As hypothesized, the soleus onset timing did not differ between the groups, despite the common insertion to the Achilles' tendon with the gastrocnemius. It has been suggested previously (Jacobs et al., 1993) that the soleus may play a different role during the loading phase of stance while running than the gastrocnemius, with the gastrocnemius contributing more to force dissipation. If this is the case, both foot strike patterns would to utilize the soleus for elastic energy storage to the same extent.

Because FFS runners contact the ground with increased ankle inversion and plantarflexion, (Williams et al., 2000) it was initially hypothesized that they may also have an earlier peroneus longus onset to provide increased ankle stability at ground contact. With an increased inversion angle and the foot in a plantarflexed position, the ankle is in a less stable position and more prone to inversion ankle sprains. Even though FFS runners have increased ankle inversion at initial contact, RFS runners also land in an inverted position (Williams et al., 2000). Therefore, although the degree of inversion may differ between the foot strikes, the peroneus longus preactivation patterns required to maintain lateral ankle stability appear to be similar.

Although it was hypothesized that preactivation patterns would change pre- to post-run, this was not observed for any of the four muscles. The plantarflexors and dorsiflexors did experience some level of fatigue as evidenced by the decreases in peak MVC following the exhaustive run (Table 5.1), but it appears that the maximal strength capabilities of the muscle did not affect timing of the contraction when running.

Integrated EMG

Although differences between the foot strikes were observed in EMG onset timing between the medial gastrocnemius and the tibialis anterior, there was only a difference in the iEMG for the tibialis anterior. Despite the earlier onset of the medial gastrocnemius in FFS runners, a similar magnitude of gastrocnemius contraction was present in both running styles prior to initial contact. These results indicate that RFS and FFS runners have similar gastrocnemius activation levels immediately prior to initial contact, however, the purpose of the muscle activity may differ. Because RFS runners exhibit a ground reaction force impact peak and loading rate of up to 7 times that of FFS runners, (Lieberman et al., 2010) it is likely that RFS runners utilize the gastrocnemius to aid in the dissipation of the impact forces during the loading phase of stance. With absence of an impact peak and a lower loading rate in FFS running, the gastrocnemius activity may instead contribute more to ankle stability and elastic energy storage during the loading phase.

The increased preactivation iEMG in the tibialis anterior of RFS runners was likely the result of having to maintain the ankle in a dorsiflexed position for initial contact. Unlike the gastrocnemius which may have similar iEMG preactivation between the foot strikes but for different functional purposes, the dorsiflexed ankle position in RFS runners requires larger amounts of tibialis anterior preactivation. A dorsiflexed position when FFS running would not allow for correct foot position at initial contact and essentially cause the runner to strike the ground with the rearfoot.

Likewise with onset of preactivation, there were no changes in pre- to post-run iEMG values which goes against the initial hypothesis. It is possible that preactivation iEMG did change at different time points within the run, but only pre- and post- values were evaluated. Wu, et al. (2007) evaluated gastrocnemius and tibialis anterior EMG activity during a 20 minute run and reported an increase in peak EMG activity in the middle of the run, that returned to initial levels after 20 minutes. This would likely be the result of larger motor units being activated mid-run and then exhausted by the end. Since only pre- and post- measures were taken for the current study, it is unclear if a similar preactivation pattern would have been observed.

Although the assessment of step frequency was not a primary purpose of the study, the differences between the current and previous studies requires additional discussion. Unlike previous studies, there was no difference between the RFS and FFS runners' step frequency. Because FFS runners have a greater knee flexion angle to allow for forefoot contact with the ground, shorter steps are generally needed to accommodate the differing kinematics, making the step frequency higher for a given speed (Williams et al., 2000). In the present study, statistical tests were nearing significance ($P = 0.06$) and finding a larger sample may provide a statistically different sample. However, it is also possible that by using habitual FFS runners instead of asking habitual RFS runners to run with a FFS, and having all runners wear similar footwear, that the difference between the groups was not observed.

Further, both groups decreased their step frequency at the end of the exhaustive run suggesting that both foot strike patterns increased their stride length as exhaustion set in. Larson, et al. (2011) reported that a large number of FFS runners had changed their foot strike pattern during a long distance running event, which may be the result of increasing step frequency. All runners in the current study, however, continued to maintain their original strike pattern. It is suggested, therefore, that perhaps a longer run duration may be beneficial to compare exhaustive running muscle profiles.

Limitations of the study

Before drawing conclusions based on the results discussed, it is important to identify the limitation of this study. One limitation of this study was the limited number of complete EMG profiles achieved from each subject. Wire lead malfunction during the run often allowed for pre-run data, but not post-run data. Additionally, because of the nature of the exhaustive running protocol and securing the EMG electrodes and leads with cohesive bandages, perspiration may have greatly affected EMG readings by increasing conduction of the electrical signal when comparing post-run to pre-run results. Another limitation is that data was only analyzed at the beginning and end of the 5 km run, preactivation patterns that may have changed during the run were not available. Finally, because this study only compared preactivation data, conclusions as to how the musculature was activated during the stance phase was not available. This may be important to understand as it is during stance that ankle stability and force dissipation are needed.

Conclusion

In conclusion, only the tibialis anterior and medial gastrocnemius onset timing and tibialis anterior iEMG differed between the foot strikes and there were no changes in muscle activity timing or magnitude following the exhaustive run in either group. The increased tibialis anterior activity helps to maintain a dorsiflexed ankle joint position for initial contact in RFS runners. Despite similar iEMG values of the medial gastrocnemius in FFS and RFS runners, the function of the muscular activity may differ between the foot strikes. The gastrocnemius preactivation activity in FFS runners may function primarily for elastic energy storage and ankle stability, while RFS runners may utilize the preactivation for dissipation of the much larger vertical ground reaction force and loading rate associated with making initial contact with the heel. Further, the lack of differences in lower extremity muscle preactivation following the 5 km exhaustive run protocol suggest that neuromuscular mechanisms prior to initial contact may not be altered in experienced RFS or FFS runners. It remains unclear if increased distance, higher intensity intervals, or other training factors may influence preactivation of these muscles while running.

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6. SUMMARY AND CONCLUSIONS

The purpose of this study was to identify differences in foot kinematics and lower extremity muscular activation between habitual RFS and FFS runners before and after an exhaustive run. The main objectives to achieve this purpose were to (1) explore differences in discrete joint kinematics of the foot using a multi-segment foot model, (2) compare foot joint coupling variability, and (3) evaluate leg musculature preactivation onset and magnitude via iEMG.

Fifteen habitual RFS runners (27.7 ± 5.05 years, 179 ± 7.69 cm, 77.9 ± 10.3 kg) and 15 age and sex matched FFS runners (27.4 ± 6.34 years, 179 ± 7.02 cm, 61.2 ± 33.6 kg) participated in this study. After foot strike verification procedures, all participants were given 12 mm sole running sandals (Mono sandal, Luna, Seattle, WA) and a 10 day accommodation program. After accommodation, participants returned for testing and performed a maximal 5 km treadmill run with appropriate retroreflective markers on their right foot to identify a seven segment foot model and EMG electrodes positioned over the medial gastrocnemius, soleus, peroneus longus, and tibialis anterior. Motion capture and EMG data were captured both at the beginning and end of the exhaustive run. Motion capture data was processed for five consecutive steps at both time points to identify single joint kinematics for the following function articulations: rearfoot complex, calcaneonavicular, calcaneocuboid, medial forefoot, lateral forefoot, and 1st MTP. With the single joint kinematics, vector coding measurements were calculated and used to identify joint coupling variability of adjacent foot articulations. Neuromuscular preactivation onset and iEMG were also identified for the four leg muscles and compared between and within the groups.

Foot strike differences

Differing joint kinematics between RFS and FFS runners revealed that FFS runners appear to have a more supinated foot and ankle position at initial contact and through early stance. Supination in the ankle is an unstable position, but supination in the foot is thought to create a more rigid foot. Because FFS runners rely on foot stability while making initial contact with the distal foot, a more supinated foot posture will allow for the necessary bony foot stability. The supinated foot in the beginning of stance may also prevent overpronating, or reaching early foot pronation which has been theorized to contribute to injuries such as plantar fasciitis and tibial stress fractures (Pohl et al., 2009; Pohl et al., 2008; Wilder & Sethi, 2004). The ankle in FFS runners, however, may be more at risk for acute injuries such as ankle sprains. Given that the participants in the current study were healthy, non-injured runners, future research should investigate these ideas by comparing with injured running populations.

Although single joint kinematics are useful in understanding motion, joint coupling has become a common measurement to assess how the kinetic chain works to coordinate movement. Greater variability in these coordination patterns has been associated with the ability to avoid injury by being able to adapt to varying stresses (Hamill et al., 1999). Between the foot strikes, variability increased with medial midfoot-forefoot couplings in FFS runners and increased variability in lateral rearfoot-midfoot and midfoot-forefoot couplings in RFS runners. This medial and lateral trend suggests that each foot strike pattern has different areas of the foot that may be more susceptible to running related injuries.

Between the foot strikes, the RFS runners had increased tibialis anterior preactivation magnitude (iEMG) and an earlier onset of activation than FFS runners. This would be necessary for RFS runners to maintain their dorsiflexed foot posture for initial contact. The FFS runners, in contrast, had an earlier medial gastrocnemius onset, but no difference in preactivation iEMG. Although iEMG values were similar for both foot strike patterns, the earlier gastrocnemius onset may indicate different muscular activity function between the foot strikes. The preactivation activity in FFS runners may aid in elastic energy storage and ankle stability while the preactivation in RFS runners may aid in force dissipation of the large loading rates which are characteristic of RFS running (Lieberman et al., 2010).

Exhaustive Run Effect

As a result of the exhaustive run, both RFS and FFS groups had increased range of motion for subphases 1, 3, and 4 of stance. The increased range of motion during early stance resulted in a more pronated foot and therefore more mobility which would increase stress on dynamic stabilizers. Range of motion increases in subphases 3 and 4 resulted in a more supinated foot and more rigidity which would be necessary for adequate push-off. Only at the medial forefoot and during the second subphase of stance were RFS kinematics observed to be different compared to FFS runners' kinematics at the end of the exhaustive run. During this subphase, RFS runners increased their medial forefoot dorsiflexion range of motion while FFS runners did not change.

When comparing the foot strike patterns during an exhaustive run, both groups demonstrated increases in joint coupling variability at the end of the run. This increase may be the neuromuscular system's response to exhaustion and its attempt to avoid overstressing tissues and injury (Ferber & Pohl, 2011). Where other studies have reported decreased variability in injured populations (Hamill et al., 1999; Herb et al., 2014; Miller et al., 2008), variability in healthy populations may increase in order to disperse stress to various tissue and therefore avoiding the breakdown of tissues through repetitive forces.

Despite the many differences in kinematics and joint coupling variability observed at the end of the exhaustive run, preactivation patterns of the four muscles investigated were not found to change in either group. Preactivation of the leg muscles helps to assure proper foot positioning and stability at initial contact as well as potentiation for a successful stretch shortening cycle (Aura & Komi, 1986). Despite the kinematic changes in initial contact angle and during early stance at the end of the run, the timing and magnitude of the muscles in preparation for these events does not appear to be altered.

The varying foot kinematics and neuromuscular preactivation observed between the foot strike patterns suggests that RFS and FFS runners likely are at risk for different types and locations of running related injuries. Although FFS running is thought to be a more “natural” form of running (Bramble & Lieberman, 2004; Lieberman et al., 2010), overall injury rates appear to be similar between the foot strike patterns (Goss & Gross, 2012; Warr et al., 2015). As this study was largely exploratory in nature and conducted on uninjured runners, no conclusion can be made as to which foot strike pattern may have more ideal foot kinematics. Therefore, more research is needed to understand how the varying kinematics and kinetics between the foot strike patterns contributes to specific running related injuries.

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APPENDICES

Appendix A: Multi-Segment Foot Model

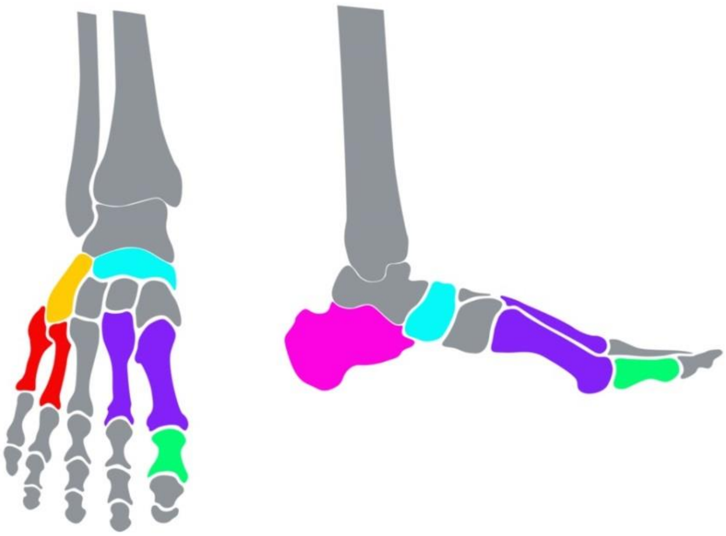


Figure A- The segments identified by the six-segment foot model: calcaneus (pink), navicular (blue), cuboid (yellow), medial rays (purple), lateral rays (red), and hallux (green).

Table A- Anatomical landmarks identified using the Davis Digitizing Pointer and surface markers.

Segment	Anatomical Landmark Identified	
Calcaneus	Sustentaculum Tali	Identified by Davis Digitizing Pointer
	Apex of Calcaneus	
	Peroneal Tubercle	
Navicular	Proximal, dorsal corner	
	Proximal, plantar corner	
	Distal, plantar corner	
Cuboid	Proximal, dorsal corner	
	Proximal, plantar corner	
	Distal, plantar corner	
Hallux	Base of hallux	
	Distal tip of hallux	
	Medial hallux	
Medial forefoot	1st and 2nd metatarsal heads	Identified by surface marker
Lateral forefoot	4th and 5th metatarsal heads	
Leg	Medial malleolus Lateral malleolus Tibial tuberosity	

Appendix B: Joint Coordinate Systems Definitions

Rearfoot functional articulation. The motions of the calcaneus with respect to the leg were formed using the mediolateral axis of the leg segment, the anteroposterior axis of the calcaneus, and the floating axis as computed by their cross product.

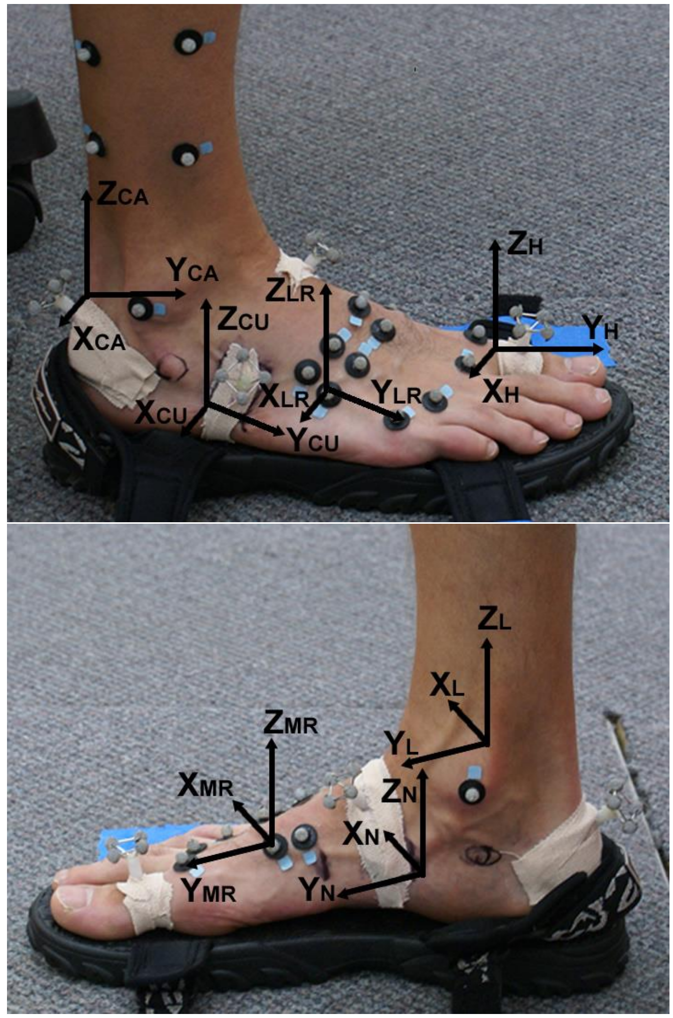
Midfoot functional articulations. The midfoot consists of two functional joints, the calcaneus and navicular articulation (calcaneonavicular) and the calcaneus and cuboid articulation (calcaneocuboid). For each of these articulations, the JCS was defined by the mediolateral axis of the calcaneus, the anteroposterior axis of the distal segment (the navicular or cuboid), and a floating axis calculated by their cross-product.

Forefoot functional articulations. Similar to the midfoot functional articulations, the forefoot consisted of a medial and lateral JCS to define the articulation of the medial rays (metatarsals 1 and 2) with respect to the navicular and lateral rays (metatarsals 4 and 5) with respect to the cuboid. The JCS of the medial forefoot functional articulation was defined by the mediolateral axis of the navicular, anteroposterior axis of medial rays, and a floating axis of their cross product. The lateral forefoot functional articulation was defined by the mediolateral axis of the cuboid, anteroposterior axis of the medial rays, and a floating axis of their cross product.

First metatarsophalangeal functional articulation. Three-dimensional motion of the 1st MTP (motion of hallux with respect to the medial rays) were calculated from the JCS formed by the mediolateral axis of the medial rays, anteroposterior axis of the hallux, and a floating axis of their cross product.

Appendix C: Multi-Segment Foot Model Coordinate Systems

Figure C- Lateral (above) and medial (below) view of the marker positioning for the multi-segment foot model. Cartesian coordinates are labeled for the calcaneus (CA), cuboid (CU), lateral rays (LR), hallux (H), medial rays (MR), navicular (N), and leg (L).



Appendix D: Statistical Power Analysis

A *priori* power analysis was done to find an appropriate sample size for the present study. Using joint coupling means and standard deviations from previously collected walking data using the same foot model (Table D), the Cohen's d value and effect size were calculated. The effect sizes from the sample data indicated a medium effect size (Cohen's d large effect size = 0.8, medium = 0.5, and small = 0.2). Calculations for a medium effect size (Cohen's $f = 0.25$, $\alpha = 0.05$, and $\beta = 0.20$) indicated that a minimum of 24 subjects were needed.

Table D. Mean and standard deviation of vector coding angles for individuals walking in two different shoe conditions.

Vector coding joint	Mean 1	SD1	Mean2	SD2	Cohen's d	Effect Size
Leg-Rearfoot transverse plane	70.48	4.94	15.64	41.24	1.87	0.68

Appendix E: Inclusion and Exclusion Criteria

Inclusion criteria

- Run minimum of 10 miles per week
- Use of either a rearfoot or mid/forefoot strike pattern for a minimum of 1 year
- Age 18-40 years old
- Have previously ran a 5 km run
- Able to run briefly at 3.35 $m \cdot s^{-1}$ (7.5 mph) at completion of a 5 km run

Exclusion criteria

- Lower extremity injury within the previous 6 weeks
- Major lower extremity surgery
- Wear custom molded foot orthotics
- Cardiovascular problems
- Uncontrolled asthma

Appendix F: Phone Screen Interview

How old are you? _________

How many miles do you average running per week? ______________

Have you had any previous lower extremity injury in the past 6 weeks that prevented you from continuing your normal running activity? □ Yes □ No

If yes, what injury?
Hip ____
Knee ____
Ankle ____

Do you have a history of any lower extremity surgery? □ Yes □ No
If so, what was done? ____

When you run, what part of the foot makes contact with the ground first?
□ Heel □ Midfoot □ Forefoot

Have you ever been diagnosed with any foot pathologies? _____________________

Have you ever been told you have high or low arches? □ Yes □ No

What is your shoe size? ______________

Do you have a history of cardiovascular disease? □ Yes □ No

Do you have asthma or exercise induced asthma? □ Yes □ No

If yes, do you use medication to control it? □ Yes □ No

Do you have a history of asthma or exercise induced asthma? □ Yes □ No

Female participants: Are you pregnant? □ Yes □ No

Appendix G: Health History Questionnaire

For your safety, a list of conditions that would make you unable to participate in this study has been prepared. Please answer the following questions to the best of your ability to assure your safety and inclusion in this study:

Sex ☐ M ☐ F

☐ Yes ☐ No Are you between the ages of 18 and 40 years old?

What's your age? ___________

☐ Yes ☐ No Do you run at least 10 miles a week regularly?

☐ RFS ☐ FFS Do you consider yourself a rearfoot (RFS) or forefoot (FFS) striker?

☐ Yes ☐ No Have you ran for at least one year with your current strike pattern?

☐ Yes ☐ No Do you have a history of asthma or exercise induced asthma?

If yes, is it controlled by medication? ☐ Yes ☐ No

☐ **Yes** ☐ **No** **Do you have any known sensitivity to any liquid adhesives or methyl salicylate**?

☐ Yes ☐ No Do you have a history of cardiovascular disease?

☐ Yes ☐ No Have you ever been told by a health care provider to avoid vigorous exercise?

☐ Yes ☐ No Have you ever had a lower extremity injury that caused you to decrease the amount of physical activity you undertake? If yes, please complete the following:

☐ Yes ☐ No Hip injury(ies)

If yes, approximately how many injuries?__________________________

☐ Yes ☐ No Knee injury(ies)

If yes, approximately how many injuries? __________________________

☐ Yes ☐ No Ankle/foot injury(ies)

If yes, approximately how many injuries? __________________________

☐ Yes ☐ No Have you had a lower extremity injury, in the last 6 weeks, that caused you to decrease the amount of miles you run per week?

If yes explain____

☐ Yes ☐ No Do you currently have any lower extremity pain or injury(ies)?

☐ Yes ☐ No Have you ever had major orthopedic surgery on your lower extremities?

☐ Yes ☐ No Have you been diagnosed with a foot pathology (i.e. hallux valgus)?

☐ Yes ☐ No For female participants: Are you pregnant or do you have reason to believe that you may be pregnant?

☐ **Yes** ☐ **No** **Are you currently employed by UWM?**

Comments/Notes:

Appendix H: Informed Consent Form

UNIVERSITY OF WISCONSIN – MILWAUKEE
CONSENT TO PARTICIPATE IN RESEARCH

THIS CONSENT FORM HAS BEEN APPROVED BY THE IRB FOR A ONE YEAR

1. General Information

Study title:

- Foot joint coupling and EMG patterns in habitual forefoot and rearfoot runners

Person in Charge of Study (Principal Investigator):

- The Principal Investigator (PI) for this study is Stephen Cobb, PhD, ATC, CSCS. Dr. Cobb is a faculty member in the Department of Kinesiology. The co-principal investigator for the study is Rhiannon Seneli, MS, ATC. Rhiannon is a doctoral student in the College of Health Sciences. Emily Gerstle is another co-investigator on this study. Emily is a masters student in the Department of Kinesiology.

2. Study Description

You are being asked to participate in a research study. Your participation is completely voluntary. You do not have to participate if you do not want to.

Study description:

The purpose of this project is to identify what foot motion and muscular activity differences exist between the different running patterns to better understand patterns of musculoskeletal overuse injuries commonly seen in runners. This study will compare runners who run with a rearfoot strike pattern (heel-to-toe strike) to those who run with a forefoot strike pattern (toe-to-heel). A foot model that divides the foot into 6 segments will be used to evaluate foot motion and how foot motion is coupled with adjacent joints. The muscle activity of the leg muscles will be compared between the two running patterns as well. As biomechanics of motion can change after fatigue, observation will take place before and after a fatiguing run.

This study will further the current knowledge on the differences between forefoot and rearfoot strike running mechanics. The results will provide clinicians and researchers invaluable information regarding the effect of foot strike pattern on running mechanics and joint coupling. Such knowledge will be an important component to comprehensive intervention programs that alter running mechanics to help prevent running related injuries.

Initial participant screening will be done via telephone call and further screening will take place along with, data collection, analyses, and storage in the Musculoskeletal Injury Biomechanics laboratory (Enderis 132); 15 habitual runners (age 18 – 45 years) with rearfoot strike patterns and 15 habitual runners (age 18 – 45 years) with forefoot strike patterns will be recruited from the University, surrounding community, and local running and triathlon groups.

As a participant in this study, you will be asked to participate in **two study sessions** in the lab where you will participate in an evaluation of running for identification of strike pattern, maximal voluntary contractions of ankle musculature, motion analysis of running mechanics before and after a fatiguing run on a treadmill. The total estimated completion time is 2.5-3 hours across two testing sessions.

3. Study Procedures

What will I be asked to do if I participate in the study?

If you agree to participate you will be asked to report to the Musculoskeletal Injury Biomechanics Laboratory (END 132) for testing. All procedures and measurements involved in the testing session will be performed by the PI or a co-investigator. Information of participants who do not qualify will be destroyed.

INITIAL PHONE SCREEN (~5-10 min)

- *Medical History Questionnaire*
    - Includes questions pertaining to your physical activity level and previous lower body injury(ies) and surgeries, pregnancy, and presence of diseases/illness that may exclude participation. This also includes questions about your age.

To participate in the study, you must meet the following criteria

- Age 18-45
- Run a minimum of 10 miles per week
- Female participants must have a women's size 6 -10 shoe size
- Male participants must have a men's size 7-13 shoe size
- Must never have had major surgery to the lower extremity
- Must have not had an injury to the lower extremity in the last 6 weeks
- Must not have diagnosed asthma or exercise induced asthma that is not adequately controlled with medication to permit moderate intensity exercise
- Must not have a history of cardiovascular disease
- Female participants must not be pregnant

IN-PERSON SCREENING (Visit 1) (~30 min)

- *Informed Consent Process*

If you agree to participate in the study, you will be asked to complete the following:

- *Foot strike pattern evaluation*

  You will be asked to run over a walkway and force plate to evaluate running form. Ground reaction force as measured by the force plate and video of your foot (video does not record above the waist) will be used to identify foot strike pattern.

- *Foot posture screen*

  A digital photograph of each of your feet will then be taken while you are standing with 10% of your body weight on the foot to be measured and also with 90% of your body weight on the foot to be measured. This photograph will not have any of your identifiers attached to it, so individuals who see it will not know that it is your foot. If you choose not to have your foot photographed, you will not be eligible to complete the study.

ELECTROMYOGRAPHY (EMG) AND MAXIMUM MUSCLE CONTRACTIONS (Visit 2) (~ 30 min)

- Eight EMG electrodes will be attached to your skin on your leg to measure electrical activity in your muscles. Your skin will be prepared through any necessary shaving and cleaning to assure good skin contact.

- A 3-5 minute warm up will take place on a treadmill.

- Maximal voluntary muscle contractions will be recorded for four different ankle motions. This is an "all-out" muscle contraction. You will sit in a muscle strength tester called a dynamometer and your leg will be secured to prevent undesired movement. There will be 3 "all-out" contractions for each of the four motions, each lasting 3 seconds.

RUNNING DATA COLLECTION (Visit 2) (~40 min)

- Reflective markers will be placed on your foot and leg to identify bony landmarks. They will be placed directly on your skin and secured using liquid adhesive and double-sided tape.

- A calibration and ten successful running trials over the walkway and force plate will take place. They will be performed at a speed of 4.0 m/ s ± 10% (8.96 mph ± 10%).

- Data will be collected using a 10-camera Motion Analysis system which will collect the position of the reflective markers, but not any images of your person. If you choose not to be recorded during the gait trials, you will not be eligible to complete the study.

FATIGUING RUN (Visit 2) (~30-45 min)

- The fatiguing run will take place on a treadmill at your selected training pace. Your rate of perceived exertion (RPE) will be collected to identify how hard you feel you are working. After 30 minutes of running, the incline on the treadmill will increase 1^{o} every 5 minutes until you are fatigued and feel you cannot run any longer at the set pace. The run will be terminated when you have at least a score of 8 on the RPE scale and/or you can no longer maintain the set pace.

POST-FATIGUE RUNNING DATA COLLECTION (Visit 2) (~20 min)

- The reflective markers previously placed will be assessed to assure proper placement exists. If needed, markers that have fallen off will be replaced. As in the previously explained running data collection, 10 more running trials will be performed at the same speed of the first 10 trials. RPE will be assessed at the end of the final trial.

4. Risks and Minimizing Risks

What risks will I face by participating in this study?
The potential risks other than muscle soreness or tightness for your participation in this research study are minimal.

Physical Risks:

Likely:

- Minor muscle soreness and/or tightness

Less Likely:

- Musculoskeletal injury such as muscle strain
- Allergic reaction to the liquid adhesive used to secure the reflective markers

Highly unlikely:

- Due to the physical activity demand of the study, there is an increased risk of cardiovascular injury. However, based on your completed medical questionnaire, age, and activity level, risks for this type of injury have been greatly reduced.

Protection of Physical Risks:

To reduce the above risks, appropriate warm-up has been incorporated before the running gait trials. If you feel any soreness or irritation while participating in this study, please tell the investigators as soon as possible. If you are injured, experience allergic reaction to the liquid adhesive used to secure the reflective markers, or experience shortness of breath while participating in this research study, initial first aid and/or appropriate emergency measures will be provided/initiated by the Principal investigator, who is a Licensed Athletic Trainer. If you are a UWM student you will be referred to the Norris Health Center for follow-up care. Non-students will be referred to their primary care physician and will be responsible for all expenses incurred.

Risks to Privacy and Confidentiality:
Less Likely:

- Since a photograph and video will be taken of your foot, this might increase risks to your privacy.
- Since your private information will be collected for this study, there is always a risk of breach of confidentiality.

Protection of Risks to Privacy and Confidentiality:
All data will be stored in a locked filing cabinet in a locked room. All data will be given a letter and number that is uniquely associated with you. This code will NOT contain any partial identifiers (i.e. last four digits of your SSN) and will be stored in a separate locked office in a locked filing cabinet. No identifiers will be stored with the research data. Only those individuals with an active role in this study will have access to the research data and only the PI and Co-investigators will have access to identifying information. When all participants' have completed active participation in the study and data collection is completed, the code will be destroyed. All appropriate measures to protect your private information will be taken.

5. Benefits

Will I receive any benefit from my participation in this study?

- You will be given information as to your foot strike pattern (as some individuals incorrectly identify their proper technique). This may assist you in altering your foot strike pattern in the future to perhaps change performance or reduce injury risk.

6. Study Costs and Compensation

Will I be charged anything for participating in this study?

- You will not be responsible for any of the costs associated with this research study.

Are subjects paid or given anything for being in the study?

- If following the initial screening you qualify for participation in the study, you will receive $20.00 in gift cards upon successful completion of the final running analysis.

7. Confidentiality

What happens to the information collected?

All information collected about you during the course of this study will be kept confidential to the extent permitted by law. We may decide to present what we find to others, or publish our results in scientific journals or at scientific conferences. Information that identifies you personally will not be released without your written permission. Only the PI and the co-investigators will have access to the information. However, the Institutional Review Board at UW-Milwaukee or appropriate federal agencies like the Office for Human Research Protections may review this study's records.

All data collected will be retained for 3 years following completion of the research study. The confidentiality of your data and information will be safeguarded as outlined in "Risks & Minimizing Risks" section under the "Protection of Risks to Privacy and Confidentiality" header.

8. Alternatives

Are there alternatives to participating in the study?
There are no alternatives to participating in this research study. You may choose not to participate.

9. Voluntary Participation and Withdrawal

What happens if I decide not to be in this study?
Your participation in this study is entirely voluntary. You may choose not to take part in this study. If you decide to take part, you can change your mind later and withdraw from the study. You are free to not answer any questions or withdraw at any time. Your decision will not change any present or future relationships with the University of Wisconsin Milwaukee.

If you withdraw from this study before completing the second testing session, we will destroy all information we collect about you. Your decision not to participate or to withdraw early will not result in penalty or harm, nor will it affect your grade or class standing.

10. Questions

Who do I contact for questions about this study?
For more information about the study or the study procedures or treatments, or to withdraw from the study, contact:

Stephen Cobb, PhD, ATC
Dept. of Kinesiology
PO Box 413
Milwaukee, WI 53201

(414) 229-3369

Who do I contact for questions about my rights or complaints towards my treatment as a research subject?

The Institutional Review Board may ask your name, but all complaints are kept in confidence.

Institutional Review Board
Human Research Protection Program
Department of University Safety and Assurances
University of Wisconsin – Milwaukee
P.O. Box 413
Milwaukee, WI 53201
(414) 229-3173

11. Signatures

Research Subject's Consent to Participate in Research:

To voluntarily agree to take part in this study, you must sign on the line below. If you choose to take part in this study, you may withdraw at any time. You are not giving up any of your legal rights by signing this form. Your signature below indicates that you have read or had read to you this entire consent form, including the risks and benefits, and have had all of your questions answered, and that you are 18 years of age or older.

____
Printed Name of Subject/ Legally Authorized Representative

____   ______________________
Signature of Subject/Legally Authorized Representative   Date

Research Subject's Consent to Audio/Video/Photo Recording:

It is okay to photograph my feet while I am in this study and use my photographed data in the research.

Please initial: ____Yes ____No

Principal Investigator (or Designee)

I have given this research subject information on the study that is accurate and sufficient for the subject to fully understand the nature, risks and benefits of the study.

____   ______________________
Printed Name of Person Obtaining Consent   Study Role

____   ______________________
Signature of Person Obtaining Consent   Date

IRBManager Protocol Form

Instructions: Each Section must be completed unless directed otherwise. Incomplete forms will delay the IRB review process and may be returned to you. Enter your information in the **colored boxes** or place an **“X”** in front of the appropriate response(s). If the section does not apply, write **“N/A.”**

SECTION A: Title

A1. Full Study Title:

Foot joint coupling and EMG patterns in habitual forefoot and rearfoot runners

SECTION B: Study Duration

B1. What is the expected start date? *Data collection, screening, recruitment, enrollment, or consenting activities may not begin until IRB approval has been granted. Format: 07/05/2011*

09/01/2013

B2. What is the expected end date? *Expected end date should take into account data analysis, queries, and paper write-up. Format: 07/05/2014*

08/31/2014

SECTION C: Summary

C1. Write a brief descriptive summary of this study in Layman Terms (non-technical language):

This study will compare habitual runners who run with a rearfoot strike pattern (heel strike) to those who run with a forefoot strike pattern (ball of foot strike). A foot model that divides the foot into 6 segments will be used to evaluate foot motion and how foot motion is coupled with adjacent joints. The neuromuscular function (muscle activity) of the leg muscles will be compared between the two running patterns as well. As biomechanics of motion can change after fatigue, observation will take place before and after a fatiguing run.

C2. Describe the purpose/objective and the significance of the research:

The purpose of this project is to identify what biomechanical and neuromuscular differences exist between the different strike patterns to better understand patterns of musculoskeletal overuse injuries commonly seen in runners.

C3. Cite and relevant literature pertaining to the proposed research:

Bramble, D. M., & Lieberman, D. E. (2004). Endurance running and the evolution of Homo. *Nature, 432*(7015), 345-352.

Cobb, S. C., James, C. R., Hjertstedt, M., & Kruk, J. (2011). A Digital Photographic Measurement Method for Quantifying Foot Posture: Validity, Reliability, and Descriptive Data. *J Athl Train, 46*(1), 20-30.

Cobb, S. C., Tis, L. L., Johnson, J. T., Wang, Y. T., Geil, M. D., & McCarty, F. A. (2009). The effect of low-mobile foot posture on multi-segment medial foot model gait kinematics. *Gait & Posture, 30*(3), 334-339.

Farthing, J. P., & Chilibeck, P. D. (2003). The effects of eccentric and concentric training at different velocities on muscle hypertrophy. *Eur J Appl Physiol, 89*(6), 578-586.

Goss, D. L., & Gross, M. T. (2012). Relationships Among Self-reported Shoe Type, Footstrike Pattern, and Injury Incidence. *US Army Med Dep J*, 25-30.

Guidetti, L., Rivellini, G., & Figura, F. (1996). EMG patterns during running: Intra- and inter-individual variability. *J Electromyogr Kinesiol, 6*(1), 37-48.

Hwang, S. J., Choi, H. S., & Kim, Y. H. (2004). Motion analysis based on a multi-segment foot model in normal walking. *Conf Proc IEEE Eng Med Biol Soc, 7*, 5104-5106.

Lieberman, D. E., Venkadesan, M., Werbel, W. A., Daoud, A. I., D'Andrea, S., Davis, I. S., . . . Pitsiladis, Y. (2010). Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature, 463*(7280), 531-535.

Morey-Klapsing, G., Arampatzis, A., & Bruggemann, G. P. (2005). Joint stabilising response to lateral and medial tilts. *Clin Biomech (Bristol, Avon), 20*(5), 517-525.

Nigg, B. M., Cole, G. K., & Nachbauer, W. (1993). Effects of arch height of the foot on angular motion of the lower extremities in running. *Journal of Biomechanics, 26*(8), 909-916.

Pohl, M. B., & Buckley, J. G. (2008). Changes in foot and shank coupling due to alterations in foot strike pattern during running. *Clin Biomech (Bristol, Avon), 23*(3), 334-341.

Williams, D. S., McClay, I. S., & Manal, K. T. (2000). Lower extremity mechanics in runners with a converted forefoot strike pattern. *J Appl Biomech, 16*(2), 210-218.

SECTION D: Subject Population

Section Notes…

- D1. If this study involves analysis of de-identified data only (i.e., no human subject interaction), IRB submission/review may not be necessary. Visit the Pre-Submission section in the IRB website for more information.

D1. Identify any population(s) that you will be <u>specifically targeting</u> for the study. Check all that apply: (Place an "X" in the column next to the name of the special population.)			
	Not Applicable (e.g., de-identified datasets)		Institutionalized/ Nursing home residents recruited in the nursing home
X	UWM Students of PI or study staff		Diagnosable Psychological Disorder/Psychiatrically impaired
	Non-UWM students to be recruited in their educational setting, i.e. in class or at school		Decisionally/Cognitively Impaired
	UWM Staff or Faculty		Economically/Educationally Disadvantaged
	Pregnant Women/Neonates		Prisoners
	Minors under 18 and ARE NOT wards of the State		Non-English Speaking
	Minors under 18 and ARE wards of the State		Terminally ill
X	Other (Please identify): Healthy adult runners (age 18-49)		

D2. Describe the subject group and enter the total number to be enrolled for each group. For example: teachers-50, students-200, parents-25, parent's children-25, student control-30, student experimental-30, medical charts-500, dataset of 1500, etc. Enter the total number of subjects below. If this is a multi-center study, enter the total number of subjects to be enrolled for UWM only. Total enrollment from all sites should be explained in the Protocol Summary Form.	
Describe subject group:	**Number:**
Forefoot runners	**15 will complete the study, but we anticipate screening approximately 25 runners who believe they are forefoot runners in order to identify 15 that meet our forefoot**

	strike pattern criteria (see "Running screen evaluation" in section F1 Below)
Rearfoot runners	**15 will complete the study, but we anticipate screening approximately 25 who believe they are rearfoot runners in order to identify 15 that meet our rearfoot strike pattern criteria (see "Running screen evaluation" in section F1 below)**
TOTAL # OF SUBJECTS:	**50**
TOTAL # OF SUBJECTS (If UWM is a collaborating site):	**50**

D3. List any major inclusion and exclusion criteria (e.g., age, gender, health status/condition, ethnicity, location, English speaking, etc.) and state the justification for the inclusion and exclusion:

Inclusion criteria for all participants are as follows: age 18-45, run a minimum of 10 miles per week, and shoe size 6-13.

Exclusion criteria for all participants are as follows: major lower extremity surgery, lower extremity injury in the previous 6 weeks, asthma or exercise induced asthma not controlled by medication, cardiovascular disease, and pregnant.
Participants who do not qualify will have their data/information destroyed.

SECTION E: Informed Consent

Section Notes…

- E1. Make sure to attach any recruitment materials for IRB approval.
- E3. The privacy of the participants must be maintained throughout the consent process.

E1. Describe how the subjects will be recruited: (E.g., through flyers, beginning announcement for X class, referrals, random telephone sampling, etc.)

Flyers posted on UWM campus
Announcements in UWM classrooms
Flyers to running and triathlon groups at UWM and in the surrounding community

E2. Describe the forms that will be used for each subject group (e.g., short version, combined parent/child consent form, child assent form, verbal script, information sheet): Copies of all forms should be attached for approval. If requesting to waive documentation (not collecting subject's signature) or to waive consent all together, state so and complete the "Waiver to Obtain-Document-Alter Consent" and attach:

Phone screen Health questionnaire
Consent form

E3. Describe who, where, and when consent will be obtained. When appropriate (for higher risk and complex study activities), a process should be mentioned to assure that participants understand the information. For example, in addition to the signed consent form, describing the study procedures verbally or visually**.**

Verbal consent will be achieved upon describing the study to the participant over the phone when screening inclusion/exclusion criteria. Written consent will be obtained upon arrival for the first visit to the Musculoskeletal Injury Biomechanics Lab. The study will be verbally explained to the participant in addition to the written consent form they will sign. **Any information or data from the participants will be destroyed after 5 years.**

SECTION F: Data Collection and Design

Section Notes…

- F1. Data collection instruments should be attached for IRB review.

F1. In the table below, chronologically describe all study activities.

- In **column A**, give the activity a short name.
- In **column B**, briefly describe activities conducted by the PI (recruitment, audiotaping) and describe in greater detail the activities (surveys, interviews, tasks, etc.) research participants will be engaged in. Address where, how long, and when each activity takes place.
- In **column C,** describe any possible risks (e.g., physical, psychological, social, economic, legal, etc.) the subject may ***reasonably*** encounter. Describe the **safeguards** that will be put into place to minimize possible risks (e.g., interviews are in a private location, data is anonymous,

assigning pseudonyms, where data is stored, coded data, etc.) and what happens if the participant gets hurt or upset (e.g., referred to Norris Health Center, PI will stop the interview and assess, given referral, etc.).		
A. Activity Name:	**B. Activity Description:**	**C. Activity Risks and Safeguards:**
Recruitment	Post fliers on UWM bulletin boards, contact UWM instructors to send e-mailed flier to students, contact UWM and community running and triathlon groups and ask to e-mail fliers to members. Recruitment should take 1-2 months.	No risk to participants
Phone screen interview- Health Questionnaire	Participants will be asked a basic health questionnaire over the phone to assure safety during the study and to assure they qualify based on the inclusion and exclusion criteria before being asked to come to the lab for further screening and testing. **In the phone interview, their age, gender, and inquiries of cardiovascular and pulmonary diseases will be asked to assure they qualify for the study. If the individual does not qualify or chooses not to participate, all information/data collected on the participant will be destroyed. This should take approximately 10 minutes**.	No risk to participants
Informed consent (Visit 1)	The first visit to the Musculoskeletal Injury Biomechanics (MIB) Lab (Enderis 132) will begin with verbal instruction to the participant as to all the requirements, risks, benefits, and study purpose. They will then read and sign the written informed consent which contains the same information. After hearing and reading all the risks and requirements of this study, if an individual chooses not to participate in the study, all information/data collected on the participant will be destroyed. This should take approximately 10 minutes	No risk to participants
Foot posture screen (Visit 1)	**Height and weight will be collected from the participant.** Foot arch structure will be assessed in the MIB lab (UWM, Enderis Hall 132) through partial (10%) and full weight-bearing (90%+) digital images of the foot. Ink marks from a ballpoint pen will be made on foot to identify underlying bony structures. Subjects will place foot on scale 1 m from camera lens. One picture will be taken at each of the two different weight-bearing conditions and both feet will be photographed for a total of 4 pictures. This should take approximately 20 minutes	Ink from pen may irritate skin, alcohol wipes are available to remove ink after photographs are taken. Risk of skin irritation from the pen will be minimized by using a pen with ink that conforms to the ASTM D-4236 nontoxic standard. Since a photograph will be taken of the participant's foot, this may increase risks to privacy; however, since the photograph will not

		be identifiable, this risk is minimized. All data is stored in a locked computer and each subject is assigned a number so the feet will not be associated with a name.
Running evaluation screen (Visit 1)	Due to the fact that some runners who believe they run with a forefoot strike pattern actually run with a rearfoot strike pattern, and vice-versa foot strike pattern will be evaluated by the investigator through assessment of ground reaction force patterns. Participants will be asked to run across a carpeted walkway in the MIB lab with right foot placement on the AMTI force plate (Advanced Medical Technologies, Inc, Waterford, MA). A few trials may be necessary to assure proper foot placement on the force plate without alteration of running form. Immediate results will be viewed through the computer program, Cortex (Motion Analysis Corp, Santa Rosa, CA) to determine if participant is a forefoot or rearfoot striker. This should take approximately 10 minutes.	Running requires increased cardiovascular and muscular activity. There is a slight risk of heart attack or other exercise-induced medical conditions (asthma, etc) and musculoskeletal injury as well. The student and faculty primary investigators are both certified athletic trainers and certified in CPR and AED for the professional. Other investigators are also certified in CPR. An AED is available in close proximity to the MIB lab (approx. 50 m) . These risks will be minimized by using trained runners and prescreening for cardiovascular disease. If an individual completes the running evaluation screen and the research group is already filled, their information/data will be destroyed.
EMG electrode placement (Visit 2)	Eight silver-silver chloride surface electromyography (EMG) electrodes (Telemyo 900, Noraxon, Scottsdale, AZ) will be placed on four leg muscles (2 per muscle) on the right leg. They will be placed on the skin surface 20 mm apart over the gastrocnemius medial portion, gastrocnemius lateral portion, the peroneus longus, and the tibialis anterior. Prior to skin placement, skin will be prepared through removal of hair (via shaving razor) and cleaned with an alcohol wipe. Electrodes will be connected to a battery and data storage pack via wires. The pack will be placed on the low back of the participant via a fanny pack. Wires will be secured with Elasticon tape. (15 minutes)	Shaving the hair off the skin may possibly cause minor cuts. Basic first aid will be administered if necessary to stop bleeding. Electrodes may cause minor irritation to the skin. If this occurs, participants will be instructed to treat with ice for 20 minutes upon returning home.
Leg MVC's (Visit 2)	Participants will undergo a 3-5 minute warm up on a treadmill. Each subject will then complete maximal voluntary isometric contraction (MVC) testing for ankle dorsiflexion, plantarflexion, inversion, and eversion on a Biodex dynamometer (Biodex Medical Systems, Shirley,	There is a slight risk of musculoskeletal injury during the MVC's. Risk will be reduced by allowing for adequate warm-up and familiarity of dynamometer before MVC. If any injury occurs, the test will be stopped and the individual will be

	NY). Subjects will be instructed to perform a 3 sec an “all-out” muscle contraction. During the contraction, strong verbal encouragement will be delivered to motivate a true MVC. Muscle activation intensities during running will be evaluated as a percentage of the ankle MVC’s. (20 min)	instructed to treat it accordingly with ice and any other necessary modality
Running Data Collection (Visit 2)	Reflective marker clusters (6.4 mm diameter markers) will be placed using liquid adhesive (Mastisol) and toupee tape on the right leg and foot to identify the following bony landmarks: medial and lateral malleoli, calcaneus, navicular, cuboid, metatarsals 1, 2, 4, and 5, and the hallux. Prior to running analysis, a seated and standing anatomical calibration will take place to develop local coordinate systems within each segment. During the standing calibration, additional markers will be placed on the greater trochanters and medial and lateral joint line of the knee and then removed after calibration to identify knee and hip joint centers. Marker clusters will also be positioned on the thigh segment and leg segment prior to the standing calibration and left on during running trials. Participants will then execute the running analysis portion by completing 10 successful running trials (4.0 m·s^{-1} ± 10%) in a minimal sandal using their typical foot strike pattern. The running trials will occur on a 30 m carpeted walkway with an AMTI force plate (Advanced Mechanical Technology, Inc, Watertown, MA) located at about 20 m down the walkway. Motion analysis data will be collected at 200 Hz with a 10-camera Eagle Motion Analysis System (Motion Analysis Corp, Santa Rosa, CA) and the AMTI force plate sampling at 1000 Hz will be used to assess ground reaction forces. (40 minutes)	Running requires increased cardiovascular and muscular activity. There is a risk of heart attack or other exercise-induced medical conditions (asthma, etc) and musculoskeletal injury as well. The student and faculty primary investigators are both certified athletic trainers and certified in CPR and AED for the professional. Other investigators are also certified in CPR. An AED is available in close proximity to the MIB lab (approx. 50 m). The age of participation is also maximized at 45 to avoid the increase in risk for cardiovascular problems in individuals over 45. These risks will also be minimized by using trained runners and prescreening for cardiovascular disease. There may be a slight risk of allergic reaction to the liquid adhesive used to secure the reflective markers. If an allergic reaction is experienced, UWM student participants will be referred to the Norris Health Center for follow-up care. Non-students will be referred to their primary care physician and will be responsible for all expenses incurred. If an unexpected emergency event occurs, a full report will be submitted to the IRB.the participant will be referred to Norris Health Center for immediate care. Participants may experience minor muscle soreness and/or musculoskeletal injury such as muscle strain as a result of the running gait trials. Proper warm-up will help to minimize this risk. Participants will also be asked to report any

		shortness of breath during the trials. In the case of injury or illness, initial first aid and/or appropriate emergency measures will be provided/initiated by the Principal investigator, who is a Licensed Athletic Trainer. UWM student participants who get hurt, upset, or experience shortness of breath during the trials will be referred to the Norris Health Center for follow-up care. Non-students will be referred to their primary care physician and will be responsible for all expenses incurred. If this event is unexpected, a full report will be submitted to the IRB.
Fatiguing run (Visit 2)	After 10 successful running trials, subjects will be placed on a treadmill to perform a fatiguing run. Subjects will be allowed 5 minutes for warm-up and then choose a speed that is aligned with their training pace. Level of fatigue will be assessed during the run by assessing rate of perceived exertion (RPE) on a scale of 1-10 every 5 minutes. After 30 minutes of running, intensity will increase by increasing treadmill incline 1° every 5 minutes. The fatiguing run will be terminated when the participant is reporting no less than an 8 on the RPE scale and/or can no longer continue at the set pace.	Running requires increased cardiovascular and muscular activity. There is a slight risk of heart attack or other exercise-induced medical conditions (asthma, etc) and musculoskeletal injury as well. The student and faculty primary investigators are both certified athletic trainers and certified in CPR and AED for the professional. Other investigators are also certified in CPR. An AED is available in close proximity to the MIB lab (approx. 50 m). The age of participation is also maximized at 45 to avoid the increase in risk for cardiovascular problems in individuals over 45. . These risks will also be minimized by using trained runners and prescreening for cardiovascular disease.
Post-fatigue running data collection (Visit 2)	Foot and leg reflective markers will be assessed after the fatiguing run to assure that none fell off and need to be replaced. If replacement of markers is required, seated and standing calibration will be done again but this time after the gait trials in order to not give time to recover to the participant. As in the pre-fatiguing gait trials, 10 successful trials will be conducted at the same speed of the first 10 trials. RPE will be assessed again after	Running requires increased cardiovascular and muscular activity. There is a slight risk of heart attack or other exercise-induced medical conditions (asthma, etc) and musculoskeletal injury as well. The student and faculty primary investigators are both certified athletic trainers and certified in CPR and AED for the professional. Other investigators are also certified in CPR. An AED is available in close proximity to the MIB lab (approx. 50 m). The age of participation is also

	completion of the final successful trial to see if the participant feels recovered from the fatigue. (20 minutes)	maximized at 45 to avoid the increase in risk for cardiovascular problems in individuals over 45. . These risks will also be minimized by using trained runners and prescreening for cardiovascular disease.
Data Analysis	Data analyses will be performed using Matlab (Matlab R2011b, MathWorks, Natick, MA) to filter data, calculate range of motion and joint coupling variables and SPSS to identify statistical significance.	No risk to participants

F2. Explain how the data will be analyzed or studied (i.e. quantitatively or qualitatively) and how the data will be reported (i.e. aggregated, anonymously, pseudonyms for participants, etc.):

Data will be evaluated quantitatively and all participants will be given a random code to which they will be identified.

All data will be stored in a locked filing cabinet in a locked room. All data will be given a letter and number that is uniquely associated with each individual participant. This code will not contain any partial identifiers and will be stored in a separate locked office in a locked filing cabinet. No identifiers will be stored with the data. **Those having access include the PI, Stephen Cobb, the student PI, Rhiannon Seneli, and Emily Gerstle, another graduate student in the lab.** When all participants' have completed active participation in the study and data collection is completed, the **key linking the names to the code** will be destroyed **one year after data collection is complete**. **All data will be retained for 3 years following conclusion of the research study.**

SECTION G: Benefits and Risk/Benefit Analysis

Section Notes…

- Do not include Incentives/ Compensations in this section.

G1. Describe any benefits to the individual participants. If there are no anticipated benefits to the subject directly, state so. Describe potential benefits to society (i.e., further knowledge to the area of study) or a specific group of individuals (i.e., teachers, foster children). Describe the ratio of risks to benefits.

Subjects will be given **immediate** biomechanical information as to their footstrike pattern **based on the initial footstrike analysis** and **overall group results of** pre- and post- fatiguing run biomechanical results of the footstrike pattern groups will be available upon request from the participant after data analyses have been completed **for up to one year, at which time the code linking participant to data will be destroyed**. Further understanding of foot motion between the two foot patterns will help to identify possible prevention of injury and rehabilitation strategies for leg musculoskeletal injuries.

G2. Risks to research participants should be justified by the anticipated benefits to the participants or society. Provide your assessment of how the anticipated risks to participants and steps taken to minimize these risks, balance against anticipated benefits to the individual or to society.

The risks for this study are minimal and the knowledge of the participants' footstrike pattern may help to reduce future injuries through technique changes. Additionally, the benefits to the running community are great in helping to identify possible biomechanical causes to running injuries.

SECTION H: Subject Incentives/ Compensations

Section Notes…

- H2 & H3. The IRB recognizes the potential for undue influence and coercion when extra credit is offered. The UWM IRB, as also recommended by OHRP and APA Code of Ethics, agrees when extra credit is offered or required, prospective subjects should be given the choice of an equitable alternative. In instances where the researcher does not know whether extra credit will be accepted and its worth, such information should be conveyed to the subject in the recruitment materials and the consent form. For example, "The awarding of extra credit and its amount is dependent upon your instructor. Please contact your instructor before participating if you have any questions. If extra credit is awarded and you choose to not participate, the instructor will offer an equitable alternative."
- H4. If you intend to submit to the Travel Management Office for reimbursement purposes make sure you understand what each level of payment confidentiality means **(click here for additional information).**

H1. Does this study involve incentives or compensation to the subjects? For example cash, class extra credit, gift cards, or items.

[X] **Yes**
[__] **No [SKIP THIS SECTION]**

H2. Explain what (a) the item is, (b) the amount or approximate value of the item, and (c) when it will be given. For extra credit, state the number of credit hours and/or points. (e.g., $5 after completing each survey, subject will receive [item] even if they do not complete the procedure, extra credit will be award at the end of the semester)**:**

$20 in gift certificates to area merchants

H3. If extra credit is offered as compensation/incentive, an alternative activity (which can be another research study or class assignment) should be offered. The alternative activity should be similar in the amount of time involved to complete and worth the same extra credit. If the task is a class requirement/assignment that students would be required to complete.

H4. If cash or gift cards, select the appropriate confidentiality level for payments (see section notes):

[X_] **Level 1** indicates that confidentiality of the subjects is not a serious issue, e.g., providing a social security number or other identifying information for payment would not pose a serious risk to subjects.

- Choosing a Level 1 requires the researcher to maintain a record of the following: The payee's name, address, and social security number and the amount paid.
- When Level 1 is selected, a formal notice is not issued by the IRB and the Travel Management Office assumes Level 1.
- Level 1 payment information will be retained in the extramural account folder at UWM/Research Services and attached to the voucher in Accounts Payable. These are public documents, potentially open to public review.

[__] **Level 2** indicates that confidentiality is an issue, but is not paramount to the study, e.g., the participant will be involved in a study researching sensitive, yet not illegal issues.

- Choosing a Level 2 requires the researcher to maintain a record of the following: A list of names, social security numbers, home addresses and amounts paid.
- When Level 2 is selected, a formal notice will be issued by the IRB.
- Level 2 payment information, including the names, are attached to the PIR and become part of the voucher in Accounts Payable. The records retained by Accounts Payable are not considered public record.

[__] **Level 3** indicates that confidentiality of the subjects must be guaranteed. In this category, identifying information such as a social security number would put a subject at increased risk.

- Choosing a Level 3 requires the researcher to maintain a record of the following: research subject's name and corresponding coded identification. This will be the only record of payee names, and it will stay in the control of the PI.
- Payments are made to the research subjects by either personal check or cash.
- Gift cards are considered cash.
- If a cash payment is made, the PI must obtain signed receipts.

SECTION I: Deception/ Incomplete Disclosure (INSERT “NA” IF NOT APPLICABLE)
Section Notes… • If you cannot adequately state the true purpose of the study to the subject in the informed consent, deception/ incomplete disclosure is involved.

I1. Describe (a) what information will be withheld from the subject (b) why such deception/ incomplete disclosure is necessary, and (c) when the subjects will be debriefed about the deception/ incomplete disclosure.

Appendix J: IRB Approval

UNIVERSITY of WISCONSIN
UWMILWAUKEE
Department of University Safety & Assurances

Jessica Rice
IRB Administrator
Institutional Review Board
Engelmann 270
P. O. Box 413
Milwaukee, WI 53201-0413
(414) 229-3182 *phone*
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http://www.irb.uwm.edu
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New Study - Notice of IRB Expedited Approval

Date: August 28, 2013

To: Stephen Cobb, PhD
Dept: Kinesiology

Cc: Rhiannon Senali

IRB#: 14.030
Title: Foot joint coupling and EMG patterns in habitual forefoot and rearfoot runners

After review of your research protocol by the University of Wisconsin – Milwaukee Institutional Review Board, your protocol has been approved as minimal risk Expedited under Categories 4, 6 & 7 as governed by 45 CFR 46.110.

This protocol has been approved on **August 28, 2013** for one year. IRB approval will expire on **August 27, 2014**. If you plan to continue any research related activities (e.g., enrollment of subjects, study interventions, data analysis, etc.) past the date of IRB expiration, a continuation for IRB approval must be filed by the submission deadline. If the study is closed or completed before the IRB expiration date, please notify the IRB by completing and submitting the Continuing Review form found on the IRB website.

Unless specifically where the change is necessary to eliminate apparent immediate hazards to the subjects, any proposed changes to the protocol must be reviewed by the IRB before implementation. It is the principal investigator's responsibility to adhere to the policies and guidelines set forth by the UWM IRB and maintain proper documentation of its records and promptly report to the IRB any adverse events which require reporting.

It is the principal investigator's responsibility to adhere to UWM and UW System Policies, and any applicable state and federal laws governing activities the principal investigator may seek to employ (e.g., FERPA, Radiation Safety, UWM Data Security, UW System policy on Prizes, Awards and Gifts, state gambling laws, etc.) which are independent of IRB review/approval.

Contact the IRB office if you have any further questions. Thank you for your cooperation and best wishes for a successful project

Respectfully,

Jessica P. Rice
IRB Administrator

Table K.1. Statistical findings from MANOVA testing for kinematics of the multi-segment foot model (RC = rearfoot complex, CNC = calcaneonavicular, MFF = medial forefoot, MTP = 1st metatarsophalangeal, LFF = lateral forefoot, and CC = calcaneocuboid) at initial contact (IC) and the maximum and minimum joint angle over the stance phase. Within and between-subjects factors were time (pre- and post- 5 km run) and footstrike (rearfoot vs forefoot strike). Results are reported as F-statistic (top number) and *p*-value (bottom number). Follow-up ANOVAs were used to identify differences seen in the MANOVAs.

Joint	**MANOVA (ANOVA) dF**	**Plane**	Within-subjects main effects (Time)				Between-subjects main effects (Foot strike)				Interaction effects			
			MANOVA	ANOVA IC	ANOVA max	ANOVA min	MANOVA	ANOVA IC	ANOVA max	ANOVA min	MANOVA	ANOVA IC	ANOVA Max	ANOVA min
RC	3,24 (1,26)	Sag	7.96 .001	2.23 .147	2.83 .102	2.80 .107	10.29 < .001	27.77 < .001	1.57 .222	1.08 .309	1.50 .24			
		Fron	11.30 < .001	3.02 .09	3.96 .06	20.81 < .001	3.91 .02	8.13 .01	10.15 .004	.333 .57	.12 .95			
		Tran	5.19 .007	4.84 .04	5.35 .03	.06 .81	.92 .45				2.60 .08			
CNC	3,25 (1,27)	Sag	6.94 .001	10.85 .003	9.10 .006	1.28 .27	.55 .66				.38 .77			
		Fron	3.16 .04	.66 .42	.21 .65	5.31 .03	2.90 .06				.30 .82			
		Tran	1.08 .38				2.62 .07				2.27 .11			
MFF	3,24	Sag	16.50 < .001				20.27 < .001				3.78 .02	9.49 .005	.177 .677	.037 .849
MTP	3,25 (1,27)	Sag	15.04 < .001	2.25 .15	12.17 .002	2.71 .11	.74 .54				.23 .88			
LFF	3,25	Sag	3.55 .03				2.07 .13				4.12 .02	2.88 .10		
CC	3,25 (1,28)	Sag	5.80 .004	13.54 .001	6.79 .02	3.72 .06	17.15 < .001	10.86 .003	.49 .49	2.01 .17	.64 .59			
		Fron	1.38 .27				.50 .69				.49 .70			
		Tran	3.77 .02	1.98 .17	10.82 .003	.26 .61	1.07 .38				1.84 .17			

Table K.2. Statistical findings from subphase 1 angular displacement. Separate MANOVAs were used to compare the three planes of motion for the rearfoot complex (RC), calcaneonavicular complex (CNC), and the calcaneocuboid complex (CC). Follow-up ANOVAs were used to identify any significant differences. For the medial forefoot (MFF), 1st metatarsophalangeal joint (MTP), and lateral forefoot (LFF) only the sagittal plane was examined and mixed within- and between-subjects ANOVAs were used. All results are reported as F-statistic (top number) and *p*-value (bottom number).

Joint	**MANOVA (ANOVA) dF**	Within-subjects main effects (Time)				Between-subjects main effects (Foot strike)				Interaction effects	
		MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA
RC	3,24 (1,26)	7.49 .001	.042 .84	12.42 .002	8.08 .009	2.94 .054				.381 .768	
CNC	3,25 (1,27)	5.04 .007	2.35 .14	5.69 .02	13.70 .001	8.87 < .001	20.39 < .001	4.38 .046	.114 .74	.719 .55	
MFF	(1,26)		3.72 .07				5.39 .03				.26 .62
MTP	(1,27)		2.96 .10				.93 .60				.29 .34
LFF	(1,27)		.28 .62				4.06 .05				1.98 .17
CC	3,25 (1,28)	1.39 .269				3.78 .023	6.74 .02	2.57 .12	.022 .88	.629 .603	

Table K.3. Statistical findings from subphase 2 angular displacement. Separate MANOVAs were used to compare the three planes of motion for the rearfoot complex (RC), calcaneonavicular complex (CNC), and the calcaneocuboid complex (CC). Follow-up ANOVAs were used to identify any significant differences. For the medial forefoot (MFF), 1st metatarsophalangeal joint (MTP), and lateral forefoot (LFF) only the sagittal plane was examined and mixed within- and between-subjects ANOVAs were used. All results are reported as F-statistic (top number) and *p*-value (bottom number).

Joint	MANOVA (ANOVA) dF	Within-subjects main effects (Time)				Between-subjects main effects (Footstrike)				Interaction effects			
		MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.
RC	3,24 (1,26)	4.94 .008				2.68 .070				6.12 .003	6.81 .015	.001 .973	11.99 .002
CNC	3,25 (1,27)	2.13 .122				3.89 .021				3.70 .025	.364 .55	.123 .73	9.03 .006
MFF	(1,26)		17.42 < .001				27.01 < .001				4.61 .04		
MTP	(1,27)		3.84 .06				.815 .38				.75 .39		
LFF	(1,27)		1.02 .32				2.23 .15				9.73 .004		
CC	3,25 (1,28)	.931 .440				11.26 < .001	30.68 < .001	1.16 .29	4.89 .04	2.11 .40			

Table K.4. Statistical findings from subphase 3 angular displacement. Separate MANOVAs were used to compare the three planes of motion for the rearfoot complex (RC), calcaneonavicular complex (CNC), and the calcaneocuboid complex (CC). Follow-up ANOVAs were used to identify any significant differences. For the medial forefoot (MFF), 1st metatarsophalangeal joint (MTP), and lateral forefoot (LFF) only the sagittal plane was examined and mixed within- and between-subjects ANOVAs were used. All results are reported as F-statistic (top number) and *p*-value (bottom number).

Joint	MANOVA (ANOVA) dF	Within-subjects main effects (Time)				Between-subjects main effects (Footstrike)				Interaction effects	
		MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal
RC	3,24 (1,26)	3.49 .031	6.57 .02	5.44 .03	.182 .67	1.02 .403				.023 .995	
CNC	3,25 (1,27)	.038 .990				.198 .897				.186 .905	
MFF	(1,26)		4.02 .06				3.29 .08				1.70 .20
MTP	(1,27)		25.93 < .001				6.46 .02				.12 .73
LFF	(1,27)		7.86 .009				.01 .92				.27 .61
CC	3,25 (1,28)	.623 .607				1.51 .235				.846 .481	

Table K.5. Statistical findings from subphase 4 angular displacement. Separate MANOVAs were used to compare the three planes of motion for the rearfoot complex (RC), calcaneonavicular complex (CNC), and the calcaneocuboid complex (CC). Follow-up ANOVAs were used to identify any significant differences. For the medial forefoot (MFF), 1st metatarsophalangeal joint (MTP), and lateral forefoot (LFF) only the sagittal plane was examined and mixed within- and between-subjects ANOVAs were used. All results are reported as F-statistic (top number) and *p*-value (bottom number).

Joint	**MANOVA (ANOVA) dF**	Within-subjects main effects (Time)				Between-subjects main effects (Footstrike)				Interaction effects			
		MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.	MANOVA	ANOVA Sagittal	ANOVA Frontal	ANOVA Trans.
RC	3,24 (1,26)	7.18 .001	.002	.006	.006	.246 .863				.230 .875			
CNC	3,25 (1,27)	8.93 < .001	19.59 < .001	.001 .973	.112 .74	.914 .449				.726 .546			
MFF	(1,26)		22.52 < .001				0.20 .89				.038 .85		
MTP	(1,27)		12.95 .001				.635 .43				.001 .98		
LFF	(1,27)		1.24 .28				.392 .54				1.25 .27		
CC	3,25 (1,28)	1.01 .405				2.64 .071				3.24 .038	4.25 .049	.365 .55	4.90 .04

Appendix L: Abstract presented at American Society of Biomechanics Annual Conference

IDENTIFYING TOE-OFF EVENT RUNNING ON A TREADMILL USING KINEMATIC DATA

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INTRODUCTION

Many running analysis studies are conducted on a runway with a force plate to identify stance phase using the ground reaction force. Running on a treadmill offers the opportunity to evaluate subsequent steps and evaluate running biomechanics at different time points of an uninterrupted run. Force plate instrumented treadmills have been on the market for some time, but are often expensive and many motion analysis laboratories do not possess such equipment. Therefore, some researchers have tested various methods for establishing the stance phase of running on a treadmill using kinematic data [1-3]. Previously proposed methods for identifying toe-off may require additional markers and joint measurements than what is used when tracking foot and leg motion or may have inaccurate estimations when compared to ground reaction force data. Therefore, the purpose of this research was to establish an accurate kinematic method for identifying toe-off while running on a treadmill using a multi-segment foot model.

METHODS

Twelve runners (≥ 10 miles/week, 26.3 ± 4.85 years) volunteered for the study. Six participants conducted overground running on a 25 m platform with a force plate (1000 Hz) and six ran on a treadmill. Each participant was equipped with 6.4 mm retroreflective markers identifying the leg and six-segments of the foot: rearfoot complex, calcaneonavicular, medial forefoot, first metatarsophalangeal (MTP), lateral forefoot, and calcaneocuboid. Three-dimensional positions of markers were captured at 200 Hz with a 10-camera Eagle system.

For the overground methods, five successful trials of running at 4.0 (±10%) m/s were evaluated for finding toe-off. The gold standard measurement for toe-off was the timing of when the vertical ground reaction force was less than 10 N after initial contact. For treadmill methods, one corner of the treadmill was positioned on the force plate. All subjects ran on a treadmill at 3.3 m/s and 10 s of running was captured. Five consecutive steps were used to analyze toe-off timing. Because the treadmill was not instrumented with a force plate, the gold standard measurement for comparison used a ground reaction force baseline for each subject as the threshold level. This baseline varied for each subject as the treadmill was moved between participants and was not always positioned identically.

Gold standard measurements were then compared to three different kinematic methods for determining toe-off. First, toe-off was defined as the maximum sagittal plane angle of the MTP joint after initial contact (SagM). Second method used the vertical position of the toe tip and identified its minimum to be toe-off (VTOE) [3]. The third method used the local maximum of the vertical acceleration of the toe tip after initial contact and then linearly interpolated the jerk of the toe tip to find where jerk was equal to zero (ATOE) [3]. This interpolation allowed to identify exact timing between frames. All three methods were then compared to ground reaction force data for each trial and the difference in frames was recorded. The differences were then averaged for each subjects' five trials and then mean and standard deviations were recorded for the group (overground and treadmill). Positive frame differences indicated that the kinematic method identified an earlier toe-off while negative numbers identified a later toe-off when compared to ground reaction force. Root mean square differences (RMS) were then calculated to assess error.

RESULTS AND DISCUSSION

For both overground and treadmill running, the SagM method was the most accurate with a frame error that corresponded to 0.007 s and -0.006 s respectively (Table 1). Small RMS values also indicate that this level of accuracy was seen for all subjects in each group (Table 2). The use of VTOE was moderately accurate for overground methods (-0.04 s), but not for treadmill running (0.22 s) (Table 1) and had corresponding RMS results (Table 2). The ATOE also had moderately accurate results for both the overground (-0.03 s) and treadmill running (-0.07 s) (Table 1) with similar RMS results as well (Table 2).

The maximal MTP dorsiflexion occurs as the runner is pushing off and gives an accurate estimate as to when toe-off occurs while running overground and on a treadmill. One disadvantage to this method is that it requires a model in which the MTP is identified with markers and many running studies treat the foot as a single, rigid segment. The other methods, VTOE and ATOE, do not require this, but rather just one marker at the toe tip. However, they do not appear to be as accurate and VTOE does not seem adequate for estimating toe-off with treadmill running. The results found from these two methods are similar to results found by Zeni, et al. [3] which had a difference in contact time of approximately 0.02-0.05 s when using the ATOE toe-off prediction and .16 s when using the VTOE method with treadmill running.

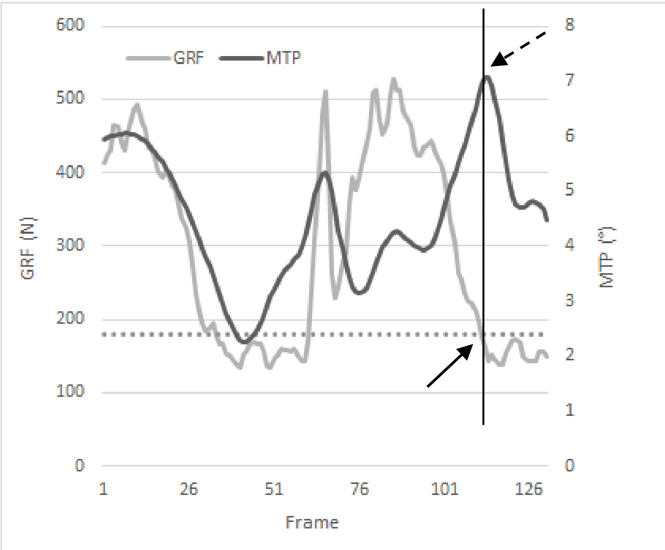

Figure 1: Ground reaction force (GRF) data and first metatarsophalangeal (MTP) sagittal plane angle while running on a treadmill. Dotted line indicates threshold for determining toe-off using GRF (solid arrow). Maximum MTP angle used to determine toe-off with kinematic methods (dashed arrow).

CONCLUSIONS

Various kinematic methods for determining toe-off during running gait have been researched and found to have varying levels of accuracy. Using maximal dorsiflexion angle of the first metatarsophalangeal joint may provide a very accurate option for identifying toe-off when running overground and on a treadmill, but would require the use of a multi-segment foot model.

REFERENCES

1. De Witt JK. *J Biomech* **43**, 3067-3069, 2010.
2. Maiwald C, et al. *Footwear Science* **1**, 111-118, 2009.
3. Zeni JA, et al. *Gait Posture* **27**, 710-714, 2008.

Appendix M: Running Log

Participant _________________________

Running Log

Date	Sandal Goal	Sandal distance	Total distance	Comfort level (5= very comfortable, 1=painful)	Notes
	1 mile			5 4 3 2 1	
	1 mile			5 4 3 2 1	
	1 ½ mile			5 4 3 2 1	
	2 mile			5 4 3 2 1	
	2 ½ mile			5 4 3 2 1	
	3 mile			5 4 3 2 1	
	3 ½ mile			5 4 3 2 1	
	4 mile			5 4 3 2 1	
	4 ½ mile			5 4 3 2 1	
	5 mile			5 4 3 2 1	

Begin the adjustment into your new sandals by completing 1 mile of your normal running distance in the sandals for your next two runs. Then, increase the distance in the new sandals by ½ mile every time you run. If the sandals cause irritation or pain, DO NOT increase amount from the previous run until irritation or pain subsides. If you are not able to build up to at least a 3 mile distance in the sandals within 10 days, please report it to the study coordinator, Rhiannon Seneli (lange3@uwm.edu) prior to your next appointment.

Appendix N: Participant Running Information Form

Please complete the following questions to the best of your knowledge.

1. On average, how many miles do you run weekly? _______________
2. How long have you been running over 10 miles a week? _________________
3. How many years of running experience total do you have? ________________
4. Have you run a 5K in the last year? ☐ Yes ☐ No

   If yes, what was your most recent 5K time? ___________________

5. What other length races have you recently run and what was the completion time?

   ☐ 10 km _________________

   ☐ ½ Marathon ________________

   ☐ Marathon __________________

   ☐ Other(specify) __________________ _______________________

6. What is your current foot strike pattern when running (what do you hit the ground with first)?
   ☐ Forefoot strike (toe) ☐ Midfoot strike (flat) ☐ Rearfoot strike (heel)

7. How many years have you run in your current strike pattern? ______________

8. What type of footwear do you regularly run in?
   ☐ Traditional running shoe ☐ Minimalist shoe ☐ Motion control shoe/Stability
   ☐ Barefoot ☐ Neutral shoe

9. How long have you been in those types of shoes? _______________________

Appendix O: Extended Joint Coupling Variability Statistical Results

Table O. Statistical results from mixed between-within subjects ANOVA for joint coupling variability in the foot and ankle. Joint couples includes functional articulations rearfoot complex (RC), calcaneonavicular (CNC), calcaneocuboid (CC), medial forefoot (MFF), lateral forefoot (LFF), and 1st metatarsophalangeal (MTP) completed in the sagittal (sag), frontal (fron), and transverse (tran) planes.

			Within-subjects		**Between-subjects**		**Interaction**	
Joint Couple	**dF**	**Phase**	**F-stat**	***P*-value**	**F-stat**	***P*-value**	**F-stat**	***P*-value**
RCsag-CNCsag	1,28	1	0.030	0.863	0.049	0.826	0.010	0.921
		2	0.492	0.489	37.375	< 0.001	1.271	0.269
		3	0.088	0.769	1.261	0.271	1.791	0.192
		4	0.600	0.445	2.878	0.101	0.127	0.724
RCfron-CNCfron	1,28	1	1.760	0.195	1.803	0.190	0.367	0.549
		2	2.717	0.110	1.100	0.303	0.078	0.782
		3	0.478	0.495	2.652	0.115	0.977	0.331
		4	0.180	0.675	0.280	0.601	0.001	0.981
RCtran-CNCtran	1,28	1	0.577	0.454	2.312	0.140	0.346	0.561
		2	14.764	0.001	0.288	0.595	1.637	0.211
		3	2.634	0.116	0.049	0.826	0.079	0.780
		4	0.003	0.954	0.115	0.737	1.254	0.272
CNCfron-MFFsag	1,27	1	0.020	0.888	4.219	0.050	0.064	0.803
		2	5.115	0.032	45.677	< 0.001	5.830	0.023
		3	0.012	0.913	0.719	0.404	0.998	0.327
		4	0.721	0.403	0.001	0.979	0.706	0.408
MFFsag-MTPsag	1,27	1	0.166	0.687	28.206	< 0.001	0.044	0.836
		2	1.435	0.241	1.758	0.196	3.257	0.082
		3	0.987	0.329	3.754	0.063	1.657	0.209
		4	0.587	0.450	0.001	0.978	5.262	0.030
RCsag-CCsag	1,28	1	0.081	0.778	4.159	0.051	1.008	0.324
		2	0.890	0.354	42.847	< 0.001	0.178	0.676
		3	0.158	0.694	0.006	0.937	0.014	0.908
		4	0.063	0.804	0.842	0.367	0.886	0.355
RCfron-CCfron	1,28	1	2.813	0.105	0.593	0.448	1.526	0.227
		2	11.475	0.002	5.557	0.026	0.357	0.555
		3	0.109	0.743	1.006	0.324	1.548	0.224
		4	0.130	0.722	1.425	0.243	0.476	0.496
RCtran-CCtran	1,28	1	0.011	0.918	3.246	0.082	0.407	0.529
		2	9.876	0.004	0.963	0.335	3.176	0.086
		3	0.268	0.609	0.407	0.529	0.073	0.789
		4	1.099	0.303	0.003	0.957	0.495	0.488
CCfron-LFFsag	1,27	1	0.110	0.742	4.287	0.048	0.125	0.726
		2	8.329	0.008	0.175	0.679	2.048	0.164
		3	0.461	0.503	4.596	0.041	1.641	0.211
		4	0.407	0.529	2.136	0.155	0.140	0.711
RCtran-RCfron	1,28	1	1.185	0.286	3.971	0.056	0.029	0.866
		2	9.589	0.004	0.014	0.907	2.610	0.117
		3	0.002	0.969	0.093	0.762	0.018	0.893
		4	0.187	0.669	0.161	0.691	0.936	0.342

Appendix P: Kinematic Time Series Graphs

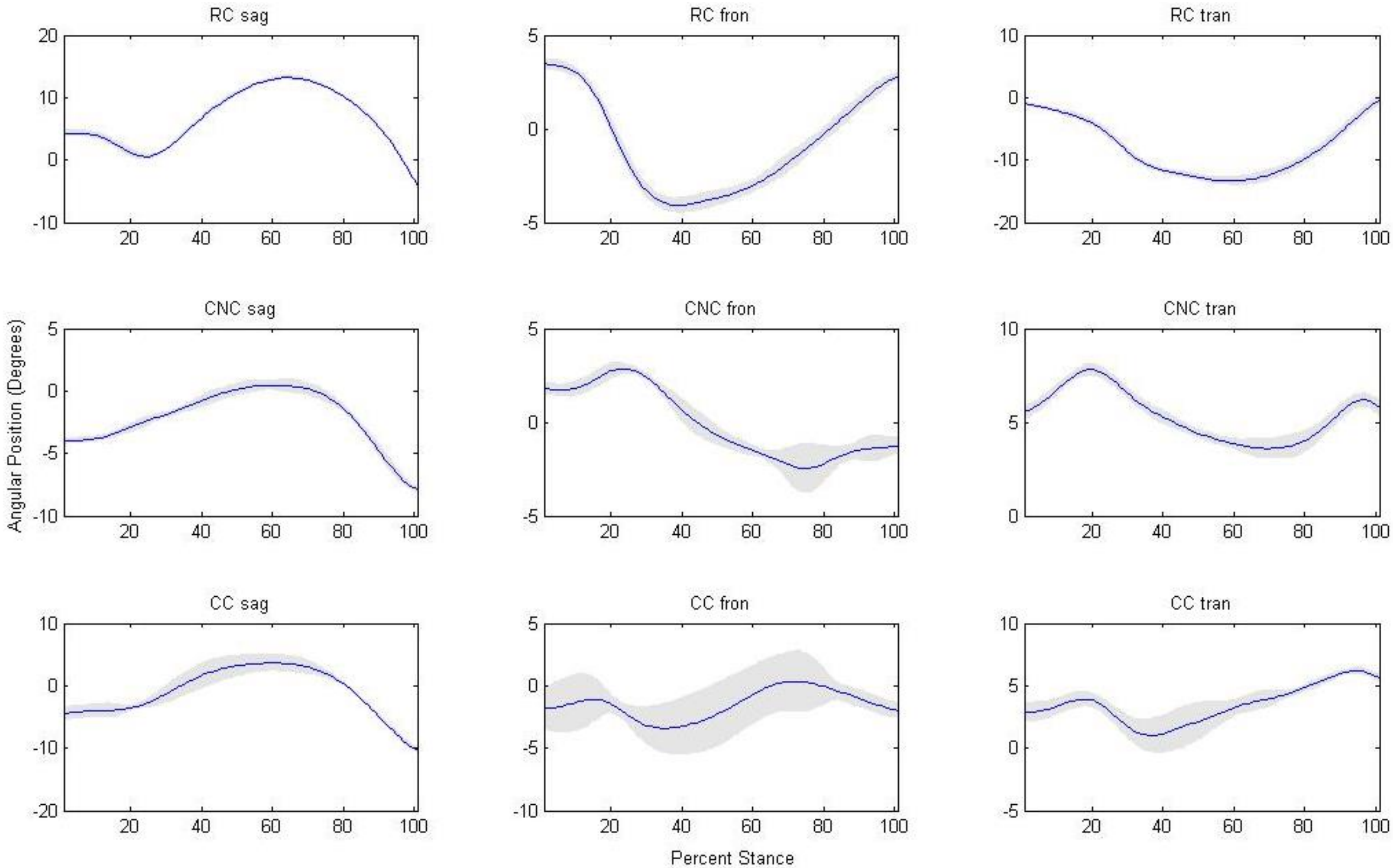

Figure P.1- Time series for joints in RFS runners at the beginning of the 5 km run. Shaded area indicates standard deviation.

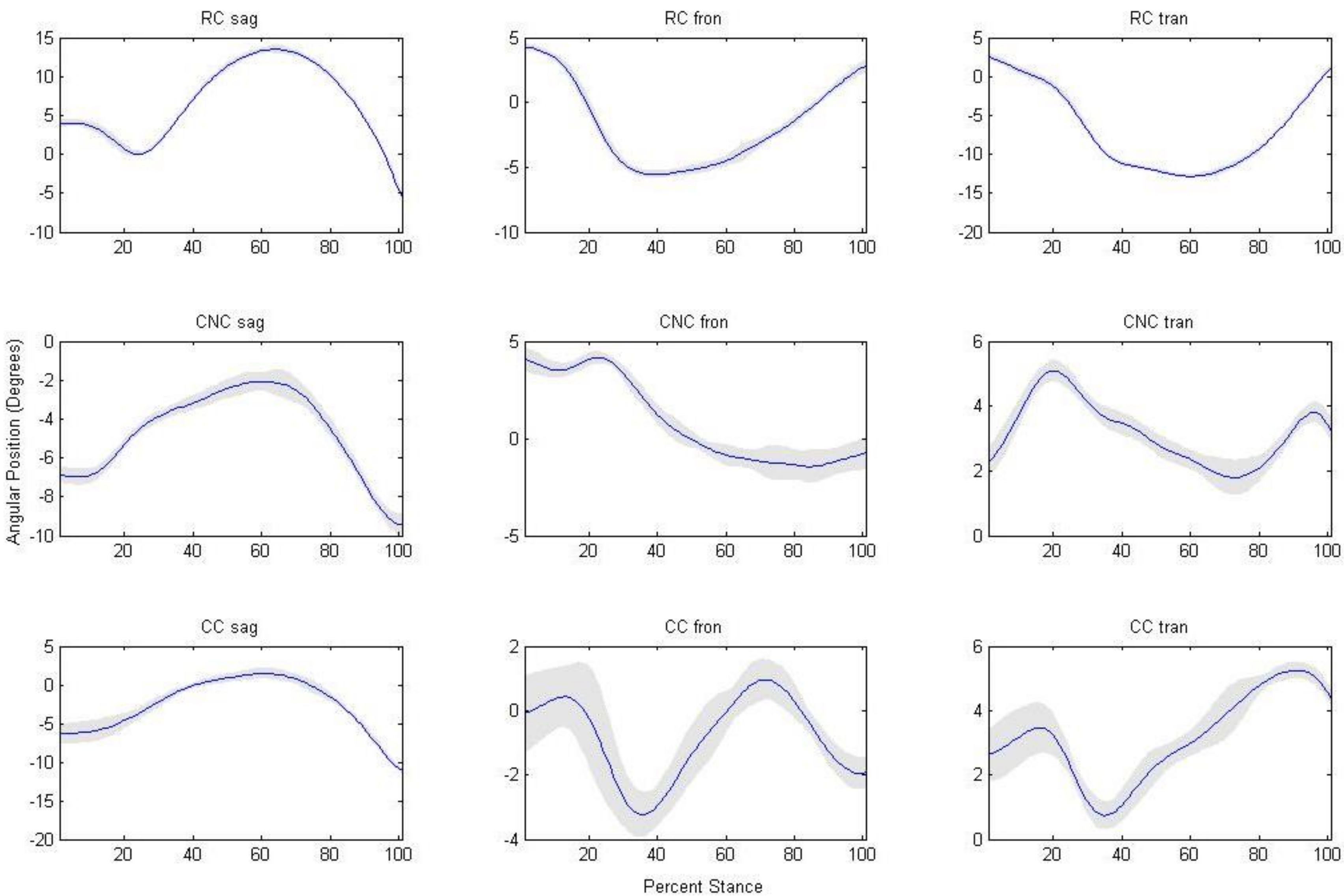

Figure P.2- Time series for joints in RFS runners at the end of the 5 km run. Shaded area indicates standard deviation.

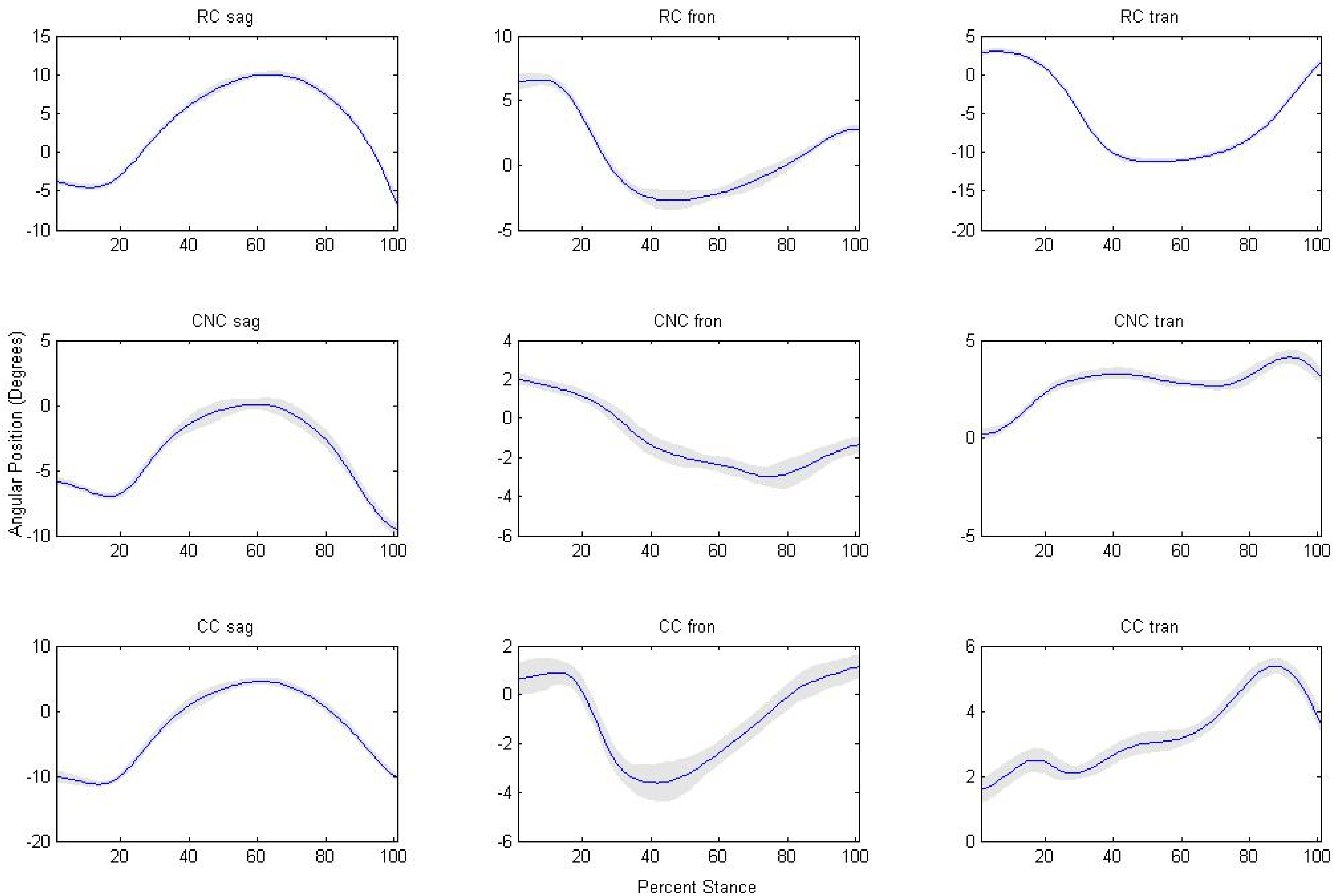

Figure P.3- Time series for joints in FFS runners at the beginning of the 5 km run. Shaded area indicates standard deviation.

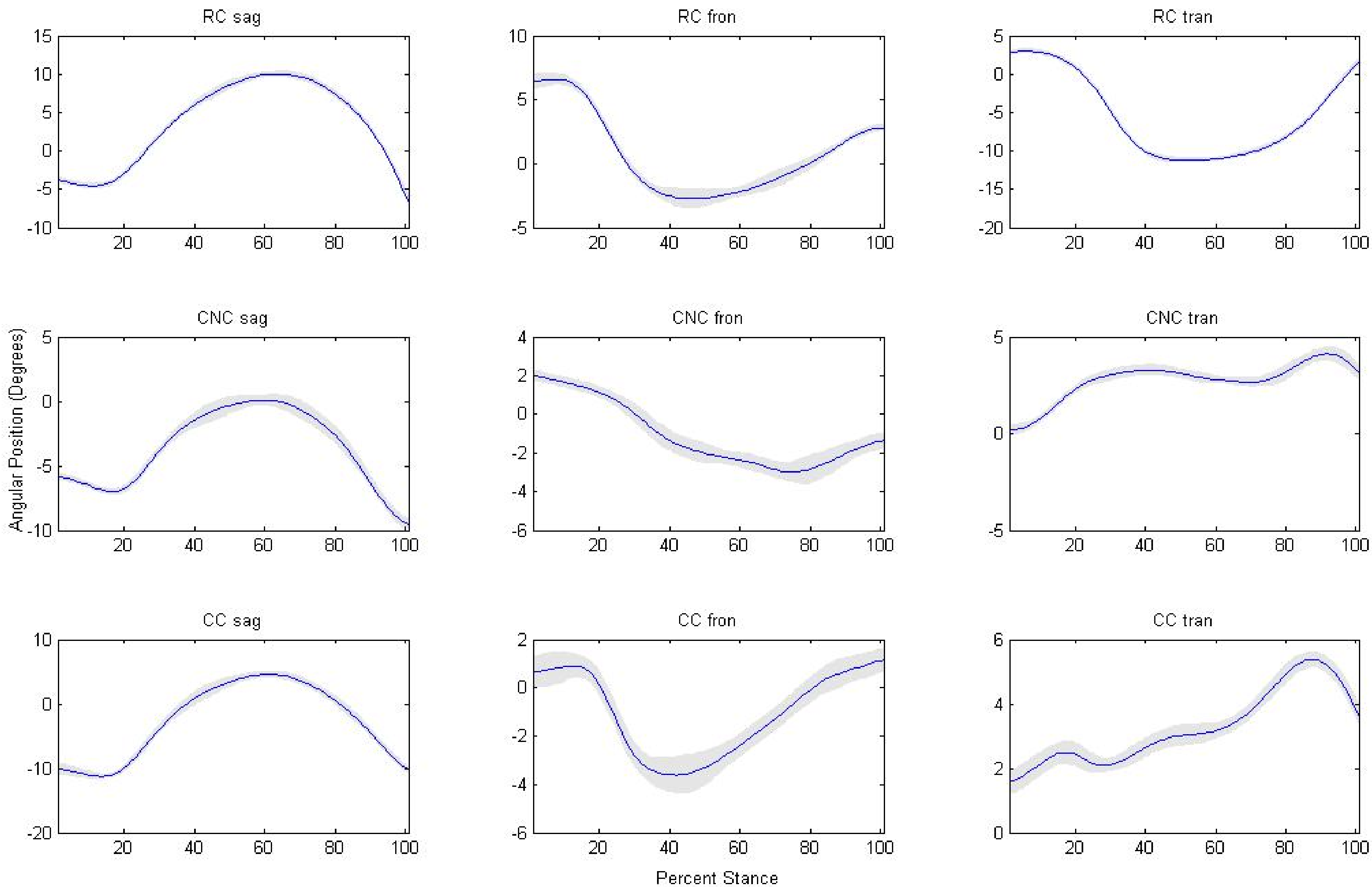

Figure P.4- Time series for joints in FFS runners at the end of the 5 km run. Shaded area indicates standard deviation.

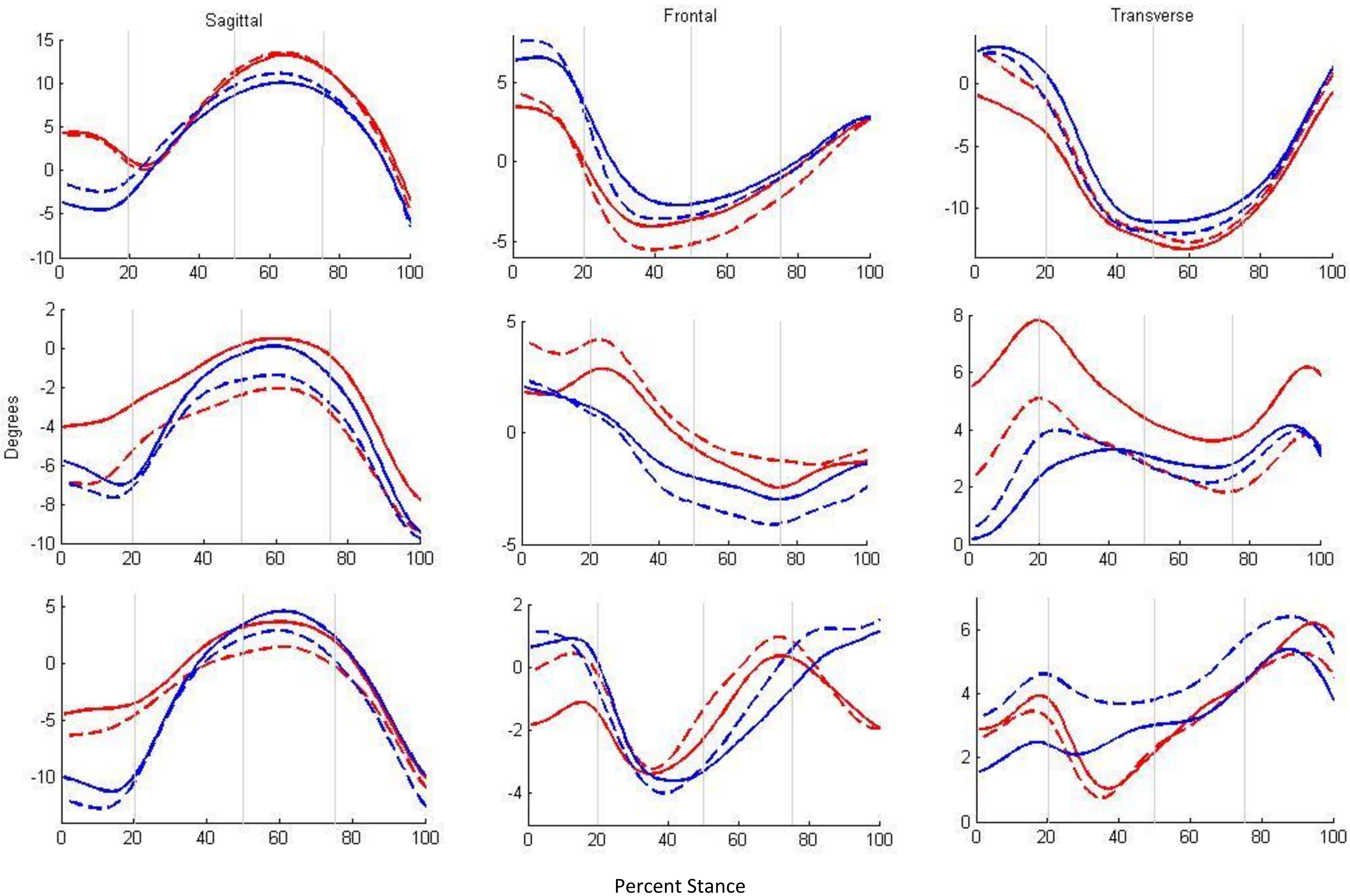

Figure P.5. Time series for the rearfoot complex (top row), calcaneonavicular complex (middle row), and calcaneocuboid complex (bottom row) for RFS (red) and FFS (blue) runners at the beginning (solid line) and end (dashed line) of a maximal 5 km run.

CURRICULUM VITAE

Rhiannon M. Seneli, MS, ATC

EDUCATION

Ph.D. in Health Sciences — University of Wisconsin-Milwaukee — 2016
Advisor: Stephen C. Cobb, Ph.D., LAT, CSCS
Dissertation title: Foot joint coupling and neuromuscular patterns in habitual forefoot and rearfoot runners
Areas of study: Biomechanics (primary), Motor Behavior (secondary)

M.S. in Kinesiology — University of Wisconsin-Milwaukee — 2011
Advisor: Ann C. Snyder, Ph.D., CSCS*D, FACSM
Thesis title: Comparison of Estimated VO2max values from the Rockport Walk Test on a non-motorized, curved treadmill to traditional overground method
Areas of study: Exercise Physiology (primary), Biomechanics (secondary)

M.S. in Exercise Science — University of Utah
Areas of study: Exercise Physiology
- Only completed one semester of course work

B.S in Exercise Science — University of Utah — 2003
Areas of study: Athletic Training

PROFESSIONAL MEMBERSHIPS

National Athletic Trainers' Association (NATA) — 2002-present
Certified Athletic Trainer
Membership #: 990883; Certification # 050502063

American College of Sports Medicine (ACSM) — 2010-present
Membership #: 680855

American Society of Biomechanics (ASB) — 2013-present
Membership #: 4860

PUBLICATIONS

Refereed Journals

Seneli, R.M., Ebersole, K.T., O'Connor, K.C., & Snyder, A.C. (2013). Estimated VO2max from the Rockport Walk Test on a non-motorized curved treadmill. *J Strength Cond Res*. 27(12): 3495-3505.

Posters/Oral Presentations

Seneli, R.M., Pomeroy, R.L., & Cobb, S.C. (2015, August). A new method for identifying toe-off event running on a treadmill using kinematic data. Poster presented at The American Society of Biomechanics Annual Conference, August 5-8, 2015.

Seneli, R.M., Joshi, M.N., Bauer, R.L., & Cobb, S.C. (2013, May). Foot motion in individuals with low arches while running barefoot and in a flat sandal. Poster presented at the American College of Sports Medicine 60th Annual Meeting and 4th World Congress on Exercise is Medicine, May 28-June 1, 2013.

Seneli, R.M. & Snyder, A.C. (2012, June). Estimated VO2max from the Rockport Walk Test on a non-motorized curved treadmill. Poster presented at the American College of Sports Medicine 59th Annual Meeting and 3rd World Congress on Exercise is Medicine, May 29-June 2.

Seneli, R.M., Joshi, M.N., Bauer, R.L., & Cobb, S.C. (2012, Dec). Multi-segment foot coupling during barefoot and shod running in individuals with low arches. Poster presented at the College of Health Sciences Fall Symposium, UW-Milwaukee.

Seneli, R.M. & Snyder, A.C. (2011, Dec). Estimated VO2max from the Rockport Walk Test on a non-motorized curved treadmill. Poster presented at the College of Health Sciences Fall Symposium, UW-Milwaukee.

Seneli, R.M., Edlbeck, B.P., Myatt, C.J., Reynolds, K.G., & Snyder, A.C. (2011, May). Comparing step length between motorized and non-motorized treadmills during walking, jogging, and running. Poster presented at the American College of Sports Medicine 58th Annual Meeting and 2nd World Congress on Exercise is Medicine, May 31-June 4.

Lange, R.M. & Mattson, J.M. (2003, June). Traumatic injury in a recreational athlete. Poster presented at the National Athletic Trainers' Association 54th Annual Meeting and Clinical Symposia, St. Louis, MO.

RESEARCH AWARDS

Wisconsin Athletic Trainers' Association Research Award ($1,000) 2013
Title: Foot joint coupling and EMG patterns in habitual forefoot and rearfoot runners
Role: Principle Investigator

College of Health Sciences Research Award ($2,000) 2013
Title: Foot joint coupling and EMG patterns in habitual forefoot and rearfoot runners
Role: Principle Investigator

Great Lakes Athletic Trainers' Association research Award ($1,500) 2013
Title: Foot joint coupling and EMG patterns in habitual forefoot and rearfoot runners
Role: Principle Investigator

HONORS

College of Health Sciences Ph.D. Scholarship Award 2013

College of Health Science Annual Fall Research Symposium 2012
Third place poster and podium presentation
Seneli, R. M., Joshi, M. N., Bauer, R. L., & Cobb, S. C. Multi-segment foot coupling during barefoot and shod running in individuals with low arches.

College of Health Science Annual Fall Research Symposium 2011
First place poster and podium presentation
Seneli, R. M. & Snyder, A.C. Estimated VO2max from the Rockport Walk Test on a non-motorized curved treadmill.

Advanced Opportunity Fellowship 2010-2013

UWM Graduate School Travel Grant (ACSM Annual Convention) 2011

TEACHING EXPERIENCE

Instructor Kinesiology	St. Ambrose University	2014-present
Adjunct Instructor- Biomechanics	University of Wisconsin - Milwaukee	2013
Graduate Teaching Assistant- Biomechanics	University of Wisconsin - Milwaukee	2012

- Biomechanics lab instructor and teaching assistant to lecture

Adjunct Instructor- Exercise Physiology	University of Wisconsin - Parkside	2012
Graduate Teaching Assistant- Athletic Training Practicum	University of Utah	2006

- Shared course teaching duties and demonstrations in lab

Clinical Instructor	University of Utah	2006-2007

- Instructed undergraduate athletic training students in clinical environment

OTHER RELEVANT EMPLOYMENT

Manufacturing Health Consultant	Kinematic Consulting, Waukesha, WI	2013
Fitness Consultant	Fitness Together, Wauwatosa, WI	2008-2009
Health and Physical Education Teacher	St. Michael's School (K-12) Santo Domingo, Dominican Republic	2007
Graduate Assistant Athletic Trainer- Women's Softball	University of Utah, Salt Lake City, UT	2006-2007
English Teacher	St. John's School (K-12) La Romana, Dominican Republic	2005-2006

Personal Trainer	Bally's Total Fitness, Brookfield, WI	2005
Athletic Trainer- Boys' Soccer	Bountiful High School, Bountiful, UT	2002-2003

LANGUAGES

Fluent in reading, writing, and speaking Spanish

COMMUNITY SERVICE

Sunday School Teacher	Davenport, IA	2015-present
Children's Primary Presidency	Davenport, IA	2014-2015
Cub Scouts Den Leader	Waukesha, WI	2013-2014
YMCA Youth Soccer Coach	Waukesha, WI	2013
LDS Missionary	Dominican Republic Santo Domingo West Mission	2003-2004