NEUROMECHANICS OF BAREFOOT AND SHOD RUNNING

A Dissertation

Presented in Partial Fulfillment of the Requirements for the

Degree of Doctor of Philosophy

with a Major in Neuroscience

in the

College of Graduate Studies

University of Idaho

by

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May 2014

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AUTHORIZATION TO SUBMIT

DISSERTATION

This dissertation of Melissa A. Thompson, submitted for the degree of Doctor of Philosophy with a major in Neuroscience and titled "Neuromechanics of Barefoot and Shod Running," has been reviewed in final form. Permission, as indicated by the signatures and dates given below, is now granted to submit final copies to the College of Graduate Studies for approval.

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Abstract

In this dissertation, I integrated empirical biomechanical analyses with detailed musculoskeletal modeling and forward dynamics simulations to gain novel insight into neural and mechanical factors that influence barefoot and shod running. This powerful combination of empirical and theoretical approaches enabled me to directly measure changes in gait mechanics as well as estimate differences in parameters relating to both performance and injury. My research chapters specifically focused on evaluating the effect of shoes, stride-length, and ankle position at ground contact, on and lower extremity kinetics and kinematics. In my first research chapter (Chapter 2), I independently evaluated the effects of shoes and changes in stride length on lower extremity kinetics. I found that reducing stride length resulted in decreased ground reaction forces and joint moments in both barefoot and shod running, while the presence or absence of shoes alone did not affect kinetics. This suggests that running barefoot leads to sensory changes that trigger a reduction in stride length, which leads to a decrease in ground reaction forces and joint moments. For my second research chapter (Chapter 3), I examined how naturally forefoot and naturally rear-foot striking runners adapt to barefoot running. The results of this study showed that natural rear-foot strike runners run similar to natural forefoot runners when running barefoot, and indicates the importance of considering foot strike position when evaluating the effects of barefoot and shod running. In my third research chapter (Chapter 4), I investigated the effects of ankle position and running shoes on running kinetics and impact accelerations. I show that, despite increased GRF impact peak magnitudes in shod running, both barefoot running with forefoot ground contact and shod running with heel

contact resulted in reduced impact accelerations as compared to barefoot running with heel contact. In my final research chapter (Chapter 5), I used a detailed musculoskeletal model and forward dynamics simulations to quantify elastic energy stored in the Achilles tendon during forefoot strike and rear-foot strike running. Preliminary results suggest that forefoot strike running results in greater utilization of elastic energy; however, better simulation data is needed before making definitive conclusions.

Acknowledgments

Though the following dissertation is an individual work, I would have never been possible without the help, support and guidance from many individuals. I would like to first thank my advisor, Dr. Craig McGowan, for taking on an “advisor-less” graduate student nearly as soon as he arrived at the University of Idaho. Craig’s patience, motivation, enthusiasm, and immense knowledge helped me continuously in the research and writing of this dissertation. I could not have imagined having a better advisor and mentor for my Ph.D study and I am honored to be his first doctoral student. Additionally, Dr. Jeff Seegmiller was vital in providing continued support and the equipment necessary to complete this research. The members of my committee, past and present, also offered valuable constructive criticism along the way.

Special thanks go to several individuals; Dr. Anne Gutmann for her gracious assistance during many steps of this project; my sister, Dr. Kristine Hoffman, for her multiple revisions of manuscripts; Dr. Chantal Vella for her endless support and for allowing me to utilize laboratory space; Dr. David Paul for his statistical advice and continued support; members of the University of Idaho’s Comparative Neuromuscular and Biomechanics Laboratory and Human Performance Laboratory, especially David Ruiz, TJ Anderson, Katie Shine, Katie Taylor, Vanessa Martinez, Amber Green and Ryan McGrath; Dr. Stephen Lee for his statistical assistance; Dr. Rodger Kram for starting my career as a researcher; Neuroscience faculty for helping me formulate my ideas and projects; and the individuals who volunteered their time to participate as research subjects. I received funding from the University of Idaho Student Grant Program that allowed me to purchase equipment that was essential for this project. Finally, I would like to thank my friends, family and dog (Beau) for their constant support.

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Chapter 1. Introduction

Locomotion is a seemingly automatic process, yet it requires complex integration of supraspinal input, centrally generated command signals and sensory feedback. The complexity of locomotion can be seen in the fact that within the lower limbs there are 244 degrees of freedom and 58 muscles, which leads to an infinite number of possibilities to create a given gait (Prilutsky & Zatsiorsky, 2002). Additionally, due to dynamic coupling, muscles are able to accelerate joints and segments that they do not span (Zajac, 2003). Despite the vast number of alternatives, the movements and muscle activation patterns of human walking and running are very stereotyped. This suggests that specific factors are integrated into neurological control signals and used to optimize gait patterns.

A primary reason why humans walk and run in stereotypical manners is that the central nervous system has organizational networks termed central pattern generators (CPGs) that control the alternating flexion and extension movements associated with gait (Marder & Calabrese, 1996). CPGs allow for ambulation without supraspinal input and can be evidenced by induced locomotion in decerebrate animals and paralyzed individuals (Forssberg et al., 1977; Forssberg et al., 1980; Pearson K., 2003). Although CPG anatomy has not been isolated in humans, there is significant evidence to suggest that CPG networks exist in the spinal cords of humans. For example, Dimitrijevic *et al*., (1998) showed that epidural spinal cord stimulation elicited step-like EMG activity and locomotor synergies in paraplegic subjects. CPGs create the alternating flexor and extensor activation seen in locomotion, but feedback from sensory receptors is essential for the stereotypical locomotor pattern seen in intact animals (Büschges & El Manira, 1998). Four principles underlie the integration of sensory feedback to CPG function: 1) sensory feedback contributes to the generation and maintenance of rhythmic activity; 2) phasic sensory signals initiate the major phase transitions in intact motor systems; 3) sensory signals regulate the magnitude of ongoing motor activity; and 4) transmission in a reflex pathway can vary during a movement (Pearson & Ramirez, 1997). While sensory feedback is an essential component of CPG output, the role that specific sensory components play in human gait has remained elusive because *in vivo* measurements are not possible. However, research suggests that several sources of afferent feedback are integrated into CPG control of gait.

Sensory sources that influence gait include energetic and mechanical feedback, which provide the individual with information about the environment and their physiological state. Mechanical feedback comes from a variety of sources including stretch receptors, joint receptors and cutaneous mechanoreceptors. Sensory feedback related to metabolic cost comes from a number of sources, including the heart, lungs, skeletal muscles and other interoceptors. It is clear that both mechanical and energetic feedback play a role in creating stereotyped gait patterns. However, it remains unknown how variations in these types of sensory feedback influence gait.

Individuals have spontaneously selected gait characteristics, termed "preferred gait" and they are also able to voluntarily modify their gait. It has been shown that preferred gait parameters, specifically stride frequency, stride length and gait velocity, are selected so that metabolic cost is minimized (Alexander R. M., 2001; Bertram, 2005; Bertram & Ruina, 2001; Zarrugh & Radcliffe, 1978). However, when gait parameters are constrained actual walking and running behavior can differ substantially from what is predicted to minimize metabolic cost (Bertram, 2005; Bertram & Ruina, 2001; Gutmann et al., 2005). For example, if individuals are required to walk or run at a specific stride length, they may select a stride frequency that does not minimize metabolic cost. This indicates that metabolic feedback is not the sole driving force behind the selection of gait parameters. Additionally, because of the time course of the underlying physical phenomena, whole body metabolic feedback takes considerably longer to be sensed than feedback that is provided by other types of sensory receptors. Yet, in many circumstances, there are immediate alterations in gait in response to changes in sensory feedback. For example, barefoot running alters cutaneous feedback from the soles of the feet and results in an immediate decrease in stride length. Thus, whole-body metabolic feedback cannot solely explain the selection of gait parameters.

Interventions, such as the presence or absence of shoes, can cause alterations in sensory feedback that lead to gait changes. Several of these gait modifications have been associated with altering the risk of musculoskeletal injuries, specifically in running. Running is one of the most popular recreational activities, but runners are also one of the most common groups to incur overuse injuries (Caspersen et al., 1984; Jacobs & Berson, 1986; Lysholm & Wiklander, 1987; Rolf, 1995). Given that an estimated 65-70% of runners are injured annually, a great deal of research has focused on running and running related injuries (van Mechelen, 1992). Musculoskeletal injuries are more common in running than walking, as running is associated with greater ground reaction forces (GRFs) and loading rates (Cavanagh & Lafortune, 1980). During the ground contact phase of running, the body's mass is rapidly decelerated resulting in impulse forces that propagate through the musculoskeletal system. These impulse forces are progressively attenuated by passive structures, such as the heel pad, articular cartilage, intervertebral discs and menisci (McNair & Marshall, 1994; Nigg et al., 1995). These forces are also attenuated by actively attenuated muscle (Derrick et al., 1998; Nigg & Liu, 1999). It has been suggested that the repetitive attenuation of impact forces results in excessive loading of biological tissues, and thus, contributes to overuse injuries (Collins & Whittle, 1989; Gill & O'Connor, 2003a; Gill & O'Connor, 2003b; MacLellan & Vyvyan, 1981; Voloshin et al., 1981).

Injury prevention efforts have focused on reducing external mechanical loading associated with running. The most common intervention aimed at reducing running related injuries is running shoes. Running shoe companies have implemented numerous material and design concepts with the goal of reducing mechanical loading. Recently barefoot running has been suggested as a potential mechanism to reduce running injuries (Lieberman et al., 2010). Barefoot running has been associated with kinetic and kinematic changes, specifically, decreased stride length and a more plantar flexed position at ground contact, which may have implications for injury prevention (McNair & Marshall, 1994).

In addition to injury prevention, a number of interventions and technique changes have been proposed to attempt to improve running performance. Running shoes, barefoot running (Altman & Davis, 2012) and alterations in spatio-temporal parameters (stride frequency & stride length (Derrick et al., 1998; Heiderscheit et al., 2011; Hobara et al., 2011; Lafortune, 1991; White & Lage, 1993) have been associated with significant kinematic and kinetic changes, which may have performance implications.

The four research chapters of my dissertation focus on understanding neural and mechanical factors related to barefoot and shod running. These studies integrate empirical biomechanical analyses with detailed musculoskeletal modeling and forward dynamics simulations to gain novel insight into the neuromechanics of barefoot and shod running. This powerful combination of empirical and theoretical approaches enabled me to directly measure changes in gait mechanics as well as estimate differences in parameters relating to both performance and injury.

In my first research chapter (Chapter 2), *The Effect of Stride Length on the Dynamics of Barefoot and Shod Running*, I independently evaluated the effects of shoes and changes in stride length on lower extremity kinetics. A significant main effect of stride length on anterior-posterior and vertical GRFs and sagittal plane knee and ankle moments was found for both barefoot and shod running. When subjects ran at identical stride lengths in the barefoot and shod conditions there were no significant differences for any of the kinetic variables that were measured. These findings suggest that sensory changes associated with barefoot running trigger a decrease in stride length, which could lead to a decrease in GRFs and sagittal plane joint moments. Thus, when evaluating barefoot running as a potential option to reduce injury, it is important to consider the associated changes in stride length. This chapter has been accepted for publication by the *Journal of Biomechanics*.

My second research chapter (Chapter 3), *The Effect of Barefoot and Shod Running on Forefoot and Rear-foot Strike Runners*, examines how naturally forefoot striking and naturally rear-foot striking runners adapt to barefoot running. The purpose of this study was to determine if barefoot running differed from natural forefoot running. The results of this study show that natural rear-foot strike runners run similar to natural forefoot runners when running barefoot. It is likely that the observed changes are due to differences in sensory feedback. While it is unclear if adopting a forefoot pattern will reduce injury, the results of this study indicate the importance of considering foot strike position when evaluating the effects of barefoot and shod running. This chapter will be submitted to *Biology Letters*.

In my third research chapter (Chapter 4), *Impact Accelerations of Barefoot and Shod Running*, I investigated the effects of ankle position and running shoes on running kinetics and impact accelerations. I show that, despite shod running being associated with increased GRF impact peak magnitudes, both barefoot running with a plantarflexed position at ground contact and shod running with a dorsiflexed position at ground contact resulted in reduced impact related accelerations as compared to barefoot running with dorsiflexed position at ground contact. This result suggests that both GRFs and impact accelerations should be considered when evaluating the potential for injury prevention. This chapter will be submitted to the *Journal of Applied Biomechanics*.

In my final research chapter (Chapter 5), *Elastic Energy Storage and Return in Forefoot and Rear-foot Strike Running*, I use a detailed musculoskeletal model and forward dynamics simulations to quantify the amount of elastic energy stored in the Achilles tendon during forefoot strike and rear-foot strike running. The results of this study show that forefoot strike running results in greater elastic energy storage, but also greater total mechanical work. Thus, forefoot strike running likely does not offer a performance advantage over rear-foot strike running. This chapter will be submitted to *Medicine and*

Science in Sports and Exercise.

Through the course of my research, I independently evaluated a number of factors that have been proposed to influence running injury risk and performance. The rationale for this approach comes from the fact that a number of gait variables are interconnected and altering one variable will result in a cascade of changes (e.g. barefoot running results in a reduction in stride length and a forefoot strike pattern). While no absolute conclusions regarding performance or injury prevention can be reached from the results of these studies, the findings suggest that stride length, foot strike and shoes are factors that should be considered when evaluating performance and injury prevention.

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Chapter 2. The Effect Of Stride Length On The Dynamics Of Barefoot And Shod Running

Thompson, M., Gutmann, A., Seegmiller, J. & McGowan, C.P. (2014). The Effect Of Stride Length On The Dynamics Of Barefoot And Shod Running. *Journal of Biomechanics*. (in press).

Abstract

A number of interventions and technique changes have been proposed to attempt to improve performance and reduce the number of running related injuries. Running shoes, barefoot running and alterations in spatio-temporal parameters (stride frequency & stride length) have been associated with significant kinematic and kinetic changes, which may have implications for performance and injury prevention. However, because footwear interventions have been shown to also affect spatio-temporal parameters, there is uncertainty regarding the origin of the kinematic and kinetic alterations. Therefore, the purpose of this study was to independently evaluate the effects of shoes and changes in stride length on lower extremity kinetics. Eleven individuals ran over-ground at stride lengths ± 5 and 10% of their preferred stride length, in both the barefoot and shod condition. Three-dimensional motion capture and force plate data were captured synchronously and used to compute lower extremity joint moments. We found a significant main effect of stride length on anterior-posterior and vertical GRFs, and sagittal plane knee and ankle moments in both barefoot and shod running. When subjects ran at identical stride lengths in the barefoot and shod conditions we did not observe differences for any of the kinetic variables that were measured. These findings suggest that barefoot running triggers a decrease in stride length, which could lead to a decrease in GRFs and sagittal

plane joint moments. When evaluating barefoot running as a potential option to reduce injury, it is important to consider the associated change in stride length.

Introduction

Given the popularity of running, a number of interventions and technique changes have been proposed to attempt to improve performance and reduce the number of running related injuries. Altering running conditions can lead to kinematic and kinetic changes that may optimize muscle/tendon function and/or reduce stress on biological tissues (e.g. tendon, ligament, cartilage). Running shoes, barefoot running (Altman & Davis, 2012) and alterations in spatio-temporal parameters (stride frequency & stride length) (Derrick et al., 1998; Heiderscheit et al., 2011; Hobara et al., 2011; Lafortune et al., 1996; White & Lage, 1993) have been associated with significant kinematic and kinetic changes, which may have implications performance and injury risk. However, some interventions, such as footwear, have been shown to also affect spatio-temporal parameters (Altman & Davis, 2012), which leads to uncertainty regarding the origin of the kinematic and kinetic alterations. Therefore, the aim of this study was to systematically manipulate shoe conditions (barefoot vs. shod) and stride length in order to understand how these conditions independently affect running dynamics.

Several previous studies have evaluated kinetic differences between barefoot and shod running, yet there is no consensus on potential differences in ground reaction forces (GRFs). For example, Divert *et al*. (2005), Lieberman *et al.* (2010) and Squadrone & Gallozzi

(2009) report decreased impact forces in the barefoot condition, whereas De Wit *et al*. (2000) and Dickinson *et al*. (1985) found no difference in impact peaks between the barefoot and shod running, and Komi *et al*. (1987) found greater impact peaks in the barefoot condition. Varied results have also been reported for the differences in the GRF active peak between barefoot and shod running. Braunstein *et al*. (2010), Divert *et al*. (2005) and Kerrigan *et al*. (2009) report greater active peaks in the shod condition, while De Wit *et al*. (2000) and Squadrone & Gallozzi (2009) found no difference in the GRF active peak between barefoot and shod running. Potential differences in the anterior-posterior GRF components are also unclear; with Divert *et al*. (2005) reporting greater propulsive forces in the barefoot condition and De Wit *et al*. (2000) finding no difference in either braking or propulsive force between conditions.

One possible reason for the discrepancy in GRFs between barefoot and shod running may be differences in kinematic alterations when individuals run barefoot. While shoes offer an obvious protective benefit, the elevated and cushioned heel of modern running shoes leads many individuals to adopt a rear-foot strike pattern, which may increase collision forces (Lieberman et al., 2010) and joint moments (Kerrigan et al., 2009). Alternatively, when running barefoot, individuals tend to have a decreased range of motion at the knee, ankle and hip (Jenkins & Cauthon, 2011); a more plantarflexed position at ground contact (Divert et al., 2005; Lieberman et al., 2010; Squadrone & Gallozzi, 2009); and a significantly shorter stride length as compared to the shod condition (De Wit et al., 2000; Divert et al., 2005; Kerrigan et al., 2009; Komi et al., 1987; Lieberman et al., 2010;

Squadrone & Gallozzi, 2009). The change in stride length is of particular interest as it has both kinematic and kinetic implications.

While differences in running kinetics associated with barefoot and shod running have received considerable attention, the independent changes associated with running in shoes (versus barefoot) and stride length have not be evaluated. In a study of shod runners, Farley & Gonzalez (1996) have shown that peak vertical GRFs decreased significantly with decreases in stride length. Derrick *et al.* (1998) report that decreasing stride length in the shod condition resulted in decreased lower extremity joint moments, though no statistics are provided. Further, Kerrigan *et al*. (2009) found that peak lower extremity joint moments were reduced in barefoot running, but there was little correlation between the decreased stride length associated with the barefoot condition and the decreased joint moments. While studies have clearly shown that barefoot running results in reduced stride lengths relative to shod running, it remains unclear how these changes influence joint dynamics. Therefore, in this study we independently evaluated the effects of shoes and changes in stride length on lower extremity kinetics. We hypothesized that peak ground reaction forces and joint moments would not differ for conditions of similar stride length and running velocity, regardless if an individual was running in shoes.

Methods

Eleven healthy, physically active adults [6 men and 5 women, age: 29 ± 5.6 yr; height: 1.63 ± 0.12 m; mass: 62.6 ± 12.1 kg] participated in this study. Subjects were required to perform a minimum of 30 minutes of physical activity at least 5 days a week, and be free of musculoskeletal injury of the lower extremities or back. The University of Idaho's Institutional Review Board approved the protocol for this study, and written informed consent was obtained from each subject (see Appendix 1).

Experimental Protocol

Subjects completed 2 testing sessions, which were separated by a minimum of 24 hours. Subjects began each session with 5-10 minutes of easy running in order to warm-up and habituate to the runway. Session 1 was used to determine the subject's preferred running gait (i.e., self-selected stride length and running velocity) in both barefoot and shod conditions. For the baseline (preferred) conditions, stride length and running velocity were averaged from 5 trials. In Session 2, subjects ran with their stride length manipulated to ± 5 and 10 % of their preferred shod stride length while barefoot and shod, for a total of 8 conditions. Both barefoot and shod stride length manipulations were based on the preferred shod condition so that in subsequent trials (e.g. +10%) the stride length was identical in both the barefoot and shod conditions. This allowed the effects of stride length and the shod/barefoot conditions to be independently evaluated. Additionally, all conditions in Session 2 were completed at the subject's preferred shod running velocity. In order to control running velocity, subjects matched their speed to a target on a motor driven pulley system that ran parallel to the runway. Stride length was controlled by having subjects match step length to strips of tape placed along the runway (Fig. 2.1). In Session 2, subjects performed 10 trials of each condition and 5 trials that were within 2% of the desired stride length and 5% of the desired running velocity were used for analysis.

Kinetics

3-dimensional motion analysis and GRF data were captured as subjects ran over a 15 m runway with a force plate (AMTI, Waterton, MA) embedded at 10 m. 16 retro-reflective markers were affixed with double-sided tape to specific landmarks according to the Modified Helen Hayes Marker set (Kadaba et al., 1990). Markers were placed on the right and left anterior and posterior superior iliac spines, lateral mid-thigh, lateral femoral epicondyle, lateral mid-shank, lateral malleolus, second metatarsal head and calcaneus. For the shod condition, heel and toe markers were placed on the shoes at the positions best aligning with the anatomical landmarks. Height, weight, leg length, and widths of the ankles and knees were measured for appropriate anthropometric scaling. 3-dimensional positions of each marker were captured at 250 Hz via a Vicon MX motion analysis system (Vicon, Oxford Metrics Ltd., UK). Marker trajectory data were filtered using a Woltring filtering routine with a predicted mean square error value of 4 mm^2. The three orthogonal components of the GRF data were recorded at 1000 Hz from the force plate in synchrony with the motion capture data. Force plate data were low-pass filtered at 30 Hz using a second-order Butterworth filter before being down-sampled and combined with the motion capture data.

Stride length was measured as the horizontal distance between ipsilateral heel marker minima. Running velocity was calculated as the horizontal displacement of each anterior superior iliac spine (ASIS) marker through the capture volume divided by the corresponding time. Running velocity was calculated for both the right and left ASIS markers and averaged. Joint moments were calculated for each trial via an inverse-dynamics model implemented in Vicon Plug-In Gait.

Statistics

Statistical differences in peak kinetic parameters were determined using a repeated-measures General Linear Model in SPSS (IBM, Armonk, NY). When a significant effect was identified, a post hoc Bonferroni pairwise comparison was performed to determine which conditions were significantly different. Statistical significance was defined as $P<0.05$.

Results

Preferred conditions

Subjects adopted a significantly shorter stride length and decreased running velocity in the preferred barefoot condition ($P<0.05$) (Table 2.1) compared to the preferred shod condition. There were no significant differences between the barefoot and shod condition for any of the kinetic parameters that were measured (Table 2.2).

Altered stride length conditions

There was a significant main effect for the relationship between stride length and the anterior-posterior GRF and vertical GRF ($P<0.05$) (Fig. 2.2). There was also a significant main effect for the relationship between stride length and sagittal plane ankle and knee moments ($P<0.05$) (Fig. 2.3). Peak values for GRFs and joint moments are shown in Figure 2.4. Pairwise comparisons showed that peak vertical GRFs in the shod -10%, -5% and +10% conditions differed significantly from the preferred shod condition ($P=0.007$, $P=0.023$ and $P=0.029$ respectively), and peak vertical GRFs in the barefoot +5% and +10% conditions were significantly greater than the preferred barefoot condition ($P=0.033$ and $P=0.004$ respectively) (Table 2.2). Peak anterior-posterior GRF in the shod +10% condition was significantly greater than the preferred shod condition ($P=0.030$), and peak anterior-posterior GRFs in the barefoot +5% and +10% conditions were significantly greater than the preferred barefoot condition ($P=0.010$ and $P=0.001$ respectively) (Table 2.2). In the sagittal plane, knee and ankle peak joint moments in the +10% condition were significantly greater than the preferred condition for both barefoot and shod running (Knee: $P=0.018$ and $P=0.010$ respectively, Ankle: $P=0.013$ and $P=0.017$ respectively) (Table 2.2). In the frontal and transverse planes there were no significant differences between any of the stride length conditions for any of the kinetic parameters that were measured. When subjects ran at an identical stride length (e.g. -10%) in both the barefoot and shod condition there were no significant differences in any of the kinetic parameters that were measured.

Discussion

The goal of this study was to independently evaluate the effects of shoes and changes in stride length on lower extremity kinetics. Few studies have evaluated the effect of either stride length (Derrick et al., 1998; Heiderscheit et al., 2011; Hobara et al., 2011) or shoes (Bonacci et al., 2013; Kerrigan et al., 2009) on running kinetics. Because barefoot running leads to a decrease in stride length, the results of studies comparing barefoot and shod running are complicated and it remains uncertain if stride length, shoes or other factors lead to altered kinetics. To the best of our knowledge this is the first study to independently evaluate the effect of shoes and stride length on running kinetics.

The results of this study support our hypothesis that that peak GRFs and joint moments would not differ for conditions of similar stride-length and running velocity, regardless if an individual is wearing shoes. Further, we found that stride length has a significant effect on GRFs and joint moments. Specifically, anterior-posterior and vertical GRFs, and sagittal plane knee and ankle joint moments increased with increasing stride length in both the barefoot and shod conditions.

The increased sagittal plane joint moments that were observed with increases in stride length can be explained, in part, by changes in the moment arm relative to the resultant GRF. The moment arm of the GRF depends on the point of force application relative to a joint axis of rotation and the orientation of the ground reaction force vector. With a decrease in stride length the heel is located more underneath the COM at initial contact and there is a decrease in peak hip and knee flexion (Heiderscheidt et al., 2011), which would act to reduce moment arms to the GRF (Derrick et al., 1998). Further, though information regarding changes in stride length are not reported, Braunstein *et al.* (2010) showed that wearing running shoes increased the moment arm of the GRF at both the ankle and knee in comparison to barefoot running.

Increased GRF magnitude is also an important factor in the greater peak sagittal plane joint moments that were observed with increased stride length. The greater GRFs are likely due to changes in center in mass (COM) trajectory, leg joint angles and lower extremity stiffness (Derrick, 2004; Heiderscheit et al., 2011; Lafortune et al., 1996). Increasing stride length has been shown to increase COM excursion, COM velocity and the distance from the heel and COM at ground contact, as well as decrease leg stiffness, all of which potentially influence GRFs and joint moments (Derrick et al., 1998; Farley & Gonzalez, 1996; Heiderscheit et al., 2011). These kinematic changes are similar to those reported for comparisons of barefoot and shod running, and while not addressed in the current study, the kinematic changes associated with barefoot running may also be a factor of stride length rather than the result of wearing shoes.

Our comparison of the preferred barefoot and shod conditions is consistent with previous studies, which have shown that when running barefoot individuals adopt a shorter stride length (Altman & Davis, 2012; Jenkins & Cauthon, 2011; Nigg, 2009). Further, we have also shown that when running over-ground, on average, inexperienced barefoot runners run slower than they do in the shod condition. The majority of studies that have evaluated barefoot and shod running have utilized a motor driven treadmill (e.g. Divert et al., 2005; Kerrigan et al., 2009; Squadrone & Gallozzi, 2009), which as compared to over-ground running, have been associated with changes in temporal factors (Elliott & Blanksby, 1976; Nelson et al., 1972) kinematics (Nigg et al., 1995; Wank et al., 1998) and kinetics (Riley et al., 2008). While treadmills create a standardized environment, running velocity is determined by the motor speed rather than being selected by the runner. Alternatively, over-ground running allows the runner to freely accelerate and decelerate, thereby allowing for more accurate analysis of a runner's preferred velocity. Velocity changes have important kinetic implications, as increased running speeds are associated with greater joint moments and GRFs (e.g., Arampatzis et al., 1999; Hamill et al., 1983). Increased running speeds have also been linked with increased injury risk (Hreljac, 2004). While we were interested in observing both running velocity and stride length changes in the preferred gait conditions, our primary goal was to examine the effect of stride length. Therefore, to avoid the complications associated with running velocity, we used a motorized pulley system to ensure that individuals ran at their preferred shod running velocity for the altered stride length conditions.

Our findings of increased GRFs and sagittal plane joint moments are comparable to the results of Derrick *et al*. (1998) and Heiderscheit *et al*. (2011) who examined the effect of stride length on shod running dynamics. Our results show a significant main effect of stride length on anterior-posterior and vertical GRFs, and sagittal plane knee and ankle moments in both barefoot and shod running. At a given stride length, we did not observe significant differences between barefoot and shod running for any of the variables tested. Further, we found no significant differences between the preferred barefoot and shod conditions, despite an average difference in stride length of 8.5%. It is possible that no significant differences were observed between the preferred conditions because there was considerable individual variation in the magnitude of stride length and velocity changes (stride length range = 1-17%, velocity range = 0-9%). While we found no significant differences between preferred barefoot and shod conditions, previous studies have reported significant differences in joint moments and GRFs between barefoot and shod running (Bonacci et al., 2013; Kerrigan et al., 2009). These studies also report differences in frontal and transverse plane moments between barefoot and shod running, whereas in the present study we only found stride length effects in the sagittal plane. However, there is little consensus in the literature as to the specific differences in GRFs and joint moments between barefoot and shod running (Bonacci et al., 2013; Kerrigan et al., 2009). For example, Kerrigan *et al*. (2009) found significantly greater frontal and transverse plane hip joint moments in shod running, whereas Bonacci *et al*. (2013) did not find significant differences in hip joint moments between barefoot and shod running. It is possible that the inconsistencies are due to methodological differences, such as over-ground versus treadmill running, the use of standardized running shoes, and/or the inverse dynamics model that was used in analysis. Further, the experience level of subjects that were tested and inherent variability between subjects could also lead to inconsistencies. Based on our results, we are confident that stride length, and not footwear, lead to the kinetic changes that were shown in the present study.

The results presented here suggest that decreasing stride length, whether barefoot or shod, produces kinetic changes that may be beneficial for prevention of running related injuries. High vertical GRFs and increased joint moments have been associated with greater risk of running related injuries (Edwards et al., 2009; Scott & Winter, 1990; van Gent et al., 2007). Our results show that decreases in stride length can reduce both vertical GRFs and joint moments. The findings of Edwards *et al*. (2009) support this conclusion, as they report that a 10% reduction in stride length decreases the risk of stress fracture. It is important to note that if individuals are to maintain the same running velocity while decreasing stride length there would have to be a corresponding increase in stride frequency, which could also potentially increase injury risk. However, Edwards *et al*. (2009) report that strain magnitude plays a more important role in stress fracture development than the total number of loading cycles. Future prospective studies should be aimed at determining if decreasing stride length can prevent running injuries.

The results presented here suggest that barefoot running itself may not lead to kinetic changes that could potentially reduce running related injuries. Rather, barefoot running triggers a decrease in stride length, which could lead to a decrease in GRFs and sagittal plane joint moments. Our results suggest that many of the biomechanical benefits attributed to barefoot running may potentially be achieved by shortening stride length, even while wearing shoes. When evaluating barefoot running as a potential option to reduce injury, it is important to consider the associated change in stride length.

Table 2.1. Preferred stride length and running velocity for barefoot and shod running.

Condition	Stride Length (m)	Velocity (m/s)
Barefoot	2.13 $\pm$ 0.31*	3.18 $\pm$ 0.48*
Shod	2.32 $\pm$ 0.36	3.31 $\pm$ 0.47

Mean (SD) values for preferred stride length and running velocity for barefoot and shod running. *Indicates significant difference between barefoot and shod conditions ($p<0.05$).

Table 2.2. Kinetic parameters.

		-10%		-5%		Preferred		+5%		+10%	
		BF	Shod	BF	Shod	BF	Shod	BF	Shod	BF	Shod
Joint Moments (N•m•kg^{-1})	Ankle Dorsiflexion	2.59 (0.33)	2.64 (0.36)	2.55 (0.37)	2.58 (0.41)	2.60 (0.45)	2.64 (0.36)	2.85 (0.43)	2.82 (0.49)	3.01 (0.54)^	2.97 (0.46)^
	Ankle Adduction	0.21 (0.12)	0.17 (0.15)	0.21 (0.08)	0.17 (0.14)	0.22 (0.13)	0.14 (0.06)	0.24 (0.13)	0.19 (0.15)	0.26 (0.24)	0.17 (0.13)
	Ankle Internal Rotation	0.43 (0.23)	0.61 (0.36)	0.41 (0.16)	0.72 (0.42)	0.66 (0.44)	0.49 (0.22)	0.55 (0.23)	0.63 (0.35)	0.59 (0.25)	0.63 (0.35)
	Knee Flexion	1.88 (0.28)	2.05 (0.43)	1.89 (0.48)	2.21 (0.53)	1.97 (0.38)	2.25 (0.46)	2.48 (0.47)	2.37 (0.34)	2.55 (0.42)^	2.67 (0.44)^
	Knee Varus	1.17 (0.55)	1.66 (0.82)	1.22 (0.32)	1.80 (0.65)	1.35 (0.76)	1.60 (0.56)	1.38 (0.58)	1.76 (0.92)	1.50 (0.58)	1.96 (0.43)
	Knee Internal Rotation	0.27 (0.22)	0.42 (0.25)	0.32 (0.16)	0.41 (0.23)	0.36 (0.19)	0.34 (0.18)	0.36 (0.20)	0.39 (0.19)	0.34 (0.15)	0.64 (0.46)
	Hip Flexion	3.10 (1.07)	2.79 (1.35)	2.85 (1.02)	3.03 (1.29)	3.07 (0.83)	2.94 (0.72)	3.25 (1.02)	3.13 (0.94)	3.89 (1.71)	3.70 (1.18)
	Hip Adduction	1.86 (0.83)	2.11 (0.61)	2.25 (0.57)	2.40 (0.47)	2.35 (0.44)	2.45 (1.17)	2.49 (0.72)	1.93 (1.08)	2.57 (0.73)	2.75 (0.45)
	Hip Internal Rotation	0.21 (0.16)	0.21 (0.16)	0.22 (0.13)	0.22 (0.13)	0.16 (0.06)	0.22 (0.17)	0.20 (0.06)	0.27 (0.20)	0.28 (0.24)	0.16 (0.05)
GRF (BW)	AP	0.40 (0.09)	0.40 (0.08)	0.41 (0.07)	0.42 (0.08)	0.44 (0.07)	0.43 (0.07)	0.52 (0.09)^	0.49 (0.11)	0.54 (0.08)^	0.49 (0.09)^
	ML	0.14 (0.07)	0.17 (0.06)	0.16 (0.04)	0.19 (0.06)	0.14 (0.03)	0.17 (0.07)	0.18 (0.07)	0.19 (0.07)	0.18 (0.07)	0.19 (0.08)
	Vertical	3.02 (0.40)^	3.13 (0.48)^	3.09 (0.38)	3.22 (0.44)	3.16 (0.44)	3.35 (0.49)	3.52 (0.48)^	3.53 (0.50)	3.60 (0.50)^	3.62 (0.46)^

Mean (SD) values for preferred stride length and running velocity for barefoot and shod running. * Indicates significant difference between barefoot and shod conditions at a given stride length. ^ Indicates a significant difference from the respective preferred condition ($p<0.05$).

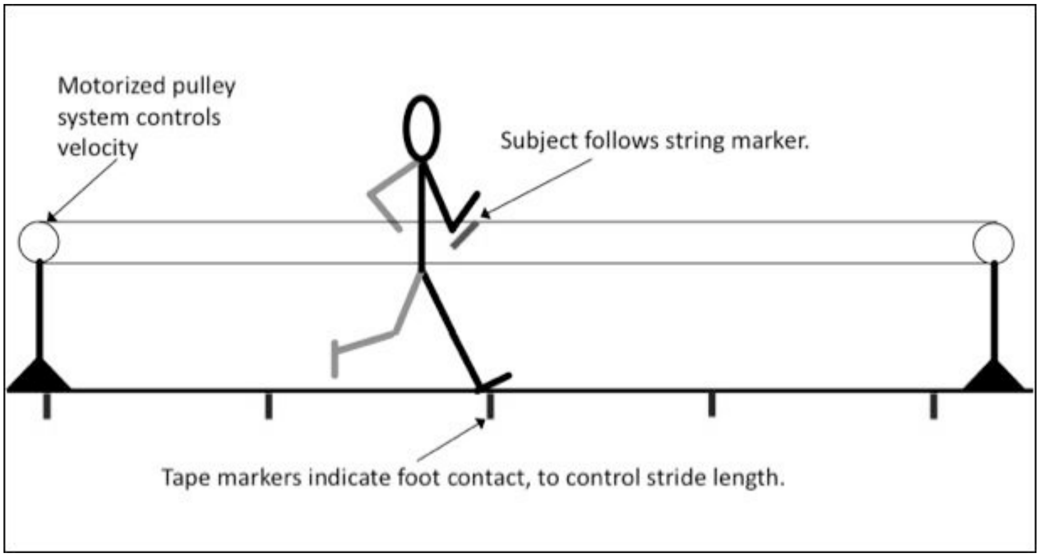

Figure 2.1. Experimental set up. The experimental set-up used for the stride-length manipulation trials. Subjects followed a string marker that was pulled at the correct velocity by a motorized pulley system. Stride length was controlled by having subjects match their footfall to tape markers placed on the floor.

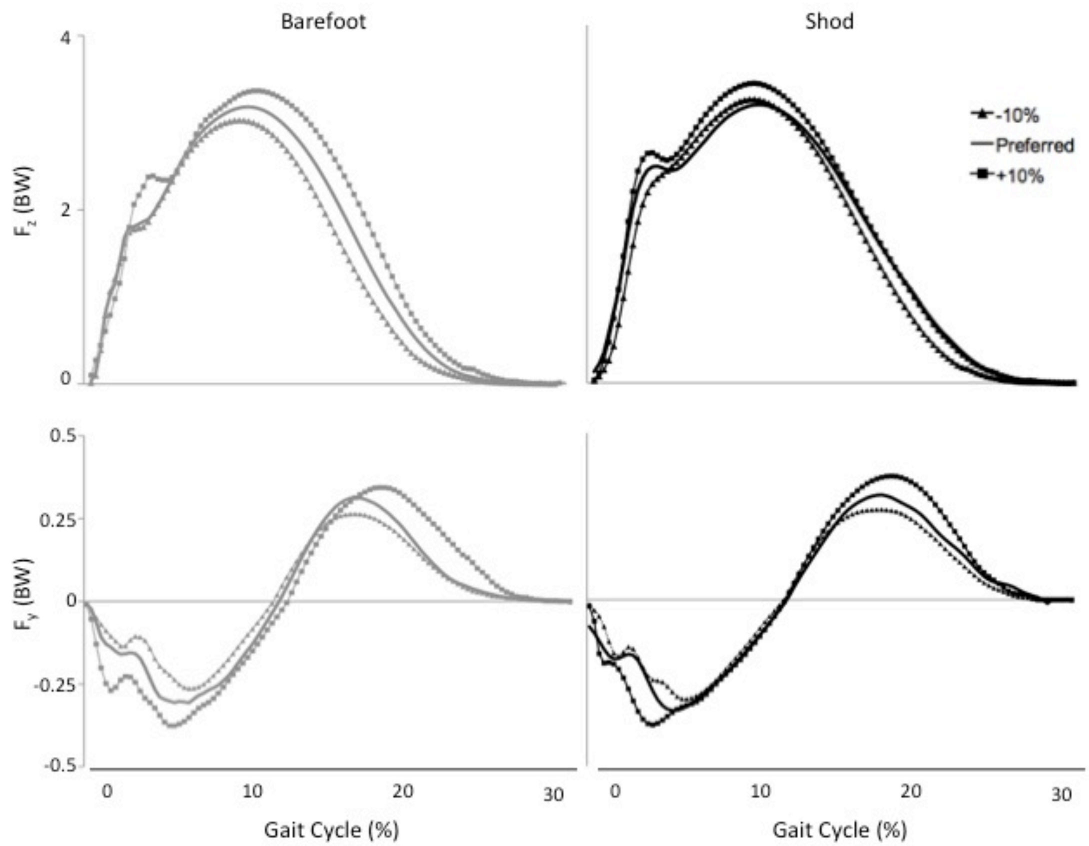

Figure 2.2. Vertical and anterior-posterior GRFs. Ensemble average vertical (F_Z) and anterior posterior (F_Y) GRFs for barefoot (gray) and shod (black) running at -10% (triangles), +10% (squares), and preferred (solid line) stride lengths.

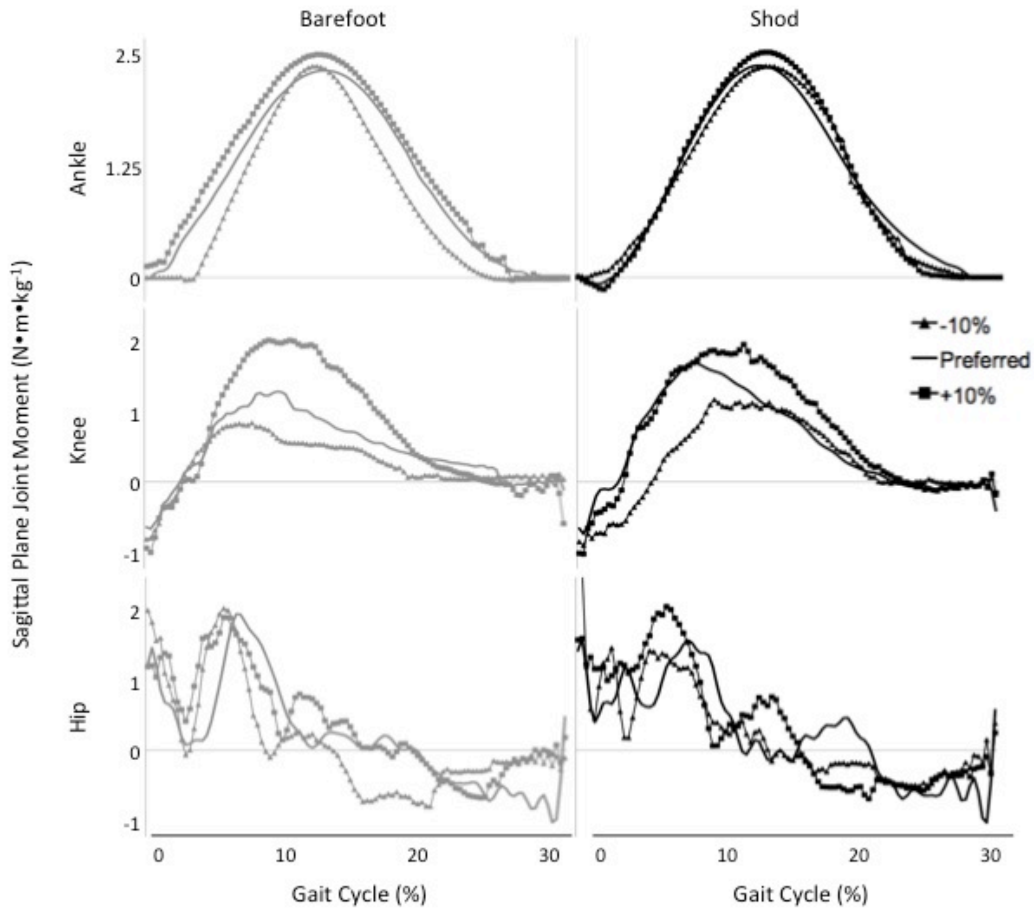

Figure 2.3. Sagittal plane joint moments. Ensemble average sagittal plane joint moments at the ankle, knee, and hip for barefoot (gray) and shod (black) running at -10% (triangles), +10% (squares), and preferred (solid line) stride lengths. Note: figures have different scales.

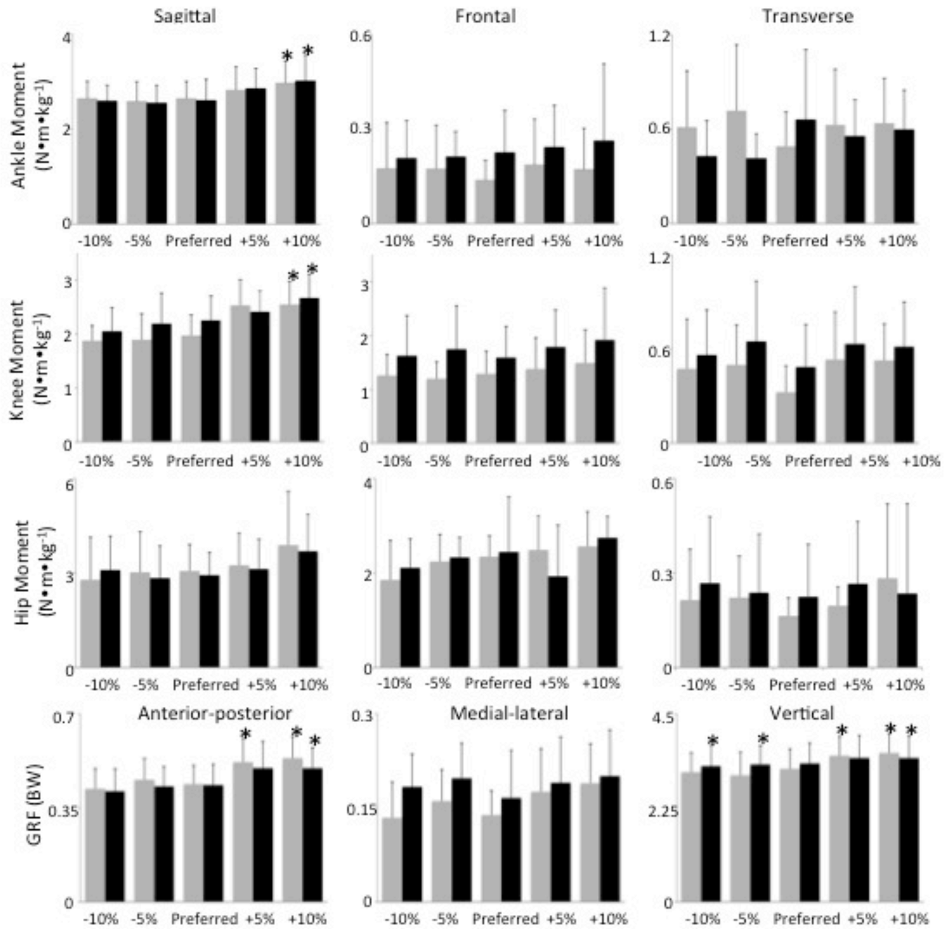

Figure 2.4. **Peak joint moments and GRFs**. Peak joint moments and GRFs ankle for barefoot (gray) and shod (black) running at in each of the stride length conditions. *Indicates significant difference from preferred condition, p<0.05. Note: figures have different scales.

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Chapter 3. The Effect Of Barefoot And Shod Running On Forefoot And Rear-Foot Strike Runners

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Introduction

Barefoot running has received much attention in the scientific literature as of late. The interest has been driven by findings of potential performance benefits and reduced risk of injury (for review see Rixie et al., 2012). It has been proposed that barefoot running triggers gait alterations that may lead to these proposed benefits (Tam et al., 2013). Gait alterations that have consistently been associated with barefoot running include a decrease in stride length and switching from a rear foot strike (RFS) pattern in the shod condition to a forefoot strike (FFS) pattern when running barefoot. However, not all individuals contact the ground with their heel. Some individuals naturally contact the ground on their forefoot, even when wearing cushioned running shoes. The purpose of this study was to determine if barefoot running differed from natural FFS running. We hypothesized that natural FFS individuals would not change their gait significantly when switching from shod to barefoot running.

Methods

Ten naturally FFS individuals [5 men and 5 women; age: 28 ± 5.9 yr; height: 1.71 ± 0.08 m; mass: 70.8 ± 10.3 kg] and ten naturally RFS individuals [5 men and 5 women; age: 29 ± 6.0 yr; height: 1.70 ± 0.09 m; mass: 65.3 ± 8.6 kg] volunteered to participate in this study. Subjects were required to perform a minimum of 30 minutes of physical activity at least 5 days a week and be free of musculoskeletal injury. Foot strike pattern was determined by visual observation. FFS was defined as ground contact on the ball of the foot (metatarsal heads) and RFS was defined as ground contact at the heel (calcaneus). The University of Idaho's Institutional Review Board approved the protocol for this study (Appendix 1).

3-dimensional motion analysis and ground reaction force (GRF) data were captured as subjects ran over a 15 m runway with a force plate (AMTI, Waterton, MA) embedded at 10 m. Subjects were instructed to run with their preferred running gait (i.e., self-selected stride length and velocity) in both barefoot (BF) and shod (SHOD) conditions. 10 strides were used to calculate participant mean data for each condition.

16 retro-reflective markers were affixed with double-sided tape to specific landmarks according to the Modified Helen Hayes Marker set (Kadaba et al., 1990). 3-dimensional marker positions were captured at 250 Hz via a Vicon MX motion analysis system (Vicon, Oxford Metrics Ltd., UK). Marker trajectory data were filtered using a Woltring filtering routine with a predicted mean square error value of 4 mm^2. The three orthogonal components of the GRF data were recorded at 1000 Hz from the force plate in synchrony with the motion capture data. Force plate data were low-pass filtered at 30 Hz

using a second-order Butterworth filter before being down-sampled and combined with the motion capture data.

To test our hypothesis, several multivariate analyses were performed in R 3.1.0 (R Development Core Team, Vienna, Austria). A two-sample Hotelling test was used to compare FFS and RFS groups for limb position at ground contact (dependent variables: 3D joint angles at the ankle, knee and hip). One-sample Hotelling tests were used to compare peak kinematics (dependent variables: 3D joint angles at the ankle, knee and hip) and kinetics (dependent variables: 3D GRFs and impact peak magnitude) between the BF and SHOD condition for both FFS and RFS running. For each significant multivariate result ($p<0.05$), uni-variate t-tests with Bonferroni corrections were performed to find which variable(s) made the significant difference.

Results

The FFS and RFS groups both showed a significant decrease in stride length when switching from shod to barefoot running (Table 3.1). When switching from shod to barefoot running the RFS group had an 8.0% reduction in stride length on average ($p=0.008$), and the FFS group experienced a 6.3% decrease ($p<0.001$). There were no significant differences between the RFS and FFS groups in stride length ($p=0.177$) or running velocity ($p=0.160$).

Position at ground contact

There was a statistically significant difference in position at ground contact between the FFS and RFS condition (p=0.0002). Post hoc tests revealed that the FFS SHOD and RFS SHOD conditions differed significantly in terms of sagittal plane ankle position (p<0.0001) and knee internal rotation (p=0.026). RFS individuals switched from a dorsiflexed position at ground contact in the SHOD condition [9.8(5.1^o)] to a more plantarflexed ankle position at ground contact when running BF [-3.3(8.7^o)] (Table 3.2). FFS individuals maintained a plantarflexed position when switching to BF running, with the amount of plantarflexion at ground contact increasing only slightly between SHOD [-4.1(6.3^o)] and BF running [-5.3 (4.6^o)] (Fig. 3.1). Additionally, the FFS group had greater knee internal rotation in both the BF and SHOD conditions [SHOD: -13.8(7.0^o), BF: -14.4(5.9^o)] than the RFS group [SHOD: -22.5(8.9^o), BF: -20.2(13.9^o)] (Table 3.2).

Peak values

The RFS runners showed a statistically significant difference in peak kinetics between the SHOD and BF conditions (p=0.0008). Post hoc tests revealed that the RFS SHOD and RFS BF conditions differed significantly in terms of vertical GRF (vGRF) (p=0.006) and medio-lateral GRF (mlGRF) (p=0.008). There were no significant differences in peak kinematics or kinetics between the FFS and RFS condition or between the SHOD and BF conditions for the FFS group (Table 3.3).

Discussion

The FFS subjects in the present study did not experience kinematic or kinetic changes when switching from shod to barefoot running. However, the FFS individuals did experience a significant decrease in stride length when running barefoot. Alternatively, the RFS individuals experienced gait changes that are commonly associated with barefoot running (Jenkins & Cauthon, 2011). Specifically, RFS individuals experienced a significant decrease in stride length, switched from a dorsiflexed to a plantarflexed position at ground contact and saw a significant reduction in impact peak magnitude. These changes reflect that when running barefoot, the RFS group ran with kinematics similar to the FFS group running barefoot or shod.

The primary kinematic difference between FFS and RFS runners was the ankle position at ground contact. RFS individuals contacted the ground with the ankle in a dorsiflexed position, which requires the tibialis anterior to decelerate ankle plantarflexion as the foot is lowered to the ground and has been associated with increased pressures in the anterior compartment of the lower leg (Kirby & McDermott, 1983; Diebal et al., 2012). Alternatively, FFS runners adopted a plantarflexed position at ground contact with the triceps surae acting to slow ankle dorsiflexion after initial contact. This plantarflexed position places the Achilles muscle-tendon complex in an eccentrically loaded position (Hammil & Gruber, 2012) and has been associated with higher Achilles tendon strain (Perl et al., 2012) and plantar flexor moments (Williams et al., 2000; Kulmala et al., 2013).

In the present study, both FFS and RFS runners reduced their stride length when running barefoot. A reduction in stride length is commonly associated with barefoot running and is thought to be due to the fact that shoes limit proprioception by blocking stimulation of the foot’s mechanoreceptors (Robins & Waked 1998; Lieberman, 2012). In

RFS runners, the improved proprioception when barefoot is thought to trigger both a reduction in stride length and a plantarflexed position at ground contact (Lieberman, 2012). With the FFS runners already assuming a plantarflexed position, there continues to be a reduction in stride length when running barefoot, indicating that the barefoot condition allows for greater perception of impact (Robbins & Gouw, 1991; Robbins & Hanna 1987).

Similar to previous studies, we found that the magnitude of vertical impact peak was lower in shod FFS runners than shod RFS runners (Lieberman et al., 2010). It has been proposed that FFS runners absorb impact through compression of the medial longitudinal arch of the foot, eccentric contraction of the triceps surae, and stretch of the Achilles tendon leading to a reduction in impact peak magnitude and loading rate (Lieberman et al., 2010). Alternatively, in RFS running impact absorption is limited to the heel pad and shoe leading to higher impact peak magnitudes and loading rates. Barefoot RFS is commonly associated with impact peaks magnitudes of 1.5-2.5 body weights and high vertical loading rates (Doud et al., 2012). The cushioned heel of running shoes attenuates this impact, but the magnitude of the impact peak and loading rate in shod RFS runners remains on average greater than in FFS runners. We have further shown that when RFS runners adopted a plantarflexed position at ground contact in the barefoot condition impact peak magnitude was reduced to levels comparable to the FFS group. Alternatively, there was no difference in the magnitude of the impact peak when FFS runners ran barefoot. These results support the notion that RFS runners adopt a FFS pattern when running barefoot in an attempt to reduce impact peaks or high loading rates. It is important to note that there is considerable individual variation in the presence and magnitude of the impact peak in FFS and RFS runners. In the present study, 50% of FFS runners had an impact peak on at least 3 of the trials used for analysis. Of these runners, only 30% saw a reduction in impact peak magnitude when switching to barefoot running. Alternatively, 90% of the RFS individuals had an impact peak present and nearly all of these individuals saw a reduction in impact peak magnitude when switching to barefoot running.

Certain limitations should be considered when interpreting the findings of the present study. First, it is important to note that the subjects were not pair matched. Therefore, some of the differences between groups may be attributable to anthropometric and velocity differences. Additionally, we did not distinguish between forefoot strike (initial contact with the ball of foot) and midfoot strike (simultaneous contact with heel and ball of foot) runners, which may be associated with kinematic and kinetic differences (Doud et al., 2012; Robbins & Hanna, 1987).

In conclusion, the results presented here show that natural RFS runners run similar to natural FFS runners when running barefoot. It is likely that the observed changes are due to differences in proprioception. While it is unclear if adopting a FFS pattern will reduce injury, the results of this study indicate the importance of considering foot strike position when evaluating the effects of barefoot and shod running.

Table 3.1. Gait Parameters

	RFS		FFS	
	SHOD	BF	SHOD	BF
Stride Length/ Leg Length	**2.53 (.27)**#	**2.34 (.24)**	**2.62 (.23)**#	**2.46 (.24)**
Velocity (m/s)	3.13 (.30)	3.01 (.28)	3.35 (0.56)	3.19 (.54)

Data are mean (standard deviation). Significant results are represented with bold numbers. (*) indicates a significant difference between foot strike pattern (FFS vs. RFS) for a given shoe condition (barefoot or shod) ($p<0.05$). (#) indicates a significant difference between shoe conditions (barefoot vs. shod) for a given foot strike pattern ($p<0.05$).

Table 3.2. Kinematics at Ground Contact

	RFS		FFS	
	Shod	BF	Shod	BF
Ankle Dorsiflexion (^{o})	**9.8 (5.1)*** #	**-3.3 (8.7)**#	**-5.3 (4.6)***	-4.1 (6.3)
Ankle Adduction (^{o})	-1.9 (3.6)	1.9 (7.3)	-0.6 (3.4)	0.8 (2.4)
Ankle Internal Rotation (^{o})	8.05 (12.0)	4.2 (16.3)	0.5 (13.6)	-3.3 (11.3)
Knee Flexion (^{o})	11.8 (7.9)	13.9 (8.8)	14.3 (4.7)	13.9 (6.9)
Knee Varus (^{o})	-1.6 (7.4)	2.9 (8.1)	2.2 (5.9)	3.7 (5.6)
Knee Internal Rotation (^{o})	**-22.5 (8.9)***	**-20.2 (13.9)***	**-13.8 (7.0)***	**-14.4 (5.9)***
Hip Flexion (^{o})	38.1 (7.5)	35.4 (8.5)	34.8 (8.1)	33.6 (7.7)
Hip Adduction (^{o})	7.5 (4.0)	4.8 (5.3)	7.6 (3.8)	6.3 (3.7)
Hip Internal Rotation (^{o})	12.2 (8.6)	13.5 (10.2)	13.8 (19.3)	12.9 (18.2)

Data are mean (standard deviation). Significant results are represented with bold numbers. (*) indicates a significant difference between foot strike pattern (FFS vs. RFS) for a given shoe condition (barefoot or shod) ($p<0.05$). (#) indicates a significant difference between shoe conditions (barefoot vs. shod) for a given foot strike pattern ($p<0.05$).

Table 3.3. Ground Reaction Forces

	RFS		FFS	
	Shod	BF	Shod	BF
Peak vGRF (BW)	2.44 (.33)	2.21 (.49)	2.46 (.42)	2.48 (.42)
Peak positive apGRF (BW)	0.24 (.06)	0.24 (.12)	0.29 (.05)	0.26 (.10)
Peak mlGRF (BW)	0.12 (.06)	0.09 (.05)	0.17 (.07)	0.12 (.06)
Impact Peak (BW)	**1.77 (0.20)** * #	**1.52 (0.17)** #	**1.51 (0.26)** *	1.52 (0.10)

Data are mean (standard deviation) in units of body weight (BW). Significant results are represented with bold numbers. (*) indicates a significant difference between foot strike pattern (FFS vs. RFS) for a given shoe condition (barefoot or shod) (p<0.05). (#) indicates a significant difference between shoe conditions (barefoot vs. shod) for a given foot strike pattern (p<0.05).

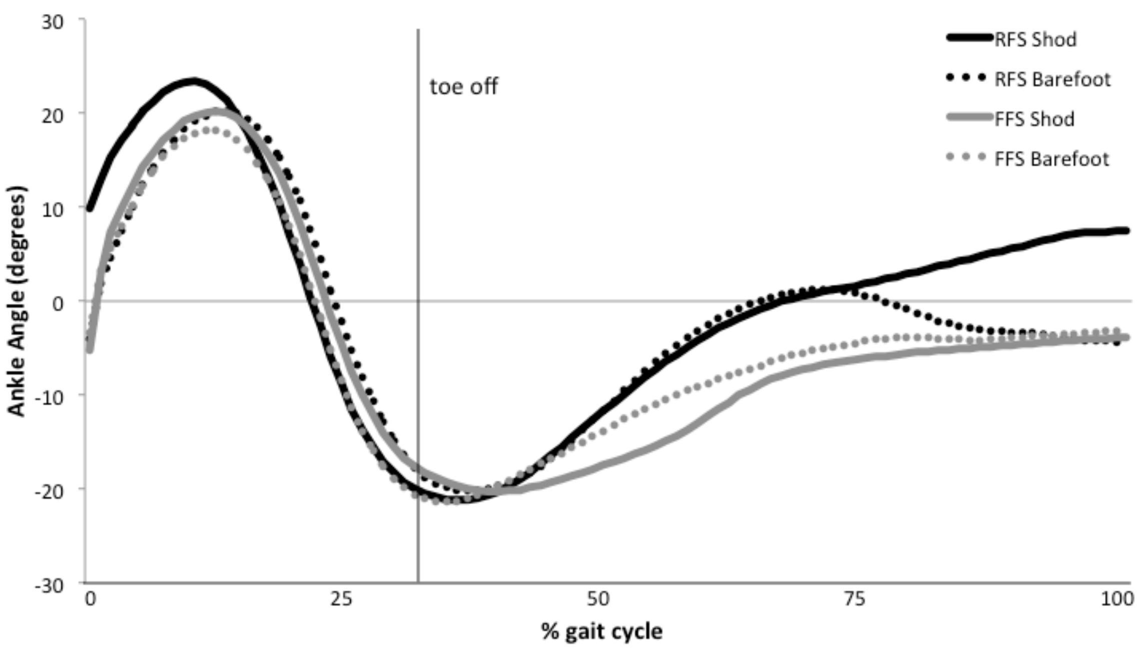

Figure 3.1. Sagittal plane ankle angle. Ensemble average sagittal plane ankle angle for a complete step. RFS (gray) and shod (FFS) running in the shod (solid) and BF (dashed) conditions. There was a significant interaction of foot strike and shoe condition for sagittal plane ankle position at ground contact (i.e. 0%).

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Chapter 4. Impact Accelerations Of Barefoot And Shod Running

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Introduction

Running is one of the most popular recreational activities, but runners are also one of the most common groups to incur overuse injuries (Caspersen et al., 1984; Jacobs & Berson, 1986; Lysholm & Wiklander, 1987; Rolf, 1995). Given that an estimated 65-70% of runners are injured annually, a great deal of research has focused on running and running related injuries (van Mechelen, 1992). A number of interventions have been proposed to try and reduce running related injuries, the most common of which is running shoes. Recently, barefoot running has been suggested as a potential mechanism to reduce running injuries (Lieberman, 2012). Barefoot running has been associated with kinetic and kinematic changes, specifically, decreased stride length and a more plantarflexed position at ground contact, which may have implications for injury prevention. (McNair & Marshall, 1994)

During the ground contact phase of running, the body's mass is rapidly decelerated resulting in impulses of force that propagate through the musculoskeletal system. These impulses are progressively attenuated as they travel to the head, by passive structures such as the ground, shoe midsole, heel pad, articular cartilage, intervertebral discs and menisci (McNair & Marshall, 1994; Nigg et al., 1995). Impulses can also be actively attenuated by eccentric activation of the muscles crossing the hip, knee, and ankle joints (Derrick et al., 1998; Nigg & Liu, 1999). It has been suggested that the repetitive attenuation of impact forces may contribute to overuse injuries (Collins & Whittle, 1989; Gill & O'Connor, 2003a; Gill & O'Connor, 2003b; MacLellan & Vyvyan, 1981; Voloshin, Wosk, & Brull, 1981).

Impulsive loading on the body during running can be assessed by measuring ground reaction forces and accelerations caused by impact (Mizrahi et al., 2000). Body segment acceleration is dependent on the magnitude of the ground reaction force (GRF) and the damping effects of the body's passive and active shock absorbers (Derrick et al., 1998). A primary objective of modern running shoes is to attenuate impact forces, with the goal to minimize running related overuse injuries that stem from the repetitive application of these forces. Running shoes have been shown to reduce tibia accelerations (McNair & Marshall, 1994), yet are associated with higher rates of loading and impact peak forces (Lieberman et al., 2010). It has also been suggested that running shoes may limit proprioceptive feedback and hence lead to an increase in running related overuse injuries (Robbins & Hanna, 1987).

Most individuals contact the ground with their heel when running in traditional running shoes that have an elevated heel. In heel strike runners, a transient impact force is generated when the heel contacts the ground. This impact transient typically has a high rate and magnitude of loading and is thought to contribute to the high incidence of running related injuries (McNair & Marshall, 1994). The forefoot strike pattern that is adopted by most individuals when running barefoot may have important implications for attenuating impact forces. It has been suggested that the greater ankle compliance during forefoot running decreases the effective mass of the body that collides with the ground, resulting in reduced impact peaks and loading rates (Lieberman et al., 2010). While differences in GRF's between barefoot and shod running have been the focus of several studies, to the best of our knowledge, how wearing shoes and a forefoot strike position independently affect impact accelerations has yet to be evaluated.

Therefore, the aim of the present study was to investigate the effects of ankle position and running shoes on running kinetics and impact accelerations. We hypothesized 1) that both running shoes and a plantarflexed position in barefoot running would reduce the propagation of impact accelerations to the tibia and 2) heel strike running whether barefoot or shod would result in higher impact peak magnitudes. Knowledge of how shoes and ankle position at ground contact independently affect the propagation of impact accelerations and running kinetics may help to understand the mechanism of running related overuse injuries.

Methods

Experimental Protocol

Ten healthy, physically active heel strike runners [5 men and 5 women, age: 26 ± 7.3 yr; height: 1.74 ± 0.09 m; mass: 65.6 ± 10.2 kg] participated in this study. Subjects were required to perform a minimum of 30 minutes of physical activity at least 5 days a week and be free of musculoskeletal injury of the lower extremities or back. Subjects provided written informed consent prior to participating in the study. The University of Idaho's Institutional Review Board approved the protocol for this study (Appendix 1).

Subjects ran in three conditions: shod (SHOD), barefoot (BF) and barefoot while heel striking (BFHS). The testing session began with 5-10 minutes of easy running to allow subjects to warm-up and habituate to the runway. When switching between conditions subjects were allowed as much time as they desired to familiarize themselves with the condition.

Kinematics and Kinetics

3-dimensional motion analysis and GRF data were collected as subjects ran over a 15 m runway with a force plate (AMTI, Waterton, MA) embedded at 10 m. Ten strides from each subject were used to calculate averages for each condition. Sixteen retro-reflective markers were fastened with double-sided tape to the bilateral anterior and posterior superior iliac spines, lateral mid-thigh, lateral femoral epicondyle, lateral mid-shank, lateral malleolus, second metatarsal head and calcaneus according to the Modified Helen Hayes Marker set (Kadaba et al., 1990). For the shod running, heel and toe markers were placed on the shoes at the positions best projecting the anatomical landmarks. Height, weight, leg length, and widths of the ankles and knees were measured for appropriate anthropometric scaling. 3-dimensional marker positions were captured at 250 Hz via a Vicon MX motion analysis system (Vicon, Oxford Metrics Ltd., UK) and filtered using a Woltring filtering routine with a predicted mean square error value of 4 mm^2. The three orthogonal components of the GRF data were recorded at 1000 Hz from the force plate in synchrony with the motion capture data. Force plate data were low-pass filtered at 30 Hz using a second-order Butterworth filter before being down-sampled and combined with the motion capture data. Joint kinematics and kinetics were computed via Vicon Plug-In Gait.

Impact peak magnitude was measured as the first observable peak in the vertical GRF. If the impact peak was absent, a value was not recorded. Loading rates were calculated as the change in force divided by the change in time between 20 and 80% of the period from ground contact to impact peak (following Milner et al., 2006).

Impact accelerations were measured from accelerometers placed on the antero-lateral surface of the distal lower leg and the lateral surface of the forehead. Lightweight biaxial accelerometers (Freescale Semiconductor, Austin, TX; model: MMA3202KEG) were mounted to a small piece of balsa wood with epoxy resin. Each accelerometer had a minimum 50-g range and 20 mV/g sensitivity. The combined mass of the accelerometer, balsa wood and epoxy was less than 3 g. The balsa wood mounted accelerometers were secured as firmly as possible to the leg with coban wrap and to the head with an elastic band. One axis of the accelerometer was oriented with the longitudinal axis of the tibia and the second axis was oriented with the direction of travel. This method of attachment has previously been shown to appropriately and reliably measure impact accelerations (Mizrahi, et al., 1997; Voloshin, Mizrahi, Verbitsky, & Isakov, 1998; Verbitsky, Mizrahi, Voloshin et al., 1998). Accelerometer data were collected at 1000 Hz via a Biometrics DataLOG MWX8 telemetered data acquisition device (Biometrics Ltd., Ladysmith, VA) simultaneously with motion capture and GRF data. Resultant accelerations were calculated from the two accelerometer axes, as this provides a better estimate of shock than a single axis (Lafortune, 1991). Peak resultant accelerations were measured for each analyzed stride and averaged across trials and subjects for each running condition.

Statistics

Statistical differences in the kinetic and kinematic parameters were determined using repeated-measures ANOVA in SPSS (IBM, Armonk, NY). When a significant effect was identified, a post hoc Bonferroni pairwise comparison was performed to determine which conditions were significantly different. Statistical significance was defined as $p<0.05$.

Results

Spatio-Temporal Parameters

There were statistically significant differences in stride length between the BF and SHOD conditions ($p = 0.038$) and BFHS and SHOD conditions ($p = 0.018$). There were no significant differences in running velocity between any of the conditions (Table 4.1).

Accelerometer

Figure 4.1 shows a representative example of the resultant accelerations from the head and leg accelerometers, simultaneously with the vertical GRF, for each condition. Peak resultant acceleration at the tibia was 11.32 ± 1.48, 13.55 ± 1.51 and 11.27 ± 1.73 g for the BF, BFHS and SHOD conditions respectively. There was a statistically significant difference in peak resultant tibia acceleration between the SHOD and BFHS conditions ($p=0.005$). Peak resultant acceleration at the head was 2.44 ± 0.71 g, 2.73 ± 0.97 and 2.46 ± 0.85 g for the

BF, BFHS and SHOD conditions respectively. There were no significant differences for peak resultant head accelerations between any conditions.

Kinematics

In general there was little difference in lower extremity kinematics between the three conditions (Table 4.2). However, there were significant differences in sagittal plane ankle angle at ground contact between the BF and SHOD and BF and BFHS conditions ($p<0.001$). In the BF condition, individuals contacted the ground in a more plantarflexed position; whereas in the BFHS and SHOD conditions, individuals contacted the ground in a dorsiflexed position. There was also significant difference in peak sagittal plane hip angle between the SHOD and BFHS conditions ($p = 0.040$) and BF and BFHS conditions ($p = 0.022$).

Kinetics

Select kinetic parameters are presented in Table 4.3. Impact peaks were present on 67% of BF trials, 96% of BFHS trials and 79% of SHOD trials. There were significant differences in impact peak magnitude between the BF and SHOD conditions ($p=0.004$) and BF and BFHS conditions ($p=0.005$). There was a statistically significant difference in peak vertical GRF between BFHS and SHOD conditions ($p=0.034$). There were no significant differences in loading rate, peak horizontal GRF, peak medio-lateral GRF or joint moments between any of the conditions.

Discussion

The goal of this study was to investigate the effects of ankle position and running shoes on running kinetics and impact accelerations. The results presented here support our hypotheses that 1) both running shoes and a plantarflexed position in barefoot running reduce tibia impact accelerations and 2) heel strike running whether barefoot or shod results in higher impact peak magnitudes. The results of this study are consistent with previous studies that have shown that running shoes decrease tibia impact accelerations (McNair & Marshall, 1994; Mizrahi et al., 2000). However, contrary to McNair and Marshall (1994), our results show that running barefoot reduced impact acceleration magnitudes to the level seen with running shoes. This difference could be due to the amount of plantarflexion at ground contact, as our BFHS condition was associated with greater impact accelerations than the BF or shod conditions. While McNair and Marshall (1994) report a more plantarflexed position in the barefoot condition, it is possible that their subjects adopted a more dorsiflexed position at ground contact than those in the present study, which could explain the greater impact accelerations.

In the present study, GRF impact peak magnitudes were similar in the shod and BFHS condition, however, the shod condition was associated with reduced tibia impact accelerations. GRF's are a measure of the force applied to the ground by the body and are frequently used as a proxy for forces transmitted to the skeletal system. However, the sole of the foot is the only structure that receives these loads (Derrick, 2008). The combination of GRF and accelerometer data used in the present study allowed for the transmission of impact acceleration to be evaluated. The results presented here suggest that in the shod condition the midsole of the running shoe helps to dampen some of the impact so the full force does not reach the tibia, which is evident in reduced impact peak magnitudes. However, in the BFHS condition more of the impact was transmitted to the musculoskeletal system as shown by increased tibia accelerations.

Consistent with previous studies, we have shown that the barefoot condition was associated with reduced impact peak magnitudes (Altman & Davis, 2012; Jenkins & Cauthon, 2011). The subjects in the present study also exhibited decreased tibia impact accelerations when running barefoot. These kinetic changes can likely be explained by a reduction in stride length and/or plantarflexed position at ground contact. Stride length is important to consider when evaluating impact attenuation, as it has been previously shown that reducing stride length results in decreased peak impact accelerations and impact attenuation (Derrick et al., 1998; Hamill et al., 1995). In the present study individuals ran with a significantly greater stride length in the shod condition. While longer stride lengths are typically associated with greater impact accelerations, the shod condition saw a reduction in impact accelerations as compared to the BFHS condition. The reduced impact acceleration, despite an increase in stride length, further supports the notion that the running shoe midsole helps to dampen impact forces. We have also shown that individuals ran at a similar stride length in both the BF and BFHS conditions, yet the BFHS condition resulted in greater impact accelerations. This would suggest that the plantarflexed position at ground contact helps to reduce impact accelerations. It has been proposed that contacting the ground on the forefoot/midfoot allows runners absorb impact through compression of the medial longitudinal arch of the foot, eccentric contraction of the triceps surae, and stretch of the Achilles tendon (Lieberman et al., 2010).

The results presented here may have particular importance in terms of knee injury risk, as accelerations of the tibia will be transferred directly to the knee, likely resulting in higher joint contact forces and/or loading of passive structures. Our results suggest that changing to a forefoot strike pattern or wearing running shoes will help to reduce transient tibia acceleration, thus, potentially reducing knee loading. Future prospective studies should be aimed at evaluating the relationship between impact induced accelerations and injury risk.

We observed no difference in resultant head accelerations between the different running conditions. This is consistent with previous studies that have shown little effect of gait changes on head accelerations. Significant differences in the magnitude of head accelerations have been reported with changes in stride length, but the magnitude of these accelerations remained considerably less than what was observed at the tibia (Derrick et al., 1998; Hamill et al., 1995). These findings indicate that, despite the magnitude of impact accelerations experienced at the lower extremity, active and passive structures attenuate shock before it reaches the head. It has been proposed that a number of anatomical structures have evolved to attenuate shock so that vision remains stable and the brain does not experience great shock magnitude (Bramble & Lieberman, 2004; Lieberman & Bramble, 2007).

Certain limitations should be considered when interpreting the findings of the present study. First, subjects wore their personal running shoes rather than standardized footwear. Previous studies have shown varied results in terms of differences in impact peak magnitude and loading rate with different shoes (Yan et al., 2012). Second, soft tissue movement can distort skin mounted accelerometer signals. However, we made every attempt to minimize soft tissue movement by firmly securing and using lightweight accelerometers (Forner-Cordero et al., 2008). Additionally, changing limb orientation will influence accelerometer data, therefore, we calculated resultant accelerations to better estimate the magnitude of lower extremity shock (Lafortune, 1991). Lastly, it is important to note that there is considerable individual variation in terms of kinetic and kinematic changes associated with different running conditions.

In conclusion, we have shown that both BF and shod running result in reduced impact related tibia accelerations. Shod running was associated with increased GRF impact peak magnitude, but it appears that the midsole of running shoes helps to attenuate impact forces, thus decreasing the amount of force that is transmitted through the musculoskeletal system. Barefoot running exhibited a similar decrease in impact accelerations, as well as decreased impact peak magnitude, which appears to be due to a decrease in stride length and/or a more plantarflexed position at ground contact. Evaluating both GRFs and impact related accelerations provides valuable information about the transmission of impact forces to the musculoskeletal system, and both should be considered when evaluating the potential for injury prevention.

Table 4.1. Stride length and velocity

	BF	BFHS	SHOD
Stride Length (m)	**2.13 (0.15)**$^{\wedge}$	**2.18 (0.17)**$^{\wedge}$	**2.25 (0.19)**$^{*\#}$
Velocity (m/s)	2.97 (0.19)	3.10 (0.25)	3.09 (0.25)

Data are mean (standard deviation). Significant differences are indicated in bold. * indicates a significant difference to BF. # indicates a significant difference to BFHS. ^ indicates a significant difference to SHOD. p<0.05.

Table 4.2. Lower extremity joint angles at ground contact and peak values.

		BF	BFHS	SHOD
Ankle Dorsiflexion (^{o})	At Contact	**-12.1 (7.0)**$^{\wedge\#}$	**7.4 (3.1)***	**8.9 (5.6)***
	Peak	30.0 (7.7)	30.4 (7.3)	29.0 (6.0)
Ankle Adduction (^{o})	At Contact	2.06 (6.1)	-4.2 (7.6)	-0.5 (7.2)
	Peak	7.3 (4.4)	11.4 (7.1)	9.6 (6.8)
Ankle Internal Rotation (^{o})	At Contact	-7.7 (10.5)	-13.2 (9.2)	-3.28 (14.8)
	Peak	2.2 (10.7)	3.1 (9.5)	5.2 (10.4)
Knee Flexion (^{o})	At Contact	8.8 (5.4)	3.2 (9.1)	6.3 (7.0)
	Peak	37.5 (6.3)	34.4 (4.3)	31.6 (6.1)
Knee Varus (^{o})	At Contact	5.9 (6.6)	6.0 (9.5)	3.4 (5.7)
	Peak	15.4(9.9)	22.2 (16.1)	19.5 (8.7)
Knee Internal Rotation (^{o})	At Contact	-23.1 (17.6)	-27.6 (17.1)	-30.8 (13.0)
	Peak	1.6 (7.3)	4.1 (11.5)	4.9 (12.8)
Hip Flexion (^{o})	At Contact	36.4 (11.9)	35.2 (11.2)	36.8 (12.1)
	Peak	**36.8 (12.0)** $^{\#}$	**42.4 (10.4)**$^{*\wedge}$	**38.3 (12.6)**$^{\#}$
Hip Adduction (^{o})	At Contact	4.6 (6.2)	5.2 (6.6)	5.6 (5.4)
	Peak	12.7 (8.0)	10.7 (8.5)	11.7 (5.0)
Hip Internal Rotation (^{o})	At Contact	21.9 (18.1)	24.6 (17.2)	25.4 (16.8)
	Peak	29.4 (14.9)	33.2 (13.9)	32.4 (10.5)

Data are mean (standard deviation). Significant differences are shown in bold * indicates a significant difference to BF. # indicates a significant difference to BFHS. ^ indicates a significant difference to SHOD. p<0.05.

Table 4.3. Select kinetic parameters.

	BF	BFHS	SHOD
Impact Peak (BW)	**1.58 (0.21)^**	1.81 (0.25)	**1.91 (0.21)***
Loading Rate (BW/s)	135.7 (38.2)	160.8 (33.6)	148.4 (48.9)
Peak vertical GRF (BW)	2.29 (0.26)	**2.23 (0.19)^**	**2.31 (0.23)#**
Peak anterior-posterior GRF (BW)	0.37 (0.08)	0.35 (0.06)	0.37 (0.06)
Peak medio-lateral GRF (BW)	0.08 (0.04)	0.07 (0.03)	0.08 (0.04)

Data are mean (standard deviation). Significant differences are indicated in bold. * indicates a significant difference to BF. # indicates a significant difference to BFHS. ^ indicates a significant difference to SHOD. p<0.05. BW = body weight.

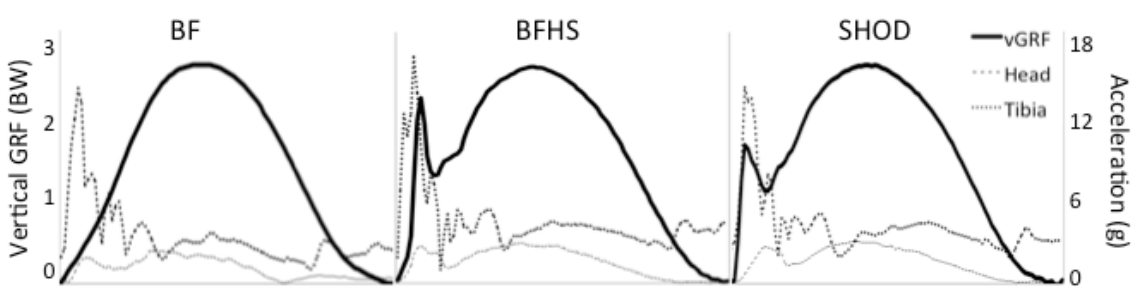

Figure 4.1. Vertical GRFs and accelerations. Typical vertical GRF in body weights (BW) and resultant head and tibia acceleration profiles for the three running conditions.

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Chapter 5. Elastic Energy Storage And Return In Forefoot And Rear-Foot Strike Running

Melissa Thompson, Anne Gutmann, Jeff Seegmiller and Craig P. McGowan

Introduction

Tendon and muscle in the lower extremities store and return elastic energy during locomotion, which greatly reduces muscular energy requirements (Alexander, 1984; Cavagna et al., 1977). Elastic energy storage is especially evident in running, where tendon and muscle are stretched and store strain energy as the body is decelerated during the first half of stance. Muscle and tendon then recoil in the second half of stance and stored strain energy helps to raise and accelerate the body into the aerial phase (Alexander, 1991). Thus, running can be explained by the spring mass model where kinetic and gravitational potential energy fluctuate in phase and are stored as elastic energy at ground contact (Blickhan, 1989). It has been estimated that efficiency in human running can be as high as 40-50% due to this conversion of gravitational potential and kinetic energy to elastic energy (Cavagna et al., 1964).

While both tendon and muscle store strain energy, muscle contributions to total mechanical work are minor as compared to tendon contributions (Alexander & Bennet-Clark, 1977). The Achilles tendon is a primary mechanism for storing elastic energy during running. Ker *et al*. (1987) estimated that the human Achilles tendon stores as strain energy 35% of the kinetic and potential energy that is lost as the body is decelerated. The arch of the human foot also stores a substantial amount of elastic energy. As the arch flattens during ground contact an estimated 17% of the kinetic and potential energy lost during ground contact is stored as strain energy in the stretched ligaments of the arch of the foot (Ker et al., 1987).

The amount of elastic energy that can be stored in a tendon is dependent on the tendon dimensions and change in tendon length (Alexander & Bennet-Clark, 1977). The magnitude of the change in tendon length is dependent on the force exerted by the muscle and external loading. In terms of the Achilles tendon, changes in gastrocnemius and soleus activation, as well as alterations in the ground reaction force (GRF) magnitude and orientation influence the amount of elastic energy storage. It has been suggested that contacting the ground with a forefoot strike (FFS) pattern when running results in greater elastic energy storage as compared to contacting the ground with a rear-foot strike (RFS) pattern (Adrigò et al., 1995; Lieberman et al., 2010). Adrigò *et al*. (1995) report higher whole body mechanical work due to greater external work in FFS running, yet a similar VO_2 for FFS and RFS running. They interpret this finding to suggest that FFS running does not require a substantial increment in VO_2 because increased elastic energy storage offsets the greater work requirement.

Evidence for elastic energy contributions during running comes from measurements of metabolic cost that are substantially lower than what would be predicted based on total mechanical work requirements. While measures of high efficiency indicate elastic energy storage, this method does not allow for energy contributions from specific sources, e.g. the Achilles tendon, to be quantified. In order to calculate elastic energy storage it is necessary to measure tendon strain. *In vivo* measurements of tendon strain can be obtained from implanted sensors (e.g. buckle transducers and implanted load cells) or from medical imaging (e.g. ultrasound and sonoelastography) (Bey & Derwin, 2012; Fleming & Beynnon, 2004). However, the use of these techniques is limited due to invasiveness and uncertain accuracy especially with dynamic movements (Bey & Derwin, 2012). An alternative method to quantify elastic energy storage and return in specific tendons is to use detailed musculoskeletal models with individual musculotendon actuators and forward dynamical simulations (Neptune et al., 2004; Zajac, et al., 2003). Thus, the purpose of the present study was to use a detailed musculoskeletal model and forward dynamics simulation to quantify the amount of elastic energy stored in the Achilles tendon during FFS and RFS running.

Methods

Experimental Data

Data for this study was collected in conjunction with Chapter 2. We collected data from ten naturally FFS individuals [5 men and 5 women, age: 28 ± 5.9 yr; height: 1.71 ± 0.08 m; mass: 70.8 ± 10.3 kg] and ten naturally RFS individuals [5 men and 5 women, age: 29 ± 6.0 yr; height: 1.70 ± 0.09 m; mass: 65.3 ± 8.6 kg]. It was required that subjects perform a minimum of 30 minutes of physical activity at least 5 days a week and be free of musculoskeletal injury of the lower extremities or back. Foot strike pattern was determined by visual observation. FFS was defined as ground contact on the ball of their foot (metatarsal heads) and RFS was defined as ground contact at the heel (calcaneus). The

University of Idaho's Institutional Review Board approved the protocol for this study (Appendix 1).

3-dimensional motion analysis and GRF data were captured as subjects ran over a 15 m runway with a force plate (AMTI, Waterton, MA) embedded at 10 m. Subjects were instructed to run with their preferred running gait (i.e., self-selected stride length and velocity) in both barefoot and shod conditions. 10 strides were analyzed for each condition.

16 retro-reflective markers were affixed with double-sided tape to specific landmarks according to the Modified Helen Hayes Marker set (Kadaba et al., 1990). 3-dimensional positions of each marker were captured at 250 Hz via a Vicon MX motion analysis system (Vicon, Oxford Metrics Ltd., UK). Marker trajectory data were filtered using a Woltring filtering routine with a predicted mean square error value of 4 mm^2. The three orthogonal components of the ground reaction force (GRF) data were recorded at 1000 Hz from the force plate in synchrony with the motion capture data. Force plate data were low-pass filtered at 30 Hz using a second-order Butterworth filter before being down-sampled and combined with the motion capture data.

EMG activity of right gluteus maximus, biceps femoris, vastus lateralis, rectus femoris, lateral gastrocnemius, soleus and tibialis anterior muscles were collected simultaneously with GRF and motion data. EMG data was sampled at 1000 Hz during using a Biometrics DataLOG MWX8 telemetered EMG system (Biometrics Ltd., Ladysmith, VA). The Biometrics bipolar SX230 preamplifier surface electrodes (diameter 10 mm and inter-electrode distance of 20 mm) were placed according to Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) recommendations. EMG data was processed via a custom program implemented in LabVIEW 2013 (National Instruments, Austin, TX). The raw EMG data was band-pass filtered from 10 to 500 Hz using a zero-lag, second-order digital Butterworth filter and rectified. For each subject, 5 consecutive strides from each trial were used to calculate mean EMG waveforms for each muscle. The EMG data was integrated with respect to time and then normalized by dividing by the peak EMG amplitude.

Musculoskeletal Model

To generate simulations of the experimental data described above, a detailed musculoskeletal model was adapted from existing models using SIMM (MusculoGraphics, Inc., Santa Rosa, CA). The model was composed of rigid segments consisting of a trunk and two legs, with each leg having a thigh, shank, patella, rear-foot, mid-foot and toes. The dimensions of the model represented a male subject with a height of 180 cm and a mass of 75 kg. Musculoskeletal geometry was based on Delp *et al*. (1990) and segment masses and inertial properties were determined using regression equations (Chandler et al., 1975; Clauser et al., 1969). The model had 13 degrees of freedom including flexion/extension at the hip, knee, ankle, mid-foot and toes of each leg, and horizontal and vertical translation and pitch rotation at the trunk. The trunk segment included the mass and inertial characteristics of the pelvis, torso, head and arms. The hip and ankle joints were modeled as frictionless revolutes. The tibiofemoral joint was modeled with a moving center-of-rotation for flexion-extension specified as functions of knee flexion angle (Delp et al., 1990). Passive torques were applied at the hip, knee and ankle joints based on Davy and Audu (1987) to model the forces applied by ligaments and other passive joint structures. The patella was constrained to move along a prescribed trajectory relative to the femur as a function of knee flexion angle (Delp et al., 1990). Contact between the foot and ground was modeled by 30 independent visco-elastic elements with coulomb friction, each attached to the foot segment in locations that described the exterior plantar surface of a shoe. Each element allowed for deformation perpendicular to the floor and represented the mechanical properties of the shoe sole and underlying soft tissue. The anterior–posterior and vertical force calculations as well as the determination of shoe specific parameters are presented in detail in Neptune *et al*. (2000).

The model was driven by 25 Hill-type musculotendon actuators (Fig. 5.1) per leg combined into 11 functional groups based on anatomical classifications with muscles in each group receiving the same excitation pattern. The muscle groups were defined as PSOAS (iliacus, psoas), GMAX (gluteus maximus, adductor magnus), VAS (3-component vastus), HAMS (medial hamstrings, biceps femoris long head), SOL (soleus), BFsh (biceps femoris short head), GAS (medial and lateral gastrocnemius), RF (rectus femoris) and TA (tibialis anterior) (Fig. 5.2). Individual muscle excitations were modeled using bimodal patterns characterized by an onset, duration and magnitude. The contraction dynamics were governed by a Hill-type model formulation (Schutte et al., 1993) and the activation dynamics were modeled by a first-order differential equation (Raasch et al., 1997), with activation and deactivation time constants of 50 and 65 ms, respectively.

Dynamic Optimization

Running simulations of a complete gait cycle were generated for each condition from heel-strike to the following ipsilateral heel-strike. A simulated annealing algorithm (Goffe et al., 1994) was used to fine-tune the muscle excitation patterns and initial joint velocities to minimize the difference between the simulated and experimental kinetic and kinematic data for both FFS and RFS running. Muscle stress and activation was minimized to ensure equal recruitment of agonist muscles and limit co-contraction of antagonist muscles. The timing of muscle excitations (onset and offset) can be constrained so that muscle activations occur in periods of the gait cycle consistent with experimental EMG data. The tracked variables included the left and right hip, knee and ankle angles, trunk translations (horizontal and vertical) and the horizontal and vertical ground reaction forces. The optimizations were continued until all tracking variables were within two standard deviations of the group averaged experimental data.

Muscle & Tendon Mechanical Power & Work

To examine differences in elastic energy storage and return between the FFS and RFS simulations, muscle fiber, tendon and muscle tendon unit (MTU) mechanical power and work was analyzed for the lateral gastrocnemius (LGAS), medial gastrocnemius (MGAS) and soleus (SOL) muscles. We analyzed each muscle separately, but because they share a common tendon, we also pooled the data to reflect the overall function of the plantar flexors and Achilles tendon. Muscle fiber, tendon and MTU power was calculated as the product of the corresponding force and velocity at each instant in time throughout the gait cycle. Positive, negative, total and net mechanical work done by the muscle fibers, tendon and MTU were calculated as the time-integral of the corresponding power. Mechanical work was analyzed for the stance phase and a complete stride. Additionally, the relative tendon contribution to total MTU work was calculated.

Results

Tracking

The kinematic and kinetic data from the FFS simulation replicated the experimental data almost always within ± 2 S.D. of the group-average (Fig. 5.3). Kinematic and kinetic data from the RFS simulation do not emulate the experimental data as closely (Fig. 5.4). Specifically, a large impact peak is present in the horizontal GRF of the RFS simulation and the timing of GRF's in the RFS simulation occur later than the experimental data. Further, in the RFS simulation, knee angle tracking deviates from the ± 2 S.D. bandwidth at several points.

EMG

In several muscle groups simulation EMG differed considerably from experimental EMG (Fig. 5.5). This indicates the need for further optimization.

Work & Power

During stance energy was first stored in the Achilles tendon (negative power) and then recovered (positive power) in both the FFS and RFS simulation (Fig. 5.6). The FFS simulation exhibited greater total tendon work, less total muscle work and slightly less MTU total work than the RFS simulation (Table 5.1). The FFS simulation showed a substantially greater relative tendon contribution to total MTU work (FFS = 27.12%, RFS = 19.48%).

Analysis of mechanical work for the total stride generally followed the same trends as what was seen in the stance phase (Table 5.2). For the complete stride the FFS simulation exhibited greater total tendon work, less total muscle work and slightly less MTU total work than the RFS simulation (Table 5.1). The FFS simulation showed a substantially greater relative tendon contribution to total MTU work (FFS = 45.51%, RFS = 40.84%).

The remainder of this results section will focus on the stance phase (Table 5.1), as this is were elastic energy storage and return primarily occurs. Positive work was greater in the tendon and MTU of the individual and pooled plantar flexors of the FFS simulation than the RFS simulation. The FFS simulation exhibited greater positive muscle work for the MGAS, LGAS and pooled plantar flexors, but the RFS simulation exhibited greater positive muscle work in the SOL.

Similar amounts of negative tendon work were observed in the individual plantar flexors and pooled plantar flexors of the FFS and RFS simulations. The FFS and RFS simulations exhibited similar amounts of negative muscle and MTU work in the SOL, but greater negative work was observed in the RFS simulation in the LGAS, MGAS and pooled plantar flexors.

Net tendon work was generally similar for the FFS and RFS simulations, and low due to similar magnitudes of tendon positive and negative work. Total tendon work was greater in the FFS simulation for the individual and pooled plantar flexors. The greater total tendon work was generally due to greater positive and negative tendon work in the FFS simulation. Net muscle work was greater in FFS simulation for the individual and pooled plantar flexors. Total muscle work was generally greater in the individual and pooled plantar flexors of the RFS simulation. This difference was due primarily to generally greater negative work in the RFS simulation. Net MTU work was negative and similar between the FFS and RFS simulations for the SOL. For the FFS simulation net MTU work was considerably greater (and positive) than the RFS simulation for the MGAS, LGAS and pooled plantar flexors. Total MTU work was similar between the FFS and RFS simulations for the SOL and LGAS and greater in the RFS simulation for the MGAS and pooled plantar flexors. The greater total MTU work in the pooled plantar flexors was due primarily to greater total muscle work

Discussion

The purpose of the present study was to use a detailed musculoskeletal model and forward dynamics simulations to quantify the amount of elastic energy stored in the Achilles tendon during FFS and RFS running. This technique allowed for estimations of muscle and tendon work that would otherwise only be obtainable by invasive measures. It is not possible to make any meaningful conclusions based on the results presented here, as the RFS simulation is not accurately tracking experimental data and simulation EMG differs substantially from experimental EMG. Optimization of the both the FFS and RFS simulations is presently ongoing so that better tracking of the experimental data can be obtained. Ultimately, simulation muscle excitation patterns will be validated with experimental EMG.

Our current results suggest that FFS running is associated with a greater return of elastic energy during stance, as is evidenced by greater absolute positive work in the Achilles tendon and a greater relative tendon contribution to total MTU work. If this relationship remains with better simulation data, it would indicate that FFS running does result in greater elastic energy storage in the Achilles. This would be consistent with the findings of Adrigò *et al.* (1995) who report that greater elastic energy storage in FFS running, which offsets increased whole body mechanical work and results in similar metabolic cost to RFS running.

The musculoskeletal model and forward dynamics simulations used in the present study opens up many interesting lines of research. For example the model could be used to examine differences in segment and trunk energetics between FFS and RFS running. This technique would also allow estimation of differences in the function of the muscle SEE component between conditions. Results of SEE function during running could also be compared to measures of SEE function obtained via ultrasound (Lichtwark et al., 2007).

It should be noted that this study used the same model for both the FFS and RFS simulations. Yet, the experimental data is based off two distinct groups of runners who naturally differ in terms of ground contact. To date there is no clear indication for what leads to the difference in ankle position ground contact. It has been shown that forcing individuals to switch ground contact techniques results in mechanical differences from individuals who habitually run in that position (Stearne et al., 2014).

The limitations associated with analyzing muscle and tendon function using the present musculoskeletal model have been previously discussed in detail (Neptune et al., 2001; Neptune et al., 2004; Zajac et al., 2003). A primary limitation is that a number of assumptions are made due to lacking experimental data. Sensitivity analysis has shown that biarticular muscle moment arms and the ground contact model are two areas where more precise data would be useful (Zajac et al., 2003).

Table 5.1. Stance mechanical work.

		Soleus		Medial Gastrocnemius		Lateral Gastrocnemius		Pooled Plantar Flexors	
		FFS	**RFS**	**FFS**	**RFS**	**FFS**	**RFS**	**FFS**	**RFS**
Tendon	**+ Work**	10.22	8.68	10.15	7.23	4.47	2.81	24.84	18.72
	- Work	-12.95	-10.07	-6.65	-8.54	-3.45	-3.28	-23.05	-21.89
	Net Work	-2.73	-1.39	3.49	-1.32	1.02	-0.46	1.78	-3.17
	Total Work	23.17	18.76	16.81	15.77	7.91	6.09	47.89	40.62
Muscle	**+ Work**	7.8	8.33	8.71	3.55	7.06	2.35	23.57	14.23
	- Work	-24.61	-26.76	-0.06	-10.09	-0.02	-5.37	-24.69	-42.22
	Net Work	-16.81	-18.42	8.65	-6.55	7.05	-3.02	-1.11	-27.99
	Total Work	32.41	35.09	8.77	13.64	7.08	7.73	48.26	56.46
MTU	**+ Work**	17.78	16.89	17.79	10.56	10.57	5.02	46.14	32.47
	- Work	-37.32	-36.7	-5.64	-18.42	-2.51	-8.5	-45.47	-63.62
	Net Work	-19.55	-19.81	12.15	-7.86	8.06	-3.48	0.66	-31.15
	Total Work	55.1	53.6	23.42	28.98	13.08	13.53	91.60	96.11
Tendon contribution to MTU total work (%)		18.55	16.19	43.34	24.95	34.17	20.77	27.12	19.48

Stance phase mechanical work (units: J) done by the tendon, muscle and MTU of the SOL, MGAS and LGAS in the FFS and RFS simulations. Magnitudes of pooled plantar flexor muscle, tendon (Achilles) and MTU work are also provided, as are relative tendon contribution to total MTU work.

Table 5.2. Stride mechanical work.

		Soleus		Medial Gastrocnemius		Lateral Gastrocnemius		Total	
		FFS	**RFS**	**FFS**	**RFS**	**FFS**	**RFS**	**FFS**	**RFS**
Tendon	**+ Work**	11.61	8.68	10.19	9.33	4.63	4.17	26.43	22.18
	- Work	-16.87	-15.39	-7.59	-9.03	-3.64	-3.64	-28.1	-28.06
	Net Work	-5.25	-6.7	2.6	0.29	0.98	0.53	-1.67	-5.88
	Total Work	28.48	24.07	17.79	18.36	8.27	7.82	54.54	50.25
Muscle	**+ Work**	8.7	8.33	8.7	6.54	7.42	5.65	24.82	20.52
	- Work	-28.86	-30.85	-2.87	-10.75	-2.05	-6.5	-33.78	-48.1
	Net Work	-20.16	-22.51	5.82	-4.2	5.37	-0.84	-8.97	-27.55
	Total Work	37.56	39.18	11.58	17.29	9.48	12.15	58.62	68.62
MTU	**+ Work**	20.07	16.89	17.78	15.66	11.01	9.64	48.86	42.19
	- Work	-45.49	-46.11	-9.35	-19.57	-4.65	-9.96	-59.49	-75.64
	Net Work	-25.41	-29.22	8.43	-3.91	6.35	-0.31	-10.63	-33.44
	Total Work	65.56	63.01	27.14	35.23	15.67	19.6	108.37	117.84
Tendon contribution to MTU total work (%)		30.55	34.61	48.90	35.62	89.72	72.25	45.51	40.84

Complete stride mechanical work (units: J) done by the tendon, muscle and MTU of the SOL, MGAS and LGAS in the FFS and RFS simulations. Magnitudes of pooled plantar flexor muscle, tendon (Achilles) and MTU work are also provided, as are relative tendon contribution to total MTU work.

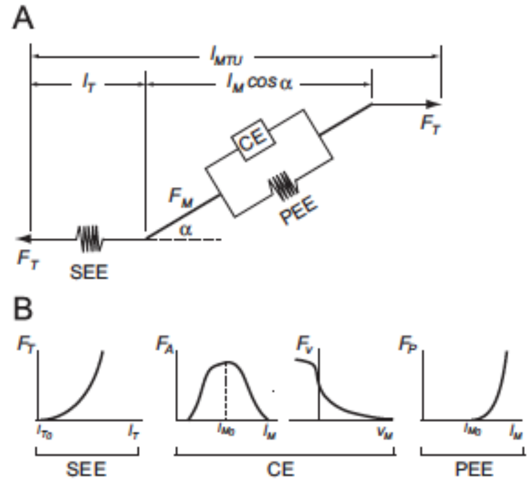

Figure 5.1. Muscle model. (A) Generalized Hill-type muscle model consisting of a contractile element (CE), parallel elastic element (PEE) and series elastic element (SEE) (Zajac & Gordon, 1989). (B) Muscle force is determined from the active force–length and force-velocity relationships in the contractile element and the non-linear spring properties of the passive elastic and series elastic elements.

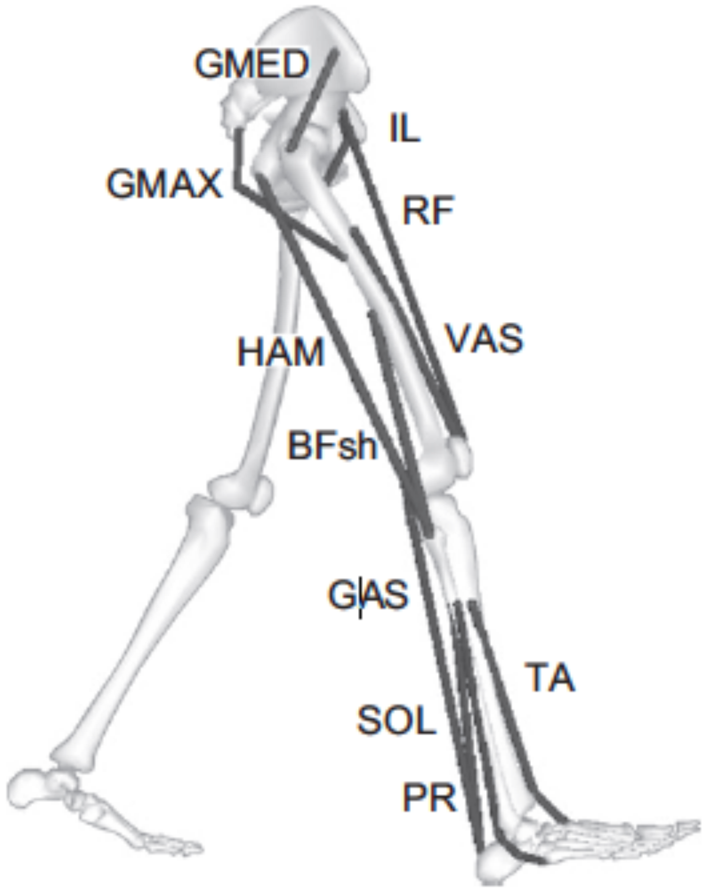

Figure 5.2. Musculoskeletal model. The musculoskeletal model consisted of rigid segments representing a trunk and two legs, each consisting of a thigh, shank, patella, rear-foot, mid-foot and toes. The 11 muscle groups were defined as IL (illiacus, psoas), GMAX (gluteus maximus, adductor magnus), GMED (gluteus medius), VAS (3-component vastus), RF (rectus femoris), HAM (medial hamstrings, biceps femoris long head), BFsh (biceps femoris short head), GAS (medial and lateral gastrocnemius), SOL (soleus, tibialis posterior, flexor digitorum longus, flexor hallucis longus), TA (tibialis anterior, extensor digitorum longus, extensor hallucis longus) and PR (peroneus longus).

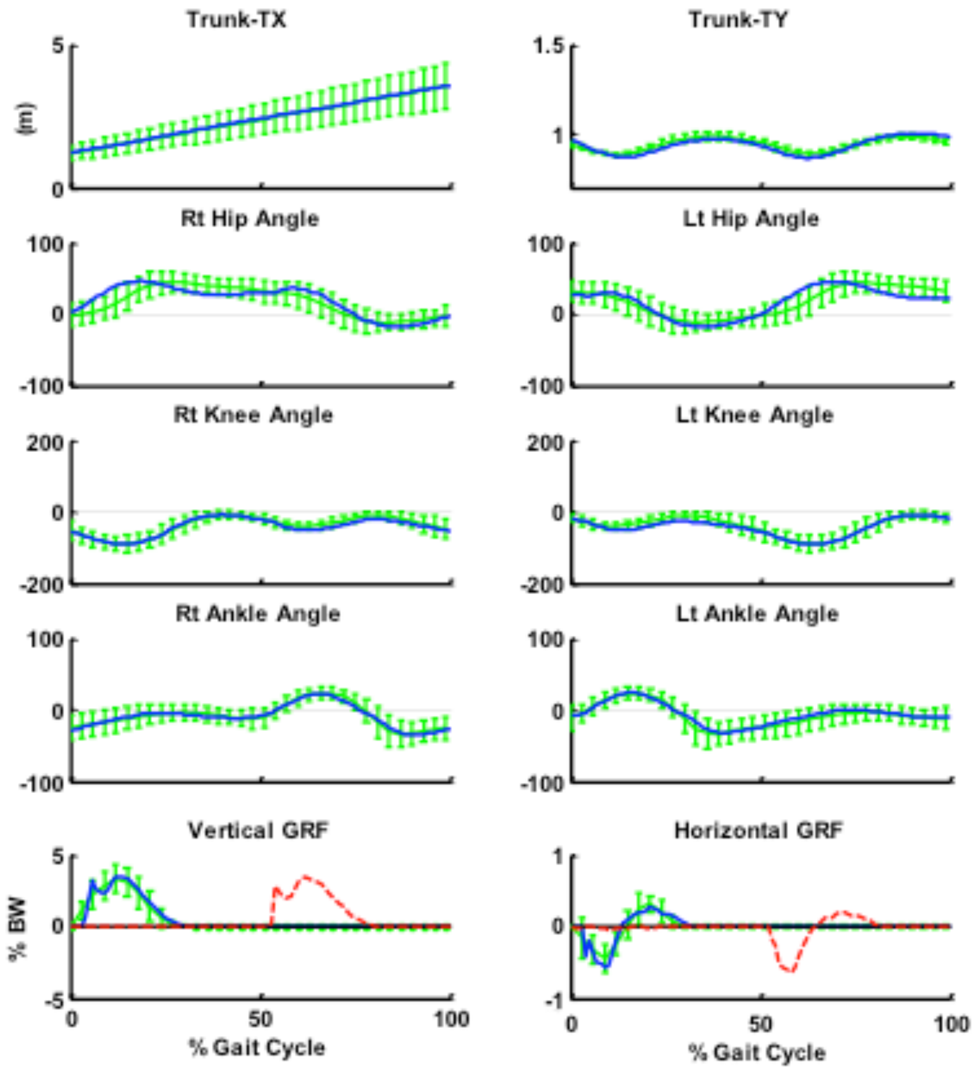

Figure. 5.3. Tracking data for the FFS Simulation. Sagittal (TX) and frontal plane (TY) trunk motion (units: meters); hip, knee and ankle joint angles for the right and left leg (units: degrees); and vertical (vGRF) and horizontal (hGRF) ground reaction forces (units: normalized to body weight). Experimental data is shown in green as mean $\pm$ 2 S.D., simulation data shown in solid blue line.

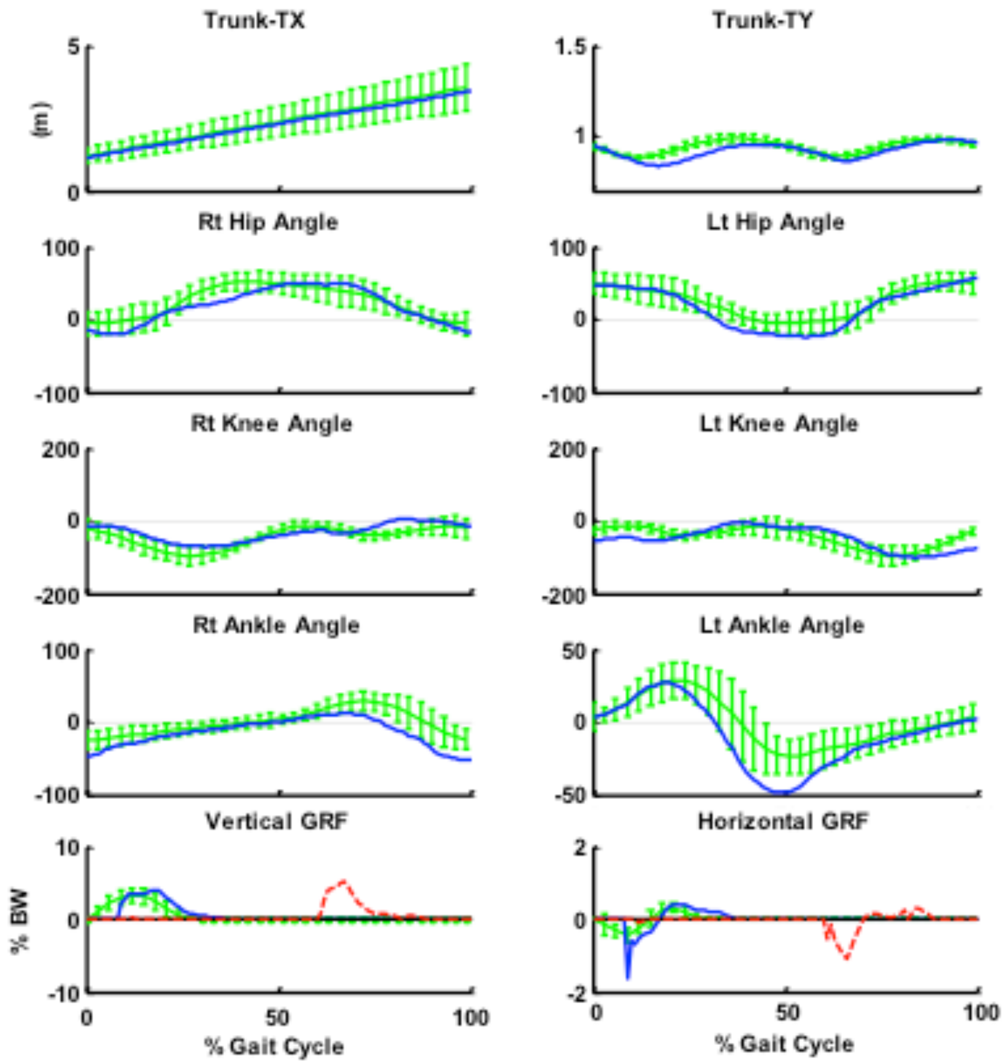

Figure. 5.4. Tracking data for the RFS Simulation. Sagittal (TX) and frontal plane (TY) trunk motion (units: meters); hip, knee and ankle joint angles for the right and left leg (units: degrees); and vertical (vGRF) and horizontal (hGRF) ground reaction forces (units: normalized to body weight). Experimental data is shown in green as mean $\pm$ 2 S.D., simulation data shown in solid blue line.

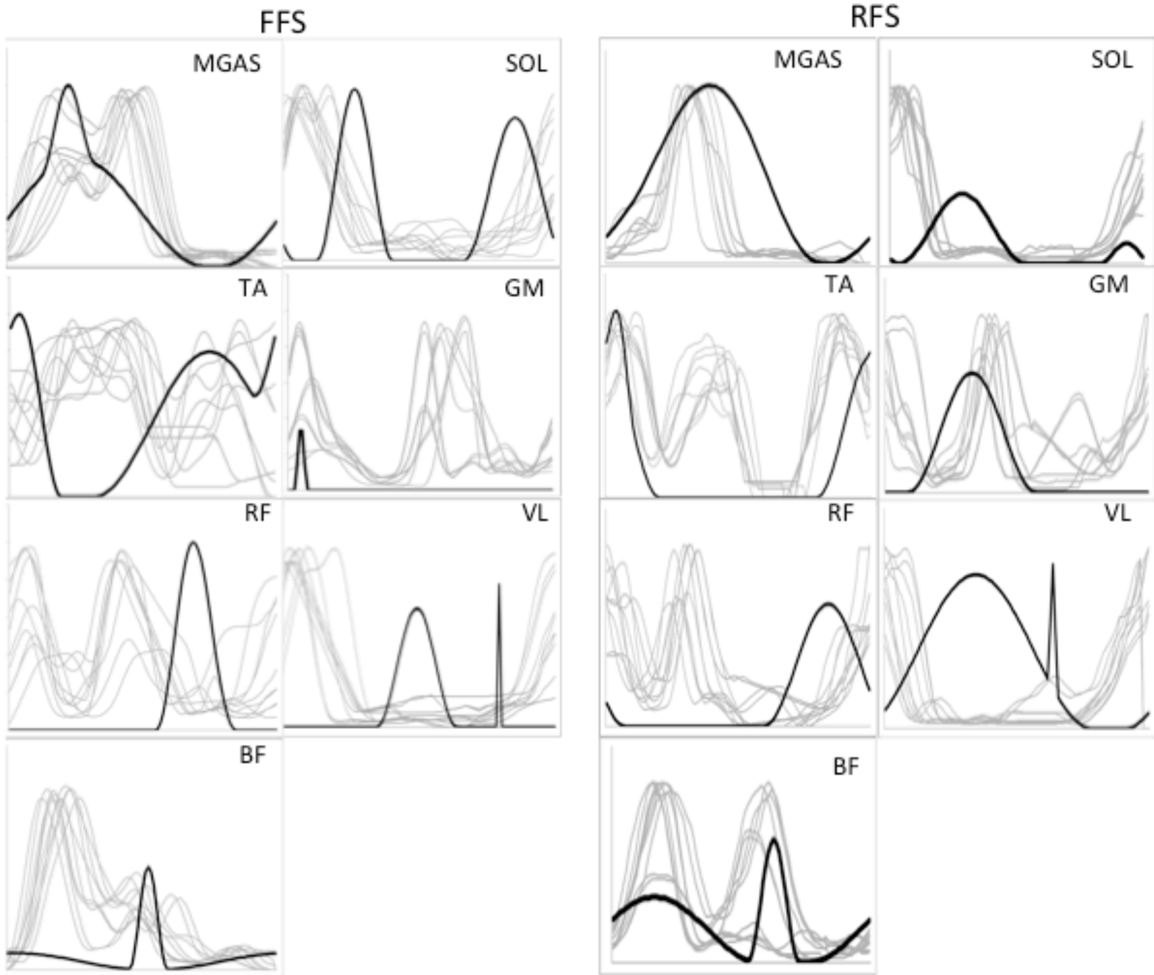

Figure 5.5. Comparison of experimental and simulation EMG data. Comparison of experimental and simulation EMG data for the FFS and RFS conditions. Single subject experimental data (grey) and simulation data (black). MGAS (medial gastrocnemius), SOL (soleus), TA (tibialis anterior), GM (gluteus maximus), RF (rectus femoris), VL (vastus lateralis), BFsh (biceps femoris).

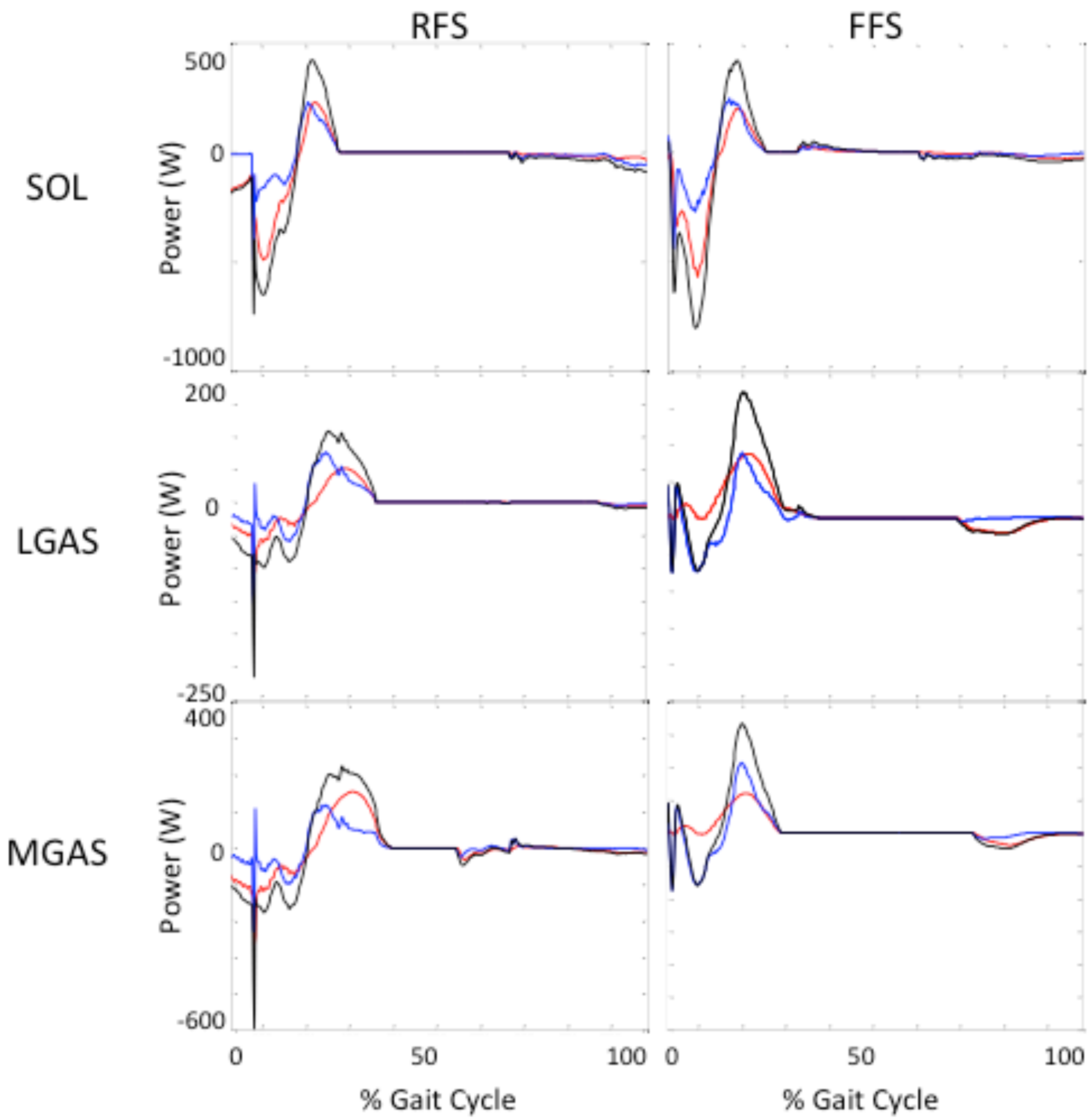

Figure 5.6. Mechanical power. Power stored and released by the SOL, LGAS and MGAS in the RFS and FFS simulations. MTU power (black) is the sum of muscle power (red) and tendon power (blue). Negative tendon power corresponds to energy stored in tendon and positive tendon power corresponds to energy released from tendon. Heel contact corresponds to 0%, ipsilateral heel contact occurs at 100%. Note different scales.

Chapter 5. Bibliography

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Chapter 6. Conclusion

The research presented here focused on understanding neural and mechanical factors related to barefoot and shod running. These studies integrated empirical biomechanical analyses with detailed musculoskeletal modeling and forward dynamics simulations to gain novel insight into the neuromechanics of barefoot and shod running. One of the complexities of analyzing gait is that changing one variable may result in cascade of subsequent changes. Therefore, the results presented do not offer a single "answer" regarding how to prevent running related injury or improve running performance, however, several important findings have been made:

1) Reductions in stride length, whether barefoot or shod, results in decreased GRFs and ankle and knee moments.
2) Barefoot running is similar to natural forefoot running.
3) Impact accelerations are reduced by wearing shoes or adopting a forefoot strike pattern when running barefoot.
4) Barefoot running itself may not be beneficial in terms of altering running mechanics. Rather, barefoot running acts as a trigger to reduce stride length and contact the ground with the forefoot, which both lead to reductions in mechanical factors related to injury risk.
5) It is important to independently examine factors such as foot-strike, stride-length and shoe condition, when evaluating how these interventions influence running mechanics.

Future Directions

While several important finding have been made through the course of these research projects, many more questions have been raised creating the possibility for further areas of exploration. First, future studies should be aimed at further understanding the role of different forms of sensory feedback to the control of locomotion. There is considerable evidence to suggest that different sensory factors influence gait, yet the specific role of different types of feedback has not been examined. Second, many potential gait modifications have been proposed, but it has yet to be determined if individuals are able to making lasting changes to their gait. Finally, long-term prospective studies are ultimately needed to assess the ability of interventions to reduce running related injury risk. Given the interaction of many gait variables multifactorial studies are likely needed to try to identify factors that influence running related injury risk.

Appendix 1

Institutional Review Board Approval Letter

University of Idaho

Office of Research Assurances
Institutional Review Board
PO Box 443010
Moscow ID 83844-3010

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To:	Dr. Craig McGowan
Cc:	Melissa Thompson
From:	Traci Craig, PhD Chair, University of Idaho Institutional Review Board University Research Office Moscow, ID 83844-3010
IRB No.:	IRB00000843
FWA:	FWA00005639
Date:	June 6, 2011
Title:	'The Role of Sensory Feedback in the Regulation of Gait Parameters'
Project:	10-264
Approved:	06/03/11
Expires:	06/02/12

On behalf of the Institutional Review Board at the University of Idaho, I am pleased to inform you that the protocol for the above-named research project is approved as offering no significant risk to human subjects.

This approval is valid for one year from the date of this memo. Should there be significant changes in the protocol for this project, it will be necessary for you to resubmit the protocol for review by the Committee.

Traci Craig

Traci Craig

Appendix 2

Copyright Letter from the Journal of Biomechanics

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Corresponding author:	**Ms. Melissa Thompson**
E-mail address:	**mathompson@fortlewis.edu**
Journal:	**Journal of Biomechanics**
Our reference	**BM6644**
PII:	**S0021-9290(14)00270-X**
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