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Effects of fatigue on muscle activation and shock attenuation during barefoot running

Carl Q. (Carl Quinn) Newton
Western Washington University

Effects of Fatigue on Muscle Activation
and Shock Attenuation During
Barefoot Running

By
Carl Q. Newton

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Moheb A. Ghali, Dean of the Graduate School

ADVISORY COMMITTEE

Chair, Dr. David N. Suprak

Dr. Kathleen M. Knutzen

Dr. Lorrie R. Brilla

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Signature:__________________________________

Date: __________________________________

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Western Washington University

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ABSTRACT

Lately, barefoot running has received attention from many acknowledged researchers and athletes alike. If alterations in running mechanics related to fatigue are found to be different while running barefoot, it may help to identify the practicality and efficacy of barefoot running in terms of injury prevention. **PURPOSE:** The purpose of this study was to investigate the effect of global fatigue on muscle activation and the rate of shock attenuation during barefoot (BFT) and shod (SHD) running in habitually shod runners. **METHODS:** Eleven well-trained runners were recruited from the community to complete protocols in two different footwear conditions (BFT and SHD). Each condition consisted of two instrumented 100-meter run trials (pre- and post-fatigue) where data for the rate of tibial acceleration (RA) and preactivation iEMG of the tibialis anterior (TA) and medial gastrocnemius (MG) were gathered. Between each trial, subjects completed a 30-minute maximal effort fatiguing run in shoes on an outdoor running track. The two conditions differed only in footwear donned for the instrumented trials, were separated by 3-7 days, and were preceded by 36 hours of non-exercise rest to reduce confounding fatigue. Three two-by-two repeated measures ANOVAs with factors Time (pre, post) and Footwear (BFT, SHD) were applied with an alpha level of 0.05. **RESULTS:** The results revealed a significant difference for RA based on condition. BFT yielded higher acceleration rates than SHD (1646.64 g/s vs. 650.93 g/s; $p < 0.001$). RA for both conditions remained unchanged following fatigue, showing an equal ability to attenuate impact shock. TA iEMG responded differently to fatigue based on condition, according to a significant interaction observed ($p = 0.031$). Following fatigue, TA iEMG increased by 8.66% SHD and decreased by 12.54% BFT ($p = 0.015$). MG iEMG revealed no significant effect of time or footwear, acting in a similar fashion in all trials. **CONCLUSION:** Our data indicate that runners may adopt different shank muscle coordination strategies while running barefoot and shod. In addition, these strategies appear to be affected differently by fatigue in order to adequately attenuate impact shock. It may be suggested to approach barefoot running with caution, especially when an athlete is in a fatigued state.

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Chapter I

The Problem and Its Scope

Introduction

Running is a quickly growing sport, and the prevalence of overuse injuries is quite high amongst elite and amateur athletes. Most of these injuries occur during training, and it has been hypothesized that alterations in running biomechanics due to fatigue may be a factor. Epidemiological studies from previous and the present decades cite similar injury rates amongst trained runners (Knobloch, Yoon, & Vogt, 2008; Reinking, Austin, & Hayes, 2007; van Mechelen, 1992; Wen, 2007). With nearly 90 percent of injuries being overuse in nature there has been considerable concern regarding prevention from a biomechanical perspective. Recently, vast technological advancements have been made in construction of running shoes, partly in attempt to quell the high occurrences of overuse running injuries (Warburton, 2001). However, no such reductions in injury rates have been observed. Although there are many possible reasons for this dilemma, recent interest has been directed towards faulty shoe design and the efficacy of barefoot run training. Lately barefoot running has received attention from many acknowledged researchers and athletes alike. Proponents of barefoot running lay claim that mechanics can be improved and injury rates can be sharply reduced; however, the research is incomplete (Jungers, 2010; Warburton, 2001).

In support of this concept, one of the foremost theories is that intrinsic foot muscles may be strengthened or better activated when one performs barefoot activities (Robbins & Hanna, 1987). Confirmation that barefoot running allows these intrinsic muscles to function more optimally is lacking. Furthermore, it is unknown if such augmentation would actually reduce injury rates as these muscles are quite small and serve minor roles in accessory function. Long term studies regarding the effect of barefoot run training in habitually shod runners is certainly warranted. Barefoot running may be an avenue for injury prevention, but long term prospective injury studies are needed where individuals are trained either completely or partially barefoot. However, before such studies can take place it is critical to understand how global fatigue affects running mechanics parameters during barefoot running.

Fatigue has been well studied in the literature regarding kinematic and kinetic changes during running. Most of these studies utilize treadmills for data collection which may not appropriately mimic overground running. Many studies also use an incremental test to induce fatigue which may have a different effect from fatigue during prolonged submaximal running such as that observed in the majority of training bouts (Hanon, Thepaut-Mathieu, & Vandewalle, 2005). There is a need to describe the effect of prolonged overground running, and the global fatigue it creates, on individuals while running barefoot. This will allow researchers to better understand how safe and practical barefoot training can be implemented in future research settings aimed at training individuals previously unaccustomed to barefoot activities. By understanding how these individuals will respond to fatigue while running barefoot, future work on barefoot training can be employed while maintaining a safe protocol and environment for such subjects.

Purpose of the Study

This study investigates the effect of global fatigue on muscle activation and kinetics during barefoot and shod running in habitually shod runners. The purpose was to examine changes in preactivation of the tibialis anterior and medial gastrocnemius muscles and rate of tibial acceleration upon impact between footwear conditions as well as before and after a fatiguing protocol.

Hypotheses

The null hypothesis is that there is no difference in muscle activation or tibial acceleration between barefoot and shod conditions following a running fatigue protocol. It was expected that fatigue may alter running mechanics differently in a barefoot runner than that of a shod runner, thus illustrating the importance of the shoe in an individual accustomed to using it.

Significance of the Study

It should be paramount to maintain physical health in recreational runners such that they can avoid injuries and remain active in their sport. If alterations in running mechanics related to fatigue are found to be more detrimental while running barefoot, it may help to identify the practicality and efficacy of barefoot running in terms of injury prevention.

Limitations of the Study

1. The study was performed on both male and female runners. Although there may be minor differences in gait mechanics between sexes, the study design called for within subjects analyses which reduces this as a confounding variable.
2. The subjects were moderately-trained habitual shod runners; the findings of the study may not apply to elite runners or individuals with substantial barefoot running experience.
3. Running intensity was not confirmed by respiratory gas analysis. The intensities performed may not have been the same in each fatigue condition, but the athletes were trained enough to have a good sense of perceived exertion.
4. All subjects used individual equipment. Footwear was not controlled for; however, subjects were asked to wear the same shoes for all shod running segments (a shoe which they were accustomed to training in).

5. Training activities, diet and sleeping patterns were not controlled during the interim between conditions. Subjects were asked to maintain their typical habits and refrain from exercise 36 hours prior to each trial; however, it is understood that these variables could still have an effect on running performance.
6. The fatigue run protocol was performed shod in both conditions. Although performing a barefoot fatiguing run for the barefoot condition may have been more practical, ethical considerations regarding prolonged barefoot running in unaccustomed runners confined the investigation to allow barefoot running only in short pre- and post-test collections.
7. Only two muscles, the medial gastrocnemius and tibialis anterior, were measured for muscle activity to obtain fatigue and coordination information regarding the primary plantar and dorsiflexors. Although other shank muscles play critical roles in ankle stability and movement, they were not included due to impedance of the accelerometer fastening materials or error regarding muscular cross-talk since activity was measured using surface electrodes.

Definition of Terms

Accelerometer: Device used to measure acceleration. In the case of this study, impact shock of the tibia is assessed from tibial accelerations during running (Mercer, Bates, Dufek, & Hreljac, 2003).

Electromyography (EMG): Measurement of electrical muscle activity to infer recruitment strategies (De Luca, 1984).

Fatigue: A breakdown in running mechanics, loss of efficiency, and /or inability to maintain submaximal exercise (Cavanagh & Williams, 1982; Elliott & Roberts, 1980; Williams & Cavanagh, 1987).

Integrated EMG (iEMG): The product of muscle activity and time expressed in Volt-sec, used to express the overall muscle effort (Smoliga, Myers, Redfern, & Lephart, 2010).

Injury: Physical hindrance that prevents an athlete from practicing regular training activities for at least one week, typically musculoskeletal overuse in nature (Knobloch, et al., 2008).

Medial gastrocnemius: Muscle located on the medial posterior aspect of the leg that acts to plantarflex the foot and flex the knee (Divert, Mornieux, Baur, Mayer, & Belli, 2005).

Pre-activation: Period of muscle activation prior to impact. In the present study, 50ms prior to footstrike was assessed (Nigg, Stefanyshyn, Cole, Stergiou, & Miller, 2003).

Running intensity: Measure of effort while running that was determined by subjects' rating of perceived exertion (Shim, Acevedo, Kraemer, Haltom, & Tryniecki, 2003).

Running shoe: Shoe worn to perform the majority of run training. Typically chosen or prescribed individually to provide benefits in cushioning or support as needed (Butler, Davis, & Hamill, 2006).

Shock attenuation: Ability of the musculoskeletal system to reduce impact forces traveling up the kinetic chain that are experienced while running (Mercer, Bates, et al., 2003).

Tibialis anterior: Muscle located on the anterior portion of the leg, acts to dorsiflex and invert the foot (Divert, et al., 2005).

Chapter II

Review of Literature

Introduction

Avoiding overuse injuries has been identified as a primary concern for runners at all levels of competition. A recent report by Reinking, Austin, and Hayes (2007) found that 68% of competitive runners experience exercise related leg pain annually that may prevent them from training and competing. The modern running shoe was first invented in the 1970s as a device to protect runners from excessive impact forces and increase performance. Running shoes have evolved over the last 40 years to help stabilize the foot and cushion footstrike landings (Lieberman, et al., 2010). In recent years, very large advancements in shoe technology have been made; however, there has been no indication of a decline in overuse injuries. An epidemiological study from 1992 indicated injury rates of about 56% in competitive runners (van Mechelen, 1992).

Today, typical training shoes are designed to prevent excessive rearfoot and forefoot motion in addition to cushion impact experienced under the heel. Recently, the topic of barefoot running has been receiving attention from acknowledged researchers in many fields of study. There has been concern regarding potential negative effect of using running shoes. Two primary issues have been discussed. First, a theory that wearing running shoes leads to atrophy of intrinsic foot muscles which could alter mechanics and lead to greater injury risk (Robbins & Hanna, 1987). Second is the idea that running shoes block sensory feedback from the plantar surface, allowing an individual to continually run with high joint impact forces uncorrected (Robbins & Gouw, 1991). Advocates of barefoot running lay claim that removing shoes may reverse both of these problems.

This review sets forth to describe the purpose of modern running shoes and investigate the efficacy and practicality of running barefoot. Particular attention has been given to critically analyze studies' methodologies and sample characteristics including anatomical structure and footstrike patterns. Ultimately, barefoot running is examined for its latent ability to minimize injury risk.

Review of the Pertinent Literature

Measurement of running gait. Gait has been well studied in the biomechanics literature. Several parameters are commonly analyzed to determine the kinematic and kinetic properties of ideal running form including stride length, frequency and variability; trunk, hip, knee and ankle angles throughout different phases; integrated electromyography (iEMG) burst duration and EMG modulation of involved muscles; as well as impact forces at footstrike and shock attenuation through the kinetic chain (Chapman, Vicenzino, Blanch, & Hodges, 2008; Dugan & Bhat, 2005; Elliott & Roberts, 1980). These measurements have also been applied to barefoot running studies attempting to shed light on the purpose of a running shoe.

Dugan and Bhat (2005) reviewed the use of various gait analysis methods and summarized the biomechanics of typical running to demonstrate the utility of different gait assessment techniques. Gait assessment through physical observation, motion analysis, force platforms, dynamic electromyography (EMG) and energy expenditure were discussed. Dynamic EMG can be used to examine neuromuscular control while running. Commonly, integration is employed whereby rectified signals are summed over time and assessed within subphases of the gait cycle. Emphasis was placed on collecting data over a large number of consecutive strides, avoiding start and stop phases, and measuring inconspicuously such that the subject can produce a natural gait pattern. The authors declared the importance of thoroughly understanding functional anatomy and synchronous movements in running biomechanics in order to properly treat or prevent injuries (Dugan & Bhat, 2005).

Multiple researchers have described the subtalar joint as playing an important role in many motions of the foot. It is characterized as having an oblique axis of rotation that is on average 23^{o} medial of the long axis of the foot and 41^{o} above the horizontal plane (Dugan & Bhat, 2005; O'Connor & Hamill, 2005). Accompanying these averages are vast individual variances ranging from 4^{o}-47^{o} in the transverse plane and 21^{o}-69^{o} in the sagittal plane (Dugan & Bhat, 2005). The oblique axis of the subtalar joint creates movements that are multiplanar which can be difficult to measure in orthogonal axes models (Dugan & Bhat, 2005). The foot structure has also been praised for its ability to adapt to terrain and disperse stresses when mobile and create a solid platform for propulsion when rigid. Intrinsic muscles, including the interossei, lumbricals, quadratus plantae, abductor hallucis, flexor hallucis brevis, abductor digiti minimi, and flexor digiti minimi brevis, all assist the plantar fascia in flexing the foot. These muscles act to support the foot in conjunction with the windlass mechanism whereby the metatarsal heads are pulled toward the calcaneus during toe-off to increase rigidity in the longitudinal arch, creating a better lever to be used in propulsion (Dugan & Bhat, 2005).

Joint coupling between the forefoot, rearfoot and shank have been studied in gait, and examined for their role in proper gait mechanics. Recently, variations in these joint couplings have been identified at different gait velocities (Pohl, Messenger, & Buckley, 2007). Twelve subjects were examined on a treadmill at four speeds: walking at 1.25 m/s, and running at 2.5, 3.0, and 3.5 m/s. Three dimensional kinematics were assessed and cross-correlational coefficients were calculated for various joint coupled movements. Peak rearfoot eversion and forefoot dorsiflexion and abduction were all lower while walking compared to running ($p<0.05$).

Peak forefoot dorsiflexion and rearfoot eversion occurred later in stance while walking. Furthermore, peak forefoot abduction and shank internal rotation occurred earlier in stance while walking. No significant differences between running speeds were observed except for the time of peak shank internal rotation, which occurred later in stance as speed increased. Correlations between rearfoot eversion and shank internal rotation were higher for running ($r>0.95$) than for walking ($r=0.49$). Running also possessed higher coupling correlations than walking for frontal plane rearfoot motion with sagittal and transverse plane forefoot motions. These differences in correlation were observed regardless of running velocity, and no significant differences were observed between running speeds (Pohl, et al., 2007). This finding points toward the need to assess running gait specifically when studying joint coupling motions in research and clinical settings. Examinations based on walking are not adequate to assume the interactions between the rearfoot and shank motions while running.

By studying these joint coupling interactions during running, researchers are able to assess coordination between the two segments and how it may pertain to a desired running form. Coordination between the subtalar and knee joints during running was examined by Stergiou, Bates, and Kurz (2003) while altering stride length. Six recreational runners were used to run over a force platform using normal strides (NS), short strides (SS) and long strides (LS). The long and short strides were set as one foot-length over and under the normal stride length. Sagittal and frontal planes 2-dimensional video data were collected to assess leg and rearfoot motion. The initial peak in the vertical ground reaction force upon footstrike was also assessed and normalized to body mass for each condition (Stergiou, Bates, & Kurz, 2003).

The impact peak increased concurrently with increasing stride length (SS, 1.677 BW; NS, 1.737 BW; LS, 2.185 BW). Also associated with increased impact forces were greater maximum angular velocity differences between the two joints (Stergiou, et al., 2003). These greater angular velocity differences were linked with higher impact peak forces and were especially increased in the LS condition. Functionally, a greater difference in angular velocities depicted antagonist relationships at either end of the tibia. The authors proposed this lack of coordination as a potential injury mechanism and that better joint coordination could be achieved by reducing initial impact peak forces (Stergiou, et al., 2003). In the case of this study, shorter strides were used to elicit improvements in knee-subtalar joint coordination.

Frontal plane movements of the foot are often assessed with kinematics of the subtalar joint. The joint axis is commonly approximated as the anterioposterior (AP) axis down the length of the foot. O'Connor and Hamill (2005) declared that these approximations may greatly limit research findings because the anatomical axis of the subtalar joint is oblique with an orientation inclined by 42^o and medial by about 23^o. They conducted a study to examine interpretations based on kinetic data from each joint axis method. Ten subjects ran across a force platform while synchronously being examined by 3-dimensional kinematics. Shoes with a removed heel counter were used so that a skin marker could be placed directly on the calcaneus. The oblique axis of the subtalar joint was calculated using a movement test where subjects rotated in the X- and Z-axes. This oblique axis compared to the standard approximation based on the Y-axis in all measures (O'Connor & Hamill, 2005).

Although range of motion did not differ, the oblique axis method showed less inversion at footstrike and greater peak eversion during midstance (O'Connor & Hamill, 2005). Furthermore, mechanical work calculated about the oblique axis was different than the AP axis. More energy absorption during the first portion of stance and energy generation in the second portion of stance were observed when using the oblique axis. Also, the subtalar joint yeilded greater positive total work than the AP axis approximation. O'Connor and Hamill (2005) concluded that although the AP axis can approximate some kinematics well, it results in very different joint moments than kinetic data obtained from the subtalar joint oblique axis. The oblique axis moment was almost entirely supination whereas the AP axis was inversion during the first half and eversion during the second half of stance (O'Connor & Hamill, 2005). These findings by O'Connor and Hamill (2005) show the importance of using a more functionally accurate approximation of the subtalar joint when analyzing gait mechanics and relation to lower extremity injury potential.

Smoothness in running has also been discussed in the literature (Hreljac, 2000). Well trained runners have been compared to well trained athletes from other sports for this measure while walking and running (Hreljac, 2000). End-point jerk cost (JC) was calculated to measure smoothness in gait. Two dimensional kinematic position data of the lateral heel throughout multiple gait cycles were collected. Essentially, the jerk (or 3rd derivative of position data) was squared and integrated by time to obtain the JC. A total of 24 subjects (12 runners, 12 non-runners) were examined while walking at 1.75 m/s and running at 3.35 m/s on a treadmill. As expressed by the position data, the runners were able to ascend and descend the heel more gradually than the non-runners (Hreljac, 2000). With every subsequent derivative calculated, this difference became even further evident. Runners displayed significantly lower JC values than non runners while walking and running. Hence it was concluded that competitive runners were inherently smoother at gait related tasks than non-runners (Hreljac, 2000). It was noted that the theory of minimum jerk concerns movement planning rather than movement execution. Since it can be assumed that all athletes planned on making their movements as smooth as possible, it was concluded that the runners were more successful in executing their plan (Hreljac, 2000).

These findings point toward the idea that runners may exhibit better coordination than non runners and that improved coordination can produce a smoother stride and potentially limit injury risk. The author suggested future studies in examination of muscular coordination in runners and how it pertains to development of a smoother gait cycle (Hreljac, 2000).

Running fatigue and its implications. Fatigue can play a critical role in limiting run performance and increasing injury risk. Biomechanical evaluations of the role of fatigue in running have been studied by many researchers. Elliot and Roberts (1980) examined kinematic changes that occur in elite middle-distance runners while negotiating a 3000-meter race simulation. Eight highly trained runners were filmed at 4 different stages while running 3000 meters on a 400-meter track. The stages were set at 500m, 1300m, 2100m, and 2900m. No statistical differences in horizontal running velocity were present between each of the four stages; therefore, the study was well controlled to examine spatio-temporal variables of stride length, stride frequency, support time and non-support time (Elliott & Roberts, 1980).

No differences in any tested variable were observed between the first three stages, but mechanics were altered slightly at the fourth stage indicating the onset of fatigue (Elliott & Roberts, 1980). Stride length decreased and stride frequency increased at the 2900m mark compared to the earlier stages. Furthermore, stance time was significantly longer and flight time was shorter in the final stage. Changes in angular positioning of various segments were also evident in the fatigued stage (Elliott & Roberts, 1980). The leg angle relative to the ground at footstrike was increased from 90.6^{o} to 93.1^{o} depicting a foot contact that was further in front of the body. This finding was expected to produce a greater braking force and reduce the runner's efficiency. A more forward trunk lean and less thigh hyperextension at toe-off were also observed when fatigued, which ultimately led to a smaller hip angle in this stage. These kinematic changes were viewed by the authors as negatively impacting each runner's efficiency, and attention was given to discussing maintenance of coordinated movement patterns (Elliott & Roberts, 1980). In order to optimize running efficiency, investigations should include examination of muscular coordination strategies.

Kinematics have also demonstrated changes while running at different intensities. Shim, Acevedo, Kraemer, Haltom, and Tryniecki (2003) studied the alteration of kinematics in nine competitive runners while running at three different intensities corresponding to the onset of blood lactate accumulation (OBLA: 4 mM blood lactate). The test intensities were set to VO_2 10% below OBLA, VO_2 at OBLA, and VO_2 10% above OBLA. The subject ran on a treadmill while sagittal plane video data were captured and later digitized for one stride at each intensity. Blood lactate revealed a non-linear rise with increasing intensity; tests 1, 2 and 3 reported 2.88, 4.05 and 6.73 mM blood lactate, respectively. Stride length, stride frequency and running velocity increased, support time decreased, and flight time remained unchanged at higher intensities (Shim, et al., 2003). Additionally, as running velocity increased, knee and thigh flexion increased during the flight phase to reduce the moment of inertia and improve angular velocity (Shim, et al., 2003). Correlation coefficients were also calculated for each variable with blood lactate. Only two variables displayed significant correlations with blood lactate: stride frequency ($r=0.51$) and the vertical oscillation of the center of gravity ($r=0.44$). Although this study did not incorporate fatigue, it did portray changes that can be expected at higher intensities that are designed to improve efficiency (Shim, et al., 2003). With the onset of fatigue, these alterations may break down leading to unfavorable running mechanics.

Whereas global fatigue may impair running mechanics, less is understood regarding the effect of localized fatigue of integral shank muscles. Researchers from the University of Calgary investigated how fatigue of the tibialis posterior affects foot kinematics during walking (Pohl, Rabbito, & Ferber, 2010). Twenty-nine subjects were recruited to undergo an exercise fatigue protocol intended to reduce the force output and function of the tibialis posterior muscle. First, the subjects were examined while walking on a treadmill at 1.1 m/s using three-dimensional kinematics. Prior to the fatiguing exercise, maximal voluntary contractions (MVC) were collected for closed chain isometric foot adduction. Subjects were constrained to a chair and instructed to perform 4 sets of 50 repetitions of concentric/eccentric foot adductions at 50% MVC. Following the protocol, MVC was retested. If MVC remained above 70% the pretest value, the protocol was repeated until 70% MVC could not be achieved indicating sufficient fatigue (Pohl, et al., 2010). Once fatigued, subjects immediately performed a posttest walking trial to compare to the initial examination. Ten consecutive strides were analyzed for each walking condition and the variables studied included: rearfoot peak eversion (EVE), EVE excursion, time to peak EVE, forefoot peak dorsiflexion (DF), DF excursion, time to peak DF, and forefoot peak abduction (ABD).

Significant increases in EVE and ABD ($+0.7^{o}$ and $+0.3^{o}$ respectively) were observed following the fatigue protocol, but the authors felt they were too small to provide substantial clinical relevance (Pohl, et al., 2010). No other significant changes were noted for any other tested variable. The authors concluded that a fatigue protocol that successfully reduces isometric adduction force by 30% has no relevant effect on rearfoot motion during walking (Pohl, et al., 2010). It was speculated that other contributing non-fatigued musculature provided extra support to control eversion and abduction. The use of electromyography to examine changes in muscle activity and understand compensation strategies by other muscles was suggested (Pohl, et al., 2010). Furthermore, it should be noted that walking and running are quite different activities.

Pohl, et al. (2007) demonstrated that foot-shank coupling actions are much different between walking and running. Running possesses higher impact forces that require greater muscular control and it could be postulated that the tibialis posterior plays a less critical role while walking than running.

Another group from Calgary tested changes in oxygen consumption and muscle activity of runners in two pairs of shoes with different cushioning properties (Nigg, et al., 2003). Twenty runners with a natural heelstrike running style donned the two shoes which differed only in the compliance of the heel. Two testing sessions were conducted to measure oxygen consumption during steady state running slightly above anaerobic threshold. The sessions reversed order of which shoe the subjects ran in, presenting a total of four 6-minute trials for each shoe to be analyzed. Muscle activity of the tibialis anterior, medial gastrocnemius, vastus medialis and biceps femoris were collected in a special testing session similar to but dysynchronous with the oxygen consumption data. The EMG data were bandpass filtered with a 20-400 Hz cutoff and converted by a root mean squared method. Pre- and postactivation were assessed 50ms before and after heelstrike in each muscle (Nigg, et al., 2003).

Mean differences between the two shoe conditions were not found for oxygen consumption, but were identified in the EMG measurements (Nigg, et al., 2003). The softer shoe displayed higher preactivation of the tibialis anterior (+3.2%), gastrocnemius (+0.9%), and biceps femoris (13.2%), while the harder shoe used higher preactivation of the vastus medialis (+7.2%). Individual differences were also observed for all measures (Nigg, et al., 2003). Some subjects were more efficient with oxygen consumption in the softer shoe, others in the harder shoe, while some showed no difference between the two. Interestingly, the vastus medials showed systematic changes based on these individual efficiencies. In all cases, the vastus medialis had more preactivation in the shoe which required greater oxygen consumption (Nigg, et al., 2003). These changes in muscle preactivation prior to heelstrike could provide intriguing discussion regarding muscular control of impact forces. The authors speculated that EMG preactivations are preprogrammed based on anticipated impact shock and related to a reflex arc event (Nigg, et al., 2003). They described this reflex mechanism as 'muscle tuning' and they considered it as involving more fast twitch muscle fibers which could be subject to fatigue (Nigg, et al., 2003). Examinations whether a reduction of muscular preactivation exists due to fatigue in various shoe conditions could provide insightful information regarding running performance and injuries.

Changes in EMG signals have been considered by many researchers to help assess fatigue while running. Hanon, Thepaut-Mathieu, and Vandewalle (2005) conducted a study aimed at developing an appropriate methodology using intregrated EMG analysis to measure running fatigue. Nine highly trained runners were examined during an incremental treadmill test consisting of 4-minute progressively increasing stages until exhaustion. The treadmill test remained at a 0% gradient only increasing belt velocity between each stage. EMG from the gluteus maximus, rectus femoris, biceps femoris, vastus lateralis, gastrocnemius and tibialis anterior were collected at 45-sec and 3-min 40-sec of each exercise stage. Data were collected long enough to assess 10 consecutive bursts in each trial, and EMG signals were then intregrated over each whole burst to obtain the iEMG. The iEMG bursts were normalized and compared to maximal iEMG recorded during previously tested 6-sec maximal voluntary isometric contractions for each muscle. EMG was also analyzed on a distance scale where the amount of muscle activity to produce a given amount of work was evaluated. The iEMG was calculated across 20 meters at the start and end of each stage and compared within and between stages. This allowed EMG per work unit to be examined as stride length and running velocity changed (Hanon, et al., 2005).

The level of activation increased for all muscles with an increase in intensity; however, the duration of activation yielded different results dependent on the muscle (Hanon, et al., 2005). The gastrocnemius and vastus lateralis tended to decrease relative burst duration as speed increased while the biceps femoris and rectus femoris increased burst duration likely due to their roles as hip mobilizing muscles (Hanon, et al., 2005). Fatigue was assessed by comparing iEMG between the start and end of each stage. The gastrocnemius was the only muscle to prove fatigue resistance throughout all stages, while all other muscles were indicating increases in iEMG from the start to the end of the final stage. The rectus femoris and biceps femoris showed the earliest signs of fatigue yielding significant differences by the third-to-final stage. The authors concluded that when examining fatigue during treadmill running of increasing velocity measuring iEMG on a distance scale is most appropriate (Hanon, et al., 2005). In this case hip mobilizing muscles are subject to fatigue sooner than other leg muscles. The hip muscles were considered the prime forward movers whereas the ankle and knee extensors were thought to regulate leg stiffness before and during the contact phase (Hanon, et al., 2005). Therefore, as speed increases the contribution of hip musculature is increased leading to earlier fatigue. In the case of prolonged submaximal running a different fatigue effect could be expected (Hanon, et al., 2005). The gastrocnemius and vastus lateralis muscles may fatigue sooner at slower speeds potentially altering running mechanics. Furthermore, over-ground and treadmill running may constitute different adaptations to fatigue and require different methodology with EMG measurements.

Apart from general fatigue during running, local fatigue of specific muscle groups have been studied on their specific effects on gait kinematics and muscular activity. Kellis and

Liassou (2009) compared the effect of fatiguing protocols of knee flexor/extensor muscles to that of ankle plantar flexor/dorsiflexor muscles on sagittal running kinematics. Fifteen well trained female distance runners were selected for the study. Maximum knee extension and flexion torques as well as ankle plantar flexion and dorsiflexion torques were collected on an isokinetic dynamometer prior to the study to establish target levels for fatigue protocols. The fatiguing exercises were performed on separate days for the knee and ankle joints at 120 deg/s until 30% peak torque could no longer be maintained. Both before and immediately following fatigue the subjects ran for 10 seconds at 3.61 m/s on a treadmill while 3-D kinematics and muscle activity were collected. Only sagittal plane kinematics were examined to match the chosen muscle groups and the data collection allowed 5 strides to be averaged and analyzed. Hip, knee and ankle angles at footstrike and toe-off were assessed, as well as maximal angular displacement during midstance and swing phases. Muscular activity of the vastus medialis, biceps femoris and gastrocnemius were also collected and normalized to previously recorded maximal voluntary isometric contractions (Kellis & Liassou, 2009).

Following the knee muscle fatigue protocol, increased knee flexion at contact and increased hip and knee flexion at toe-off were observed (Kellis & Liassou, 2009). The ankle fatigue protocol caused a decreased dorsiflexion at impact and increased knee flexion at toe-off (Kellis & Liassou, 2009). Both fatigue protocols resulted in increased EMG from the vastus medialis and gastrocnemius, but had different effects on biceps femoris activity. Biceps femoris EMG was increased after knee fatigue and decreased after ankle fatigue during the swing phase; however, no changes in pre-activation, stance or propulsion phases were observed. The authors noted that kinematic changes in running are dependent on the muscle group being fatigued, although the function of these changes were similar, in that protective kinematic adjustments were acquired to maintain joint stability (Kellis & Liassou, 2009).

The use of electromyography in running studies has been well documented. Researchers have measured muscular activity to infer movement coordination and fatigue responses; however, many different data processing techniques have been applied without much discussion regarding reliability or validity. Recently Smoliga, Myers, Redfern and Lephart (2010) tested the reliability and precision of four different EMG measurement parameters on 13 muscles while running. Fifteen competitive distance runners were recruited to run for 10 minutes on a treadmill. EMG data were collected from the vastus lateralis, semimembranosus, gluteus maximus, rectus femoris, erector spinae group, rectus abdominus, external oblique, trapezius, latissimus dorsi, anterior deltoid, middle deltoid, posterior deltoid and brachioradialis during the ninth and tenth minutes of the run. Integrated EMG (iEMG), root mean square EMG (RMS), maximum M-wave, and median power frequency (MPF) were calculated for 25 consecutive strides. EMG parameters from the ninth and tenth minute were compared using intra-class correlation coefficients (ICC) and standard error of measurement (SEM) for each muscle (Smoliga, et al., 2010).

Multiple EMG parameters were considered reliable and precise assessments of muscle activity while running; however different muscles were measured optimally using different methods (Smoliga, et al., 2010). MPF was the most reliable and precise parameter measured possessing ICC > 0.80 for 12 of the 13 muscles examined. Maximum M-wave, RMS and iEMG were all very reliable having ICC > 0.80 for 10 muscles each. The iEMG was considered superior to RMS since it had more favorable precision as measured by the SEM. The authors demonstrated that reliability of one EMG parameter does not necessarily guarantee reliability using a different method, and that choice of parameter should depend on the muscles being examined (Smoliga, et al., 2010). Shank muscles were not examined in this study, but the authors recommended the use of iEMG or MPF when studying running based on their results of the thigh muscles examined (Smoliga, et al., 2010). Furthermore, it was noted that iEMG should be normalized to stride duration rather than an arbitrary time period to obtain a more functionally specific measurement to running.

Summary. Intramuscular and surface electromyography have been used to evaluate fatigue and breakdown of gait in endurance running (Chapman, et al., 2008; De Luca, 1984). Increased EMG amplitude over time while performing a steady state submaximal exercise is accepted as a marker for muscular fatigue (De Luca, 1984). Studies have also examined EMG modulation in secondary phases of repetitive motions such as cycling or running. Decreased EMG modulation shows the muscles' inability to activate in a favorable pattern (Chapman, et al., 2008). Having delayed muscular activations following foot strike or decreased modulation in secondary EMG may lead to increased risk of injuries (Chapman, et al., 2008).

Shock attenuation during running. To achieve optimal performance and limit the occurrence of injuries, endurance runners must withstand the negative effects of fatigue. Many markers of fatigue have been indicated in prolonged running. One such measure is the runner’s ability to attenuate impact shock measured as tibial accelerations. Mizrahi, Voloshin, Russek, Verbitsky and Isakov (1997) examined how the musculoskeletal system can be affected by fatigue in attenuating impact shock. Twenty-two subjects were instrumented with an accelerometer on the tibial tuberosity and surface electrodes on the gastrocnemius and rectus femoris and studied over the course of a 30-minute fatiguing run. Each participant ran on a treadmill at a speed corresponding to their individual ventilatory threshold. Fatigue was determined and monitored based on measurements of end-tidal carbon dioxide pressure ($PETCO_2$). Following the 30-minute run, subjects were characterized by their change in $PETCO_2$ as being in a fatigue group or non-fatigue group. Twelve of the subjects demonstrated a significant decrease in $PETCO_2$ distinguishing them as fatigued (Mizrahi, Voloshin, Russek, Verbitsky, & Isakov, 1997). Accelerometer and EMG data were collected every 5 minutes for 20 seconds and normalized to the initial data taken from the start of the test. The fatigue group demonstrated a steady incline in tibial accelerations throughout the test that was significantly greater than baseline after 20 minutes of running. These accelerations paralleled decreases in $PETCO_2$. The non-fatigue group had no changes in tibial accelerations or $PETCO_2$ throughout the run test. No significant changes in EMG were observed for either group in the time or frequency domains. It was expected that fatigue would induce modifications to running mechanics that would lead to changes in muscular pre-activation patterns (Mizrahi, et al., 1997). The repetitive stresses from the fatiguing run clearly lessened the ability of some subjects to adequately attenuate impact forces; however, this decreased shock attenuation did not correlate to any changes in EMG signal. Therefore, the authors concluded that the fatigue observed was attributed to deterioration in muscle coordination rather than fatigue of the individual muscle itself (Mizrahi, et al., 1997). This raises questions regarding synergistic control of muscles while running and how fatigue can impact recruitment patterns. Furthermore, placing the subjects on a treadmill may limit the findings in that gait mechanics are altered from over-ground running.

Another study regarding the effect of running fatigue on shock attenuation was conducted by Mercer, Bates, Dufek and Hreljac (2003). Ten subjects were examined with accelerometers on the distal tibia and forehead, as well as stride parameters and oxygen consumption while running before and after a fatigue protocol. The fatiguing run in this case was a standard maximal graded exercise test. Most of stages were conducted at a 7.5% grade with increasing speed each minute until volitional fatigue. The average graded exercise test culminated after 10.1 minutes. Before and immediately following the fatigue test, subjects ran about 5 minutes at 3.8 m/s while accelerometer, stride, and oxygen consumption parameters were collected. These submaximal bouts served as the pre- and post-fatigue trials (Mercer, Bates, et al., 2003). The accelerometers were tightly secured to the distal tibia and forehead using compression bandages and recorded data for 20 seconds during the first minute of each trial (Mercer, Bates, et al., 2003). This allowed accelerations to be captured for at least 10 consecutive strides.

Oxygen consumption and heart rate were respectively 16% and 11% greater during the fatigued run. As expected, shock attenuation was on average 12% lower when running fatigued; however, no significant differences were observed for stride length or frequency (Mercer, Bates, et al., 2003). Furthermore, individual differences between subjects were reported for shock attenuation responses to fatigue (Mercer, Bates, et al., 2003). Seven of the 10 subjects displayed decreased shock attenuation, one subject remained unchanged, and two subjects increased attenuation throughout the protocol. It was assumed that the fatigue protocol being a graded exercise test elicited different responses compared to findings of previous researchers that used longer continuous runs. This test used by Mercer, et al. (2003) increased running speed gradually and was performed at a 7.5% incline. This is dissimilar to typical run training or racing and may have required different muscular recruitment. The concept that runners are less able to attenuate shock when fatigued was confirmed; however due to vast individual responses to fatigue, questions could be raised about what fatigue protocol is best suited to examine real-life situations. Running over ground does not confine a runner to a certain speed or stride length as observed on treadmill tests. These variables could affect how subjects obtain fatigue and attenuate shock.

Mercer performed a similar study with a different research team investigating the individual effects of stride length and frequency of shock attenuation (Mercer, Devita, Derrick, & Bates, 2003). They used 10 subjects of varying running ability and examined shock attenuation during conditions of altered stride characteristics. After identifying each subject's preferred stride length (PSL) and frequency (PSF), they were measured on three separate experiments consisting of three conditions each. First stride length was manipulated (+15% PSL, PSL, -15% PSL) while stride frequency was held constant at the PSF. The second experiment manipulated stride frequency (+15% PSF, PSF, -15% PSF) while controlling stride length to the PSL. The last experiment manipulated stride length and frequency concurrently to stabilize running velocity (+10% PSL/-10% PSF, PSL/PSF, -10% PSL/+10% PSF). The accelerometers were affixed to the distal anterio-medial aspect of the tibia and forehead with a tight compression bandage similar to the previous study. Subjects ran on a treadmill at 3.8 m/s and set stride frequencies to a metronome; as before, data was collected for each trial over 20-seconds to gather 10 consecutive strides (Mercer, Devita, et al., 2003).

In the first experiment, shock attenuation was significantly different with changing stride lengths and controlled stride frequency (Mercer, Devita, et al., 2003). Mean shock attenuation was 43% greater during the +15% PSL as compared to the -15% PSL. Conversely, no significant differences were observed in the second experiment when stride frequency was altered with a controlled stride length (Mercer, Devita, et al., 2003). In the experiment of concurrent changes in stride length and frequency, changes in shock attenuation were also observed. Shock attenuation was 18% greater during the +10% PSL/-10% PSF condition as compared to -10% PSL/+10%

PSF. These results indicate that changes in stride length influence shock attenuation while treadmill running independent of stride frequency (Mercer, Devita, et al., 2003). The increased shock attenuation observed with greater stride lengths was attributed to expected changes in impact magnitude. Previous research has demonstrated increased impact forces and leg stiffness with greater stride lengths. These increases in impact energy suggest a higher demand for shock attenuation (Mercer, Devita, et al., 2003). Therefore, shock attenuation during running can be viewed as being related to lower extremity compliance and kinematics at impact since these factors are directly related to stride length independent of stride frequency.

Increasing tibial accelerations during fatigued running is seen as a consistent finding in the literature. Mizrahi, Verbitsky and Isakov (2000) investigated whether an imbalance in plantarflexor and dorsiflexor muscle activity is present in conjunction with increasing impact shock. They examined 14 recreational runners over the course of a 30-minute fatiguing treadmill run. Gas analysis was monitored and the end-tidal carbon dioxide pressure ($PETCO_2$) was used to confirm that the subjects were running at about 5% above their anaerobic threshold. This protocol was described earlier by Mizrahi, et al. (1997) as instilling global fatigue in the tested runners. Acceleration of the tibial tuberosity and surface EMG from the tibialis anterior and gastrocnemius were collected for 20-seconds in the first minute and every subsequent 5 minutes for the duration of the test. This allowed data to be obtained inconspicuously to the subject for about 27 consecutive foot strikes each collection (Mizrahi, Verbitsky, & Isakov, 2000).

$PETCO_2$ steadily decreased throughout the 30-minute trial and impact acceleration increased significantly from the 15th minute onward as compared to the initial minute, both measures indicating that global fatigue developed (Mizrahi, et al., 2000). Integrated EMG remained unchanged in the gastrocnemius and significantly decreased from the 20th minute onward in the tibialis anterior. These findings portray an imbalance in the ankle plantarflexor and dorsiflexor muscle activity with increasing fatigue (Mizrahi, et al., 2000). The gastrocnemius maintained its activity while the tibialis anterior appeared to reduce due to fatigue. The authors speculated the effect of this imbalance on the mechanical loading properties of the shank (Mizrahi, et al., 2000). Increasing muscle activity on the posterior aspect and decreasing activity on the anterior aspect of the tibia could dispose this bone to excessive bending forces. These increased bending forces in conjunction with greater tibial shock could result in increased risk of bone stress injuries. Mizrahi, et al. (2000) viewed co-contractions of shank muscles as providing protection from bending forces as well as allowing proper shock absorption from impact accelerations. Therefore, fatigue from prolonged running may contribute to increased injury potential through multiple mechanisms.

Changes in tibial acceleration and shank muscle activity from global fatigue has been well documented. More recently, researchers have examined how local fatigue of shank musculature can affect such measures. A study investigated shock attenuation changes from local leg muscle fatigue in 24 active women (Flynn, Holmes, & Andrews, 2004). The subjects were instrumented with an accelerometer on the medial tibial condyle and surface electrodes on the tibialis anterior and lateral gastrocnemius. After which, they assumed a supine position in a human pendulum system designed to load the unshod foot on a vertical forceplate. Restraints were placed about the pelvis and proximal knee to control for accessory joint motion while impact force and velocity were set to that typically experienced while running (Flynn, et al., 2004). Subjects were instructed to maintain an extended knee and dorsiflexed ankle such that only heelstrike would be simulated. Testing was performed in two experimental conditions spaced one week apart: a fatigued tibialis anterior and fatigued gastrocnemius trial. The protocol consisted of maximum voluntary isometric contractions (MVC) to establish baseline EMG for normalization, pretest loading, an isometric fatigue protocol for one muscle, and posttest loading. Each fatigue protocol was completed when a 15% decrease of the mean power frequency in the respective EMG signals was achieved (Flynn, et al., 2004).

Muscle activity of the tibialis anterior and gastrocnemius increased by 25.9% MVC and 15.5% MVC, respectively following each fatigue experiment (Flynn, et al., 2004). Furthermore, peak tibial acceleration and maximal acceleration slope significantly decreased following fatigue (Flynn, et al., 2004). These results are in direct contrast to those found with global fatigue. The authors concluded that as the muscles fatigued, they became less stiff, which enhanced their shock attenuation properties (Flynn, et al., 2004). This is a clairvoyant reminder of the findings by Mizrahi, et al. (1997) that detriments to shock attenuation in running may not be due to specific muscle fatigue, but rather a lack of proper coordination. It should be noted, however, that this study by Flynn, Holmes, and Andrews (2004) does not adequately replicate impacts that would be experienced during running as it was not a dynamic test. In application to investigations of shock absorption during running a dynamic test should be performed, although it is valuable to understand the difference between individual muscle fatigue and deterioration of muscle coordination strategies.

Comparisons between different shoe types. Modern running shoe design has been evaluated for its effect on kinematics in gait, as well as kinetics, including net joint reaction forces at the hip, knee and ankle. Researchers have applied various measures to examine how different shoe types can affect these variables.

A study comparing typical stability shoes (Brooks Adrenaline) and barefoot running was conducted on 68 injury-free recreational runners (Kerrigan, et al., 2009). The subjects were instrumented for three-dimensional kinematic data collection in synchrony with ground reaction forces on an AMTI compound implemented treadmill. Kinematic and kinetic parameters were averaged across 10 consecutive strides for the two footwear conditions. Inverse dynamics was used to calculate joint reaction forces at the hip, knee and ankle (Kerrigan, et al., 2009).

As confirmed by previous studies, shod running was associated with longer stride lengths; however this did not correlate well with any increases in joint torque (Kerrigan, et al., 2009). The only strong correlate with stride length was the vertical ground reaction force, explaining 35% of the variance ($p<0.01$). Significantly higher net torques were observed for all joints measured while running with shoes compared to barefoot. The largest changes in joint torque were identified for hip internal rotation, knee flexion and knee varus, increasing 54%, 36% and 38%, respectively, while wearing shoes (Kerrigan, et al., 2009). The results of this study warrant the development of new running footwear that can provide compliance to normal foot function while minimizing increases in hip and knee reaction torques.

Many researchers agree that running shoes can be adequately prescribed based on an assessment of individual running mechanics. In order to do so, special equipment is necessary and the evaluator must be experienced with the measurements. When such procedures are not available, many individuals default to recommending shoes based empirically on foot type as determined by arch height. Recently the use of motion control and cushioned trainer shoes in high- and low-arched runners was evaluated to test the efficacy of these recommendations (Butler, et al., 2006). Forty recreational runners (20 high-arch, 20 low-arch) were examined using three-dimensional kinematics and kinetics while running in the two types of shoes. Arch height index (AHI) and AHI stiffness (AHI/kg; change in arch height while loaded divided by the change in load from 10% to 50% BW load) were calculated for each subject. The subjects ran along a 25-meter runway at 3.5 m/s. A stationary force platform was used to obtain ground reaction forces and an accelerometer attached to the tibia measured shock at impact (Butler, et al., 2006).

Clear differences were observed between shoes irrespective of arch type (Butler, et al., 2006). Peak eversion and eversion excursion were decreased by 11% and 6%, respectively, in the motion control shoe compared to the cushioned trainer. However, peak tibial acceleration and lower extremity stiffness were reduced by 20% and 2%, respectively, in the cushioned trainer when compared to the motion control shoe. Furthermore, an interaction of arch type was observed between shoes only regarding the maximal load rate of the vertical ground reaction force. The cushioned trainer reduced load rate in the high-arched runners, while it actually increased load rate in the low-arched runners when compared to the motion control shoe. The authors explained these findings to confirm and validate the common practice of recommending motion control shoes for low-arched and cushioned trainers for high-arched individuals (Butler, et al., 2006). According to these findings, arch index may be sufficient information to prescribe running shoes. However, the authors noted that arch type does not necessarily dictate running mechanics and that ideally shoes should be recommended based on running mechanics (Butler, et al., 2006).

In addition to using specific shoe types, some clinical practitioners may prescribe custom foot orthotics to patients who they suspect to have compromised foot type or function. The efficacy of using pressure data to prescribe orthotics was examined using lower extremity kinematics and pressures while running (Dixon & McNally, 2008). Researchers recruited 22 recreational runners to don semi-custom orthotics (Dixon & McNally, 2008). The orthotics were prescribed using classification of foot type based on barefoot running plantar pressure. Orthotics were provided in three main classes: low-arch, normal-arch, and high-arch. After being fitted, subjects ran across a pressure plate under two conditions: wearing orthotics in neutral shoes and wearing neutral shoes only. Three-dimensional kinematic data was also collected to evaluate rearfoot, ankle, tibial torsion, and knee angles (Dixon & McNally, 2008).

The orthotics resulted in an overall more inverted foot function (Dixon & McNally, 2008). The initial rearfoot angle at contact was more inverted and underwent less eversion through midstance. Orthotics reduced peak eversion by 2.2 degrees and eversion velocity by 11.1 deg/s compared to the shoe only condition. The ankle was more dorsiflexed at both contact and midstance while wearing the orthotics. No significant differences were observed between conditions for tibial torsion or knee flexion angles, indicating that the orthotics primarily influenced the ankle and subtalar joints only (Dixon & McNally, 2008).

Recently, custom-made medial-posted orthotics have been examined for their ability to reduce pain in patients with overuse injury symptoms (Hirschmuller, et al., 2009). Thirty-nine moderately trained recreational runners with various chronic lower extremity overuse symptoms were prescribed custom shoe orthotics for 8 weeks. The orthotics were designed based on plantar pressure during dynamic barefoot walking. An additional 42 subjects served as a control group receiving no initial orthoses. Inclusion in the statistical assessment warranted that all patients were able to maintain training habits and keep a diary of pain and comfort scales. The Pain Disability Index (PDI), Subjective Pain Experience Scale (SES) and a comfort index of orthoses (ICI) were recorded by each subject across the 8 week intervention (Hirschmuller, et al., 2009).

PDI scores for the orthoses group decreased from 4.0 to 1.6 and increased from 4.1 to 4.8 after the intervention (Hirschmuller, et al., 2009). Similarly the SES scores decreased from 29.9 to 25.9 in the orthoses group and increased from 31.6 to 32.5 in the controls. The authors reported these findings as statistically significant evidence that orthotics decrease pain (Hirschmuller, et al., 2009). Conclusions drawn from these findings may be misrepresentative of running since the pain scales utilized are primarily used to assess functioning in activities of daily living. Furthermore, the maximal scores for the PDI and SES are 70 and 100, respectively, and may be subject to a floor effect with the intended sample.

The issue regarding effects of long term orthotic use remains unaddressed. It is plausible that orthotics are only a temporary solution that could predispose patients to additional problems after substantial wear due to atrophy of intrinsic muscles in the feet.

A previous case study investigated hip joint loads with various types of footwear (Bergmann, Kniggendorf, Graichen, & Rohlmann, 1995). An active, healthy eighty-two year old man with coxarthrosis underwent a bilateral total hip replacement. A titanium hip with ceramic head was used, instrumented with three telemetric strain gauges to transmit force and moment readings from the prosthesis. Seven months post-operational surgery, data was collected for walking and jogging with different types of footwear. Eight running shoe, two hiking boot, two evening shoe, and two clog pairs were compared along with a barefoot condition to establish baseline measurements. Each shoe was evaluated and ranked based on heel and forefoot hardness (Bergmann, et al., 1995).

The subject walked (3km/hr) and jogged (6km/hr) using soft, normal and hard foot strikes for a total of 6 collections on each shoe pair. Resultant hip joint force as well as bending and torsional moments were calculated for each trial (Bergmann, et al., 1995). Each respective barefoot condition was used as a baseline measure and each shoe was calculated as its percent change from barefoot. Resultant hip joint forces and bending moments showed modest changes of -4 to +6% BW for various shoes in walking and running. Torsional moments increased substantially +20 to +26% BW for most shoes. No clear patterns were observed between shoe hardness and hip joint forces; however, each barefoot condition yielded smaller forces (Bergmann, et al., 1995). Furthermore, soft foot strikes provided decreased bending, torsion and resultant forces from normal gait in every condition. Conversely, hard foot strikes yielded increases in all respective forces (Bergmann, et al., 1995). This data suggests that hip joint forces have no dependence on shoe sole material and that softer landings or barefoot conditions may reduce forces experienced at the hip.

Speculation exists around overestimation of kinematic changes by studies on running mechanics that used markers on skin and external shoe surfaces. Critical of such previous studies, a research team from Europe designed a similar investigation using intracortical bone pins with reflective markers attached (Stacoff, Nigg, Reinschmidt, van den Bogert, & Lundberg, 2000). Five subjects were instrumented with bone pins in the lateral calcaneus and tibia to examine the skeletal movement during running. Running conditions included barefoot as well as shod with 5 insole modifications of varying stabilization and support. Very small mean differences were observed comparing barefoot to shod conditions for calcaneal eversion and tibial rotation, on the order of 1-3 degrees. Larger differences were observed between subjects (up to 10 degrees) for running kinematics, suggesting vast individual variations (Stacoff, et al., 2000). Joint coupling for barefoot and shod running were similar between footwear and changes in tibiocalaneal movement patterns were minor and not systematic across subjects. Stacoff, et al. (2000) concluded that bone movements during barefoot running were largely similar to those while shod; however, only the rearfoot was considered and foot type was not accounted for, opening their findings to potential limitations.

Footwear motion relative to the foot and its effect on subsequent forefoot and rearfoot motion has been studied during running (Morio, Lake, Gueguen, Rao, & Baly, 2009). Researchers used two different sandals with midsoles equal to typical running shoes with high and low cushioning abilities (Morio, et al., 2009). The open upper portion of the sandals allowed for placement of markers on the feet as well as the shod sole. Ten subjects ran across a force platform linked with a three-dimensional motion capture system for three footwear conditions: wearing the two different sandals and barefoot. Clear differences in foot motion while running shod and barefoot were observed. Barefoot running exhibited greater inversion and less eversion throughout the stance phase and greater adduction during the last 20% of stance. Very distinct footwear motions were observed compared to the associated foot motion (Morio, et al., 2009). Sole hardness had an effect on footwear motion; the more pliable midsole allowed foot motion more similar to the barefoot condition. The sandals effectively constrained foot motions in the frontal and transverse plane, but not for the sagittal plane. The sandal also influenced forefoot eversion during the propulsive phase. While running barefoot, the subjects displayed a great range of variability for frontal plane motion: 4 subjects everted, 5 subjects inverted, and one displayed a neutral pattern. While wearing the sandals, 9 of the 10 subjects adopted an inversion pattern during push-off (Morio, et al., 2009).

Shoe type has been identified as a factor in determining plantar pressure while running. It was found that sex plays a role in pressure response to a change in shoe type as well, warranting sex-specific running shoe designs. Researchers have compared plantar pressure in men and women while running in a typical training shoe and a minimal racing flat (Queen, Abbey, Wiegerinck, Yoder, & Nunley, 2010). Thirty-four (17 female, 17 male) subjects were chosen to perform seven 10-meter running trials of both footwear conditions. A Pedar-X in-shoe pressure system was utilized to measure plantar pressure of eight different regions of the foot via an insole placed in each shoe. The eight regions of interest were: rearfoot, medial midfoot, lateral midfoot, medial forefoot, middle forefoot, lateral forefoot, hallux and the lesser toes. Maximum force was greater for the racing flat, 2.60 BW compared to 2.35 BW for the training shoe. Total foot contact area was 92.8% for the training shoe and 90.9% for the racing flat. Women and men also differed in total contact area, having 93.2% and 90.2% respectively. Maximum force and contact area also differed based on foot region, shoe type and sex (Queen, et al., 2010). Training shoes displayed larger forces than racing flats, and men more than women at the lateral forefoot. Men also yielded greater forces than women at the medial midfoot only while wearing racing flats, and at the medial forefoot while wearing training shoes. Consequently, contact area for the medial midfoot and forefoot were greater in the training shoe and the racing flat displayed increased lateral forefoot contact area (Queen, et al., 2010).

The difference in maximal force observed by these researchers was likely due to differences in cushioning and midsole density (Queen, et al., 2010). The medial midfoot region showed a 13% decrease in plantar loading using the racing flat. Also, males showed an increase in plantar loading of the lateral side of the foot, a site which is subject to greater concern regarding stress fracture risk (Queen, et al., 2010). These findings indicated the need for caution while training in racing flats. Men, in particular, may need to consider limiting time spent in racing flats to minimize injury risk (Queen, et al., 2010).

A paired study by the same research group analyzed only the effect of shoe type on pressure distribution (Wiegerinck, et al., 2009). Maximum force, peak pressure and contact area for the training shoe and racing flat were examined for the total foot and the same eight regions described previously. The methodology was identical to the study by Queen et al. (2010), except for a different research question and statistics applied. In short, the racing flat yielded greater maximum force and peak pressure than the training shoe across the total foot (Wiegerinck, et al., 2009). Within regions, the complete forefoot and lateral midfoot had increased peak pressure in the racing flat. Maximum force increased along the lateral aspect and decreased at the medial midfoot. Contact area also increased in the lateral forefoot but decreased in the medial forefoot (Wiegerinck, et al., 2009).

The decreased force at the medial midfoot is characteristic of decreased medial support in the racing flat (Wiegerinck, et al., 2009). In training shoes with increased arch support, medial contact area is increased leading to higher medial peak forces. As alluded to by Queen et al. (2010), this study also pointed towards limiting time in racing flats due to increased lateral forces and risk of stress fractures (Wiegerinck, et al., 2009).

A point did arise regarding higher maximum force in the rearfoot of the training shoe, however, it received very limited discussion by the authors (Wiegerinck, et al., 2009). Kinematics were not recorded for either study which could help to explain the dynamic footstrike pattern of these two shoes. According to investigations of footstrike and impact forces, it seems likely that the greater forces observed in the rearfoot of the training shoe could be, at least partially, due to a more pronounced heelstrike. Having a sense of how kinematics are affected by shoe type could provide important information.

It has also been suggested that plantar pressure data can vary greatly across different footstrikes, and may therefore be an inadequate measure to evaluate kinetics during running (Maiwald, et al., 2008). A study of reproducibility of plantar pressure during barefoot running showed different results for foot region (Maiwald, et al., 2008). Recreational runners with a heel strike pattern ran multiple trials over a pressure plate recording plantar pressure distribution.

Rearfoot pressure yielded worse reproducibility than forefoot pressure indicating greater variability of rearfoot motion in heel strikers. Furthermore, almost all (93 of 95 total) subjects displayed irrelevant pressure loading in the medial midfoot region (Maiwald, et al., 2008). Therefore, interpretation of the findings by Queen et al. (2010) and Wiegerinck et al. (2009) should be considered with caution.

Shod versus barefoot running. McNair and Marshall (1994) performed a study to examine adaptations to different types of footwear. Four different running shoe designs were examined using a materials test and compared to barefoot running on various measures of shock attenuation (McNair & Marshall, 1994). The shoes differed on their midsole characteristics and price, constructed from different forms of double density ethylene vinyl acetate (EVA). Variables of interest during the materials test and dynamic running test were peak acceleration, time to peak acceleration and kinetic energy absorbed. The materials test showed significant differences between each shoe; however, the shoes did not follow a consistent trend for each test. There was a 17% range between shoes for peak acceleration, 30% range in time to peak acceleration, and 4% range in kinetic energy absorbed. Ten subjects were then ran on a treadmill with an accelerometer attached to the distal tibia. Sagittal plane kinematics were also recorded for knee and ankle angles, and all variables were averaged across eight consecutive strides. During the running examination, no differences were observed between shoes for peak acceleration or time to peak acceleration. However, the barefoot condition did yield differences from the shoes. Barefoot running displayed higher peak accelerations and a 48% faster time to peak acceleration suggesting a lessened ability to attenuate shock (McNair & Marshall, 1994). Subtle differences in knee and ankle kinematics were observed for each footwear condition. The most noticeable change was greater plantarflexion held throughout the entire stride cycle of barefoot running (McNair & Marshall, 1994).

These results portray that runners prefer to operate within a certain ‘kinetic bandwidth’ regarding tibial accelerations from impact (McNair & Marshall, 1994). It was suggested that the subtle kinematic changes between conditions were designed to create similar tibial accelerations across shoes with varying cushioning abilities (McNair & Marshall, 1994). Therefore, runners adapt to footwear such that impact shock is held constant. However, given a heelstrike pattern, barefoot running yielded greater impact forces than the system could adapt to and the shock attenuation parameters were altered differently from the shod conditions.

To further examine this concept, an analysis of kinematic and kinetic variables during the stance phase of running was conducted to compare barefoot and shod conditions (De Wit, De Clercq, & Aerts, 2000). Nine healthy distance runners were selected to run across a force platform at three different velocities (3.5, 4.5 and 5.5 m/s) while barefoot and wearing neutral shoes. Running velocities were ensured by timing between infrared photocells placed 5-meters apart and equal braking/propulsive impulses were monitored to maintain constant velocity. Two-dimensional kinematics of sagittal and frontal planes were examined with high speed cameras. An additional test was conducted with a pressure mat placed on top of the force platform to evaluate plantar pressures for the barefoot condition at 4.5 m/s.

Barefoot running showed larger loading rates for the initial vertical ground reaction force than shod conditions at all velocities (De Wit, et al., 2000). Subjects contacted the ground initially with about 14 degrees more plantarflexed position and a greater position of knee flexion. This indicated an adaptation to a flatter foot strategy which also explains findings of decreased plantar pressures at the heel. Barefoot running was also characteristic of higher step frequencies, shorter step lengths, as well as less ankle and knee motion during midstance. This contributed to greater leg stiffness observed during barefoot running (De Wit, et al., 2000). The researchers noted that there were large individual differences in frontal plane kinematics, which may have been limited by their use of two-dimensional analyses.

A study was conducted to determine if cushioning properties of different shoes have an effect on the regulation of leg stiffness during hopping and running (Bishop, Fiolkowski, Conrad, Brunt, & Horodyski, 2006). Adaptations in leg stiffness and running kinematics due to changes in footwear were evaluated. Nine healthy subjects hopped on a force platform at 2.2 Hz for 1-minute under three footwear conditions: barefoot, low-cost shoes, and high-cost shoes. Two-dimensional kinematic data was also collected for the three footwear conditions while running on a treadmill. Limb stiffness was calculated as the peak vertical ground reaction force divided by the displacement of the center of mass during the hopping activity. Stiffness of each shoe was calculated in a similar fashion on a MTS mechanical testing machine. The 2-D kinematic data was used to analyze sagittal plane knee and ankle angles at initial foot strike as well as peak knee and ankle flexion in midstance (Bishop, et al., 2006).

The low-cost shoe was significantly stiffer than the high-cost shoe (426 N/mm compared to 257 N/mm when loaded at 10 mm/s), and the high-cost shoe compressed twice as far when loaded by the machine (22.9 mm vs 11.2 mm). Peak limb stiffness increased when each subject wore shoes. Leg stiffness was significantly greater while wearing high-cost shoes as compared to barefoot (p=.002). A trend existed from barefoot to low-cost to high-cost; however, no statistical difference was reported between barefoot and low-cost (p=.092) or between low-cost and high-cost shoes (p=.091). At initial footstrike, subjects had about 12 degrees greater dorsiflexion while wearing shoes compared to running barefoot. The knee underwent a greater joint angle excursion while wearing shoes, and the ankle had greater joint excursion while barefoot (Bishop, et al., 2006).

Barefoot runners landed with the ankle more plantarflexed and absorbed most of the initial impact at the ankle, while shod runners flexed more at the knee and less at the ankle to absorb the impact (Bishop, et al., 2006). This study also provides evidence that cushioning properties of shoes effect limb stiffness. A less stiff shoe requires a runner to increase leg stiffness by flexing less at the knee and ankle during compression of the leg-surface spring (Bishop, et al., 2006).

Ground characteristics may also affect the leg-surface spring model. Rome, Hancock and Poratt (2008) described how running barefoot on soft surfaces such as sand, grass, or artificial tracks and fields can be very beneficial for healthy foot development. However, doing so harder surfaces like pavement could alter biomechanics, potentially leading to injuries or arthritic complications and culminating in reduced foot function (Rome, Hancock, & Poratt, 2008).

This spring-damper-mass model has been applied to describe the role of footwear and ground composition on force attenuation properties during running (Ly, Alaoui, Erlicher, & Baly, 2010). Ground was illustrated as having viscoelastic properties that change under load. Having softer cushioned shoes or being on a softer, more compliant ground surface greatly influenced the passive impact peak of the vertical ground reaction force. Soft shoes running on a surface with very low stiffness were able to effectively eliminate the initial passive impact peak. The authors viewed ground characteristics as having a critical role in controlling shock attenuation and should be considered in determining shoe cushioning design (Ly, et al., 2010).

Leg and foot stiffness were also examined while barefoot walking and running across four surface mats of different compliances (Wilson & Rochelle, 2009). The study was designed to describe the optimal surface compliance for damping abilities during walking and running (Wilson & Rochelle, 2009). While walking, vertical ground reaction forces were best reduced using the least rigid surface, which had a compliance of 334 kN/m. However, during running, the vertical forces were reduced better with a surface of intermediate rigidity (compliance of 1020 kN/m). These findings suggest that reductions of impact loading while barefoot running act differently from walking and that a super-cushioned surface does not provide the optimal benefits (Wilson & Rochelle, 2009).

Variations in ground surface have been linked to changes in impact forces experienced while running. Plantar pressure distribution has been studied in runners while negotiating different surfaces. Recently, peak pressure, contact area, and contact time while running on natural grass and asphalt were compared in recreational shod runners (Tessutti, Trombini-Souza, Ribeiro, Nunes, & Sacco Ide, 2010). Forty-four subjects ran 40-meters across the two surfaces at 3.33 m/s while donning Pedar-X pressure measurement insoles. The feet were divided into six regions and analyzed across conditions. While running on asphalt, peak plantar pressure was significantly greater in the central and lateral rearfoot as well as the lateral forefoot, increasing by 12.7, 12.2, and 10.8% respectively. Accompanying the increase in pressure was a 12.7% decrease in contact area and 12.1% decrease in contact time for the central rearfoot. These results indicate greater loads during rigid surface running, especially on the lateral aspects of the foot (Tessutti, et al., 2010). Due to the associated decrease in contact time, the authors speculated a change in kinematics (substantiated by previous researchers). With less time on the ground comes a reduced ability to absorb pressures, leading to a less mobile foot/ankle complex (Tessutti, et al., 2010). This could explain why the pressures were increased in particular on the lateral side; when wearing shoes the foot stays rigid and provides less impact-absorbing eversion.

With a decrease in contact time less attenuation of plantar pressure occurs helping to explain the higher impact forces previous researchers have cited from running on less compliant surfaces.

Kurz and Stergiou (2004) examined how hardness of shoe midsoles affect ankle coordination during the stance phase of running. Two running shoes of similar characteristics except midsole hardness were used (Brooks Beast and Adidas Response Cushion) in conjunction with a barefoot trial to make three footwear conditions (Kurz & Stergiou, 2004). Eight recreational runners were recruited to run on a treadmill for each condition while sagittal and frontal plane kinematics were collected. Each stance phase was separated into absorption and propulsion phases respective of the peak knee flexion observed during stance. A dynamic systems theory was used to examine ankle coordination whereby phase angles were calculated from plot of segment angular displacement by angular velocity (Kurz & Stergiou, 2004). Phase angles close to zero indicated that the foot and ankle were in-phase and more coordinated. Phase angles closer to 180 degrees were considered out-of-phase and less coordinated (Kurz & Stergiou, 2004).

There were no differences in any measures of phase angles or coordination between the two shoe conditions, but barefoot was significantly different from both shoes (Kurz & Stergiou, 2004). The sagittal plane foot-leg examination showed a more uncoordinated out-of-phase relationship for the barefoot condition during absorption. During propulsion, this reversed and the barefoot condition remained more in-phase. The frontal plane data revealed the barefoot condition to be more in-phase and more coordinated between the foot and leg during absorption and propulsion. The results indicate that the musculature surrounding the ankle must adopt a more coordinated strategy for barefoot running when impact forces are increased (Kurz & Stergiou, 2004). The less in-phase for sagittal absorption reflects a more plantarflexed footstrike and the more in-phase for the frontal plane shows more coordination regarding inversion and eversion. These findings match those of previous studies that barefoot running relies less on the subtalar joint and more on ankle dorsiflexion to absorb impact forces. These authors speculated that these coordination strategies are beneficial in their use of larger plantarflexor and dorsiflexor muscles to adapt to shock during barefoot running as opposed to relying on frontal plane dynamics (Kurz & Stergiou, 2004).

A longer comparison between barefoot and shod running was performed by evaluating 60 consecutive steps on a force platform instrumented treadmill (Divert, et al., 2005). Thirty-five recreational runners completed 2 bouts of 4-minute running at 3.33 m/s separated by 2 minutes of recovery. The running bouts were randomly assigned as barefoot or wearing neutral shoes. Passive and active peaks were evaluated on the vertical ground reaction force while braking and pushing peaks were examined for the anterior-posterior ground reaction force. Surface EMG of five shank muscles (tibialis anterior, peroneus longus, lateral gastronemius, medial gastrocnemius, and soleus) was also evaluated.

Stride duration, contact time and flight time were all significantly lower during the barefoot condition (Divert, et al., 2005). Passive and active vGRF peaks were 14.9 and 4.6% smaller and vertical impulse was 0.3% lower when barefoot. Although, the braking and pushing forces and impulses were all significantly higher while running barefoot. EMG data revealed an increased pre-activation of the plantarflexor muscles prior to foot strike. The lateral gastrocnemius, medial gastrocnemius and soleus respectively had 13.7, 23.6 and 10.8% more activity without shoes. No significant differences in EMG amplitude were discerned in any other phase of running between barefoot or shod (Divert, et al., 2005).

This study refuted findings of previous research that barefoot running yields higher vertical ground reaction forces when subjects run across a stationary platform (Divert, et al., 2005). The authors declared that when examined across 60 consecutive strides, subjects must adapt by decreasing the mechanical loading of repetitive impacts (Divert, et al., 2005). In doing so, runners reveal lower impact peaks and impulses as well as decreased temporal variables to accommodate to prolonged barefoot running.

Most previous research examining footstrike patterns of barefoot running was conducted on subjects who normally run in shoes. Therefore, Lieberman, et al. (2010) compared footstrike patterns of habitual barefoot and shod runners. Subjects were selected to fill five groups of 16 participants each. The groups consisted of: habitually shod runners, athletes who grew up barefoot but now use cushioned shoes, runners who grew up shod but now habitually run barefoot, children who have never wore shoes, and children who have worn shoes most of their lives.

Individuals who grew up running in shoes adopted a rear foot strike pattern, whereas those who grew up running barefoot, or switched to habitual barefoot styles tended to run with a forefoot strike pattern (Lieberman, et al., 2010). These forefoot strikes were characteristic of a toe-heel-toe pattern where subjects would land on their forefoot, compress to heel touchdown in stance and then push off from the forefoot. A major difference in vertical ground reaction forces was observed as result of footstrike patterns. Heelstriking, whether shod or barefoot, resulted in an initial vGRF peak at about 6.2% of stance time from impact with a very high loading rate. The toe-heel-toe running had no initial peak at this point and load rates were far more gradual throughout contact. It was explained how during the forefoot strikes translational kinetic energy from the leg was converted into rotational energy due to the initial toe-to-heel motion and controlled by the gastrocnemius and soleus muscles (Lieberman, et al., 2010). In contrast, heel-striking requires the lower extremity to absorb the full impact of that translational energy leading to the higher forces assumed.

The running shoes were identified as a major contributor to rearfoot striking patterns (Lieberman, et al., 2010). Due to the high heel cushion, running shoes provide a slant that orients the foot in 5 degrees less dorsiflexion than the sole of the shoe (Lieberman, et al., 2010). Effectively, modern running shoes facilitate heel striking by increasing the ankle dorsiflexion angle at contact.

One common component of running shoe design is the incorporation of cushioning to reduce the load of impact. Some studies examining shoes of varying compliance yield results that favor this notion of cushioning for impact control; however, other studies have shown that barefoot running can produce lower forces than those of shod running. Robbins and Gouw (1991) speculated that this could be due to neuro-regulatory mechanisms in which plantar tactile stimuli can influence motion to reduce impact forces. They believed that the sole in shoes attenuates sensations that would otherwise tell runners to ease up or alter gait such that impact could be reduced. To test their hypothesis, 20 participants were subjected to vertical and horizontal plantar loads using three different surfaces. The loads were held constant by constraining the subjects to an apparatus and normalized to each individual's plantar contact surface area. The surfaces used consisted of: smooth rigid acrylic plastic, ultra high molecular weight polyethylene (UHMWPE), and a rigid textured surface with 2mm irregularities. Subjects were instructed to provide a discomfort rating to each condition on a numerical pain scale. When vertical and horizontal impacts were below 0.4 kg/cm^2 there were no significant differences in discomfort observed between surfaces. When impacts were at or above 0.4 kg/cm^2 significant effects were found. At the higher loading, the irregular surface proved to be the most uncomfortable. The horizontal load was a critical component in the impact testing as was the irregular surface in inducing subcutaneous nociceptor stimuli. Cushioned shoes provide protection from both of these factors, allowing the runner to experience more plantar comfort (Robbins & Gouw, 1991). With the increased comfort, the runner is less persuaded by plantar sensations to reduce vertical impacts, explaining higher forces while running shod. Without adequate feedback from plantar nerves, less behavioral adaptations result, leading to greater assumed impacts and injury risk (Robbins & Gouw, 1991).

By decreasing foot pain linked with poor technique, shoes can actually perpetuate poor mechanics by limiting self-correcting behaviors and ultimately leading to injuries (Vormittag, Calonje, & Briner, 2009). Through an alteration in running mechanics, problems can occur up the kinetic chain and lead to a multitude of injuries (Vormittag, et al., 2009). Shoe manufacturers are currently marketing new designs with limited support and cushioning which claim to strengthen feet and decrease injury potential.

Reports on populations who habitually run barefoot indicate lower injury rates than their shod counterparts (Lieberman, et al., 2010; Vormittag, et al., 2009). Admittedly, there can be many explanations for this phenomenon; however, it is peculiar since the primary marketing point for shoes are protection, support and injury prevention.

Robbins and Hanna (1987) proposed that the ability of the foot to absorb shock is related to deflection of the medial longitudinal arch during loading. Furthermore they claimed that shoes increase the rigidity of the foot and disallow the arch to function properly, leading to higher injury rates (Robbins & Hanna, 1987). Associated with this theory was the premise that flat low arches can be a result of weak intrinsic muscles in the feet. Habitually wearing shoes excessively supports the foot and leads to atrophy of these muscles. These researchers set out to examine the effects on integrating barefoot activities to alter foot type (Robbins & Hanna, 1987). Seventeen recreational runners were tested for medial arch length by lateral radiograph under two partial loaded conditions of 15kg and 55kg. The subjects were required to maintain a detailed daily log recording running history, footwear, and barefoot weight-bearing activities for 4 months. They were instructed to gradually increase barefoot weight-bearing activities throughout the intervention, walking and running encouraged. Radiographs of the medial arch were recollected at one month intervals until the conclusion of the study. A significant shortening of the medial longitudinal arch was observed following treatment, describing a 4.7mm mean decrease. Associated with shortening of the arch was a successful heightening as well. These results identified the ability to rehabilitate the intrinsic foot musculature (Robbins & Hanna, 1987). The activation of such muscles can allow the foot to act more as a dynamic impact dampening structure while sparing stress on the plantar fascia. The findings from this study unveil the possibility that many running related injuries may simply be solved by promoting barefoot activity (Robbins & Hanna, 1987). Since such behavior may be impractical due to social restrains and extreme temperatures and terrains, research into designing minimalist footwear that activate and sustain intrinsic muscles is warranted.

A minimalist lightweight shoe was designed by Vibram to mimic the experience of being barefoot. Squadrone and Gallozzi (2009) studied the effectiveness of this shoe in imitating barefoot running regarding spatiotemporal variables, pressure distribution, kinematics, and economy. The Vibram Fivefinger was compared to a typical training shoe and barefoot conditions on an instrumented treadmill for 8 experienced barefoot runners. The athletes ran for 6 minutes in each shoe at 3.33 m/s to allow assessment to be based off a large number of consecutive steps. Stride length was significantly lower and stride frequency was higher while barefoot with no difference between shod conditions. Although, almost all other variables tested were different in the training shoes from the other two conditions and no difference was observed between barefoot and Vibram. The training shoe showed higher impact forces, oxygen consumption, and pressures under the heel, midfoot and hallux. The subjects landed more dorsiflexed in the training shoes and had less ankle motion throughout stance absorption, confirming findings of previous works. The Vibram Fivefinger did not differ from the barefoot condition with respect to all these variables. Peak vertical impact was lower in the Vibram than training shoe (1.59 compared to 1.72 BW). Furthermore, the sagittal kinematics and oxygen consumption closely mimicked barefoot running. The Vibram shoe required 2.8% less VO_2 compared to the training shoe indicating an improved economy likely attributable to weight reduction. These authors indicated that the Vibram Fivefinger successfully imitated barefoot running while providing a small amount of protection to the plantar surface (Squadrone & Gallozzi, 2009). This design likely satisfies the request of Robbins and Hanna (1987) to create a shoe capable of utilizing intrinsic musculature while shielding the foot from hazardous elements.

Efficacy and practicality of barefoot running. It appears from the literature that the forces experienced during running depend on many factors including surface compliance, footwear properties, anatomical structure, footstrike patterns, and running speed. Barefoot running seems to be a growing trend among previously shod populations. Although some researchers may be proponents of moving exclusively to barefoot, based on this review most might agree that such an impetuous act would be ill-conceived and could result in injury. The more sensical approach may be to use barefoot activities as a tool to strengthen feet and potentially limit injuries, while maintaining the majority of training volume in minimal shoes, lightweight trainers or racing flats.

Warburton (2001) outlined the efficacy of barefoot running associated with reductions in acute and overuse injury risk. Inversion sprains, which make up a disproportionate majority of acute running injuries, could in fact be amplified by shoe use. The soles of running shoes may limit the proprioceptive ability of the foot as discussed by Robbins and Gouw (1991), which could lead to decreased control of the ankle. Furthermore, Warburton (2001) theorized that the additional height of a running shoe provides a greater leverage arm to rotate about the subtalar joint in the event of a sprain, consequently increasing the torque and potential damage experienced.

Chronic injuries may also be reduced as a result of barefoot adaptations. Plantar fasciitis is caused from inflammation of the plantar fascia, which typically supports the medial longitudinal arch. In conjunction with Robbins and Hanna's (1987) finding that barefoot activities can strengthen intrinsic muscles of the feet, it is postulated that the plantar fascia could be spared and as the arch is depressed some of the load is transferred to other local musculature. Other overuse injuries such as ilio-tibial band syndrome, shin splints and patello-femoral pain may be reduced due to the kinematic changes associated with barefoot running. Multiple researchers have cited greater plantarflexion upon landing without shoes. Following contact, increased dorsiflexion and decreased excessive eversion have been observed throughout stance. As identified by Kurz et al. (2004), the change in barefoot kinematics allows for a more coordinated shank-foot strategy during stance and propulsion. These changes allow each impact to be absorbed in a different manner that help explain the smaller vertical forces observed with barefoot running and suggest decreased risk of injury.

Thirty minutes of weight-bearing barefoot activity is recommended as an appropriate starting point to strengthen intrinsic muscles, ligaments and adapt to a less dependent foot structure (Warburton, 2001). Increases in duration, intensity and surface compliance should be implemented gradually. After about 4 weeks, the skin on the plantar surface adapts to allow longer periods of running without receiving cuts, bruises or blisters (Warburton, 2001). Following 4 months of barefoot training, the medial longitudinal arch may begin to shorten and heighten with hypertrophy of the intrinsic muscles (Robbins & Gouw, 1991). Progressive activities to increase foot strength may include active inversion and toe flexion exercises as well as walking on the balls of the feet and running on compliant surfaces (Warburton, 2001).

In this case, barefoot activities can be considered a supplement to a run training regimen. Practical applications could involve runners applying a small percentage (5-10%) of their volume to running without shoes. Running on mid-level compliance surfaces such as artificial grass or turf may be best suited for these activities, and individuals should refrain from running fast while barefoot. Such practices could help to develop intrinsic muscles in the feet to help support the arch while most of the training can still be performed in lightweight training shoes or racing flats.

Summary

Distinct differences in research findings have been reported regarding the efficacy of barefoot running. Many studies indicate increased impact forces and risk of injury during barefoot conditions. These researchers praise running shoes for their ability to cushion and stabilize the foot to attenuate forces experienced. However, questions have been raised about these studies regarding methodological limitations. More recently, researchers have shown the necessity to average a large number of consecutive strides while assessing changes in running kinematics, pressures or impact forces. Studies that examine a greater number of strides yield opposite results that are actually in favor of barefoot running. Furthermore, footstrike patterns may play a critical role in how forces are transmitted through the kinetic chain. Previously described heelstrike patterns may be best suited for shod running, where as barefoot running may be served by forefoot strike styles.

William Jungers (2010) cited that most runners today strike the ground initially with their heel in a rear foot strike pattern. Barefoot runners adapt their footstrike based on conditions, but typically land in a midfoot or forefoot strike pattern. Forefoot strike runners also tend to have shorter stride lengths and increased leg compliance, reducing the high transient force of repeated impacts. Modern running shoes are elevated and provide a cushioned heel designed to attenuate forces from impacts during heelstrike running (Jungers, 2010). Although there are clear debates and anecdotal testimonials to both sides, there is no definitive evidence that cushioned running shoes either prevent or cause musculoskeletal injuries.

Few studies have followed a longitudinal design examining long term use of barefoot activities in previously shod populations. Robbins and Hanna (1987) showed a change in arch length with increases of barefoot activities and credited improved activation of intrinsic muscles; however this was only an assumption as soft tissues were not examined by their imaging. Further research is warranted to describe the effects of long-term barefoot running on intrinsic foot musculature development as well as overuse injury prevention.

Before such studies can take place, it is important to understand how global fatigue from prolonged running affects running mechanics in a barefoot individual. Previous research has indicated that habitual shod runners are less adapted to barefoot running than habitual barefoot runners. Prior to examining habitual shod runners in barefoot studies or entering them into barefoot training regimens, it should first be understood how fatigue affects these individuals with and without shoes. This study addresses these concerns by examining changes in shock attenuation and muscle activation before and after fatigue while running barefoot and shod.

Chapter III

Methods and Procedures

Introduction

In accordance with the research question, a study was conducted to investigate the difference in fatigue response as measured by muscle activation and shock attenuation between barefoot and shod running immediately following a fatiguing run protocol. Tibial accelerations were evaluated to discern changes in shock attenuation ability between conditions. Pre-activation and timing of tibialis anterior and medial gastrocnemius muscle activations were analyzed to identify alterations in coordinative strategies and running mechanics following the fatigue protocol.

Description of Study Population

Twelve moderately-trained male and female runners were recruited from the community via verbal solicitation and recruitment flyers posted at a local running retail store. All subjects had at least 4 years of experience training and racing recreationally, and were free of any neurological disease which could impact the EMG measurements. Subjects were also reportedly injury free for the 3 months prior to data collection, ensuring sound natural biomechanics in running gait.

Design of the Study

A repeated measures observational study was employed to examine fatigue effects in two footwear conditions. Running gait parameters were examined pre- and post-fatigue and between barefoot and shod conditions. Each subject performed both conditions in random order, assignment as determined by a coin flip. The two conditions evaluated were running shod before and after a fatiguing run (SHD) and running barefoot before and after a fatiguing run (BFT).

Both fatigue protocols were performed shod, due to ethical considerations regarding prolonged barefoot running in unaccustomed individuals.

Data Collection Procedures

Instrumentation. Subjects were fixed with a tri-axial 10G G-link accelerometer (MicroStrain Inc., Williston, VT) on the anterior-medial aspect of the distal right tibia. Double-sided tape and Coban® light-elastic athletic tape were used to adhere the accelerometer to the leg. Surface electrodes (3M, St. Paul, MN) were placed over the center of the muscle bellies of the medial gastrocnemius and tibialis anterior muscles. Electrode placement sites were shaved with a disposable razor and cleansed with alcohol wipes to reduce the electrical resistance of the skin. Two electrodes were used for each muscle, centers spaced 2 cm apart and parallel in line with the direction of the muscle fibers. Wired leads were connected to the electrodes, bundled and affixed to the leg at various points with prewrap to reduce artifact noise. The leads were connected to a small BTS PocketEMG unit (BTS Bioengineering, Garbagnate Milanese, MI, Italy) donned by the subject integrated with a belt worn around the waist. EMG data were collected on board the mobile unit and later downloaded to a working computer via BTS EMG analysis software. The accelerometer data were collected separate from the EMG data through a laptop computer and wireless transmission device placed in the center of the collection field. The two systems were time synchronized by the use of a footswitch integrated with the BTS equipment.

Prior to removing the equipment following the first condition, the electrodes were traced with a pen to document proper placement during the second condition. This reduced the likelihood of measurement error regarding difference in electrode placement.

Measurement techniques and procedures. Data for the two conditions were collected on different occasions separated by 3-7 days. The conditions were performed in random order by each subject as determined by a coin flip. Each collection day commenced with a 5-minute warm up consisting of a light jog performed around a 400-meter outdoor running track. This warm up protocol has shown to increase the accuracy of EMG measurements by increasing blood flow (De Luca, 1984). After warm up, an experienced researcher applied electrodes and the accelerometer on the subject as described in the instrumentation. Subjects then entered the protocol for one of two footwear conditions.

In the BFT condition, subjects were instructed to run barefoot for a quarter lap along the track (100 meters) while EMG and accelerometer data were collected. The intensity for this instrumented run was set as a comfortable training pace, self-selected by each individual. Time to completion was monitored and the subject was instructed to pace the run as evenly as possible. The participants only used a straight section of track for the 100-meter runs to limit alterations in running mechanics that may be imposed by a change of direction. Afterwards, the PocketEMG unit, accelerometer and wire leads were removed. The electrodes were left on the skin to avoid measurement error in replacement precision, but were traced in case replacement was necessary. Then the subjects were instructed to run a fatiguing 30-minute run in their typical training shoes at a high intensity self-selected by each individual. The goal assigned was to complete as many laps as possible in 30 minutes. Participants were verbally encouraged by the research team during the run and allowed to monitor their running time. Immediately following the fatiguing run, subjects were quickly re-instrumented with the accelerometer, wired leads and PocketEMG unit. After removing their shoes, they were instructed to run barefoot once again for a quarter lap (100 meters) along the track while data was recorded. Intensity for the second barefoot run was set equal to the comfortable pace self-selected by each individual before the fatiguing run. As before, the subject was able to monitor running time and pace evenly and closely to the pre-fatigue run. Prior to all 100-meter collection trials the subjects were instructed to step on a footswitch placed on the ground that was integrated with the PocketEMG unit. This allowed the EMG and accelerometer data to be time synchronized later in the data processing phase. Time to completion was monitored for both pre and post-fatigue 100-meter runs and compared to establish consistency in running speed. Since the subjects were experienced runners, they were able to accurately pace themselves throughout each measured run. In the event that the running times were not consistent, such occurrence was reported in the results and later discussed as a limitation of the study. Only subjects who achieved a post-fatigue run within ± 5% of their pre-fatigue run time were included in data analysis.

The protocol for the SHD condition was identical to the BFT condition, except all running was performed in the subjects' typical training shoes. Electrode and accelerometer placement were performed by the same researcher and guided by pen markings placed on the subject from the previous trial. Following the last trial in both conditions all equipment was removed and subjects were encouraged to perform a light jogging cool down and stretch to individual satisfaction.

Data processing. Electromyography and accelerometer data were processed and filtered using Excel® 2007 (Microsoft Corp., Bellevue, WA). They were time synchronized by matching the footswitch signal to the first significant acceleration spike. Footstrike and the support phase were identified from the accelerometer signal and used to derive stride time across each gait cycle in all collections. Eight consecutive gait cycles were selected for analysis that coincided approximately with the 50-meter mark of each collection trial. This point of each run was estimated as 50% of the time to completion for its respective 100-meter trial. This mark was chosen in attempt to eliminate changes in acceleration and was evaluated in an inconspicuous manner unbeknownst to the subject.

The peak impact acceleration was identified for each gait cycle. The rate of positive acceleration was calculated from the first value above 1g to the last point before the actual peak. A similar method has been used by researchers to calculate average vGRF loading rates from 20% to 80% of the peak to ensure that the calculated slope was mostly flat (Milner, Ferber, Pollard, Hamill, & Davis, 2006). All acceleration rates for the eight gait cycles were averaged for each subject to obtain an average peak tibial acceleration rate at impact for each BFT and SHD trial.

EMG data was integrated by time and normalized to stride duration for each gait cycle to obtain the iEMG. The iEMG and acceleration data were then averaged over the eight strides. To obtain the iEMG, the raw EMG signal was rectified by taking the absolute value, low-pass filtered at 20 Hz, and integrated by time. The iEMG for the 50ms prior to footstrike was analyzed as pre-activation for each muscle and 50ms following footstrike was evaluated as the beginning of the support phase. This method was demonstrated by Nigg, Stefanyshyn, Cole, Stergiou, and Miller (2003) to be a valid temporal measurement for pre- and post-activation during running.

Training procedures. The two conditions, performed in random order by each subject, were separated by a 3-7 day recovery period. Subjects were instructed to perform their regular training activities throughout the entirety of the study with no increases or decreases in intended training volume. However, subjects were instructed to refrain from competing in any races and to refrain from physical activity 24 hours prior to each collection in effort to minimize confounding fatigue.

Data Analysis

Three two-by-two repeated measures ANOVAs were used to analyze the effect of condition (BFT vs. SHD) and time (pre- and post-fatigue) on preactivation iEMG of the medial gastrocnemius and tibialis anterior muscles and the rate of acceleration of the shank at ground contact. In the event that a significant difference in preactivation iEMG was observed, a percent change from pretest to posttest was also calculated for that muscle using a two-tailed t-test. Statistical analyses were conducted using the Statistical Package for Social Sciences (SPSS) version 17.0 (SPSS Inc., Chicago, IL). From an alpha level of 0.05, a Bonferroni correction was applied when subsequent tests of simple effect were conducted in the event of a significant interaction. In such case, statistical significance was accepted at $p < 0.025$.

Chapter IV

Results and Discussion

Introduction

The purpose of this study was to examine how alterations in shank muscle coordinative strategies can be effected by the occurrence of global running fatigue. In addition to muscle activity, the average rate of tibial acceleration was obtained such that the effect of these changing muscular strategies on the rate of impact loading shock could be investigated.

Results

Subject Characteristics. One subject was unable to complete the second condition due to injury unassociated with the study and was subsequently excluded from data analysis. The remaining 11 subjects (6 male, 5 female) were 39.0 ± 14.6 years old. They were 1.73 ± 0.07 meters tall and weighed 67.0 ± 13.5 kg. The participants also reported having 19.3 ± 14.2 years of running experience and were currently running 49.0 ± 21.4 km per week.

Running Performance Variables. The 100-meter instrumented run trials were performed by the subjects in 24.99 ± 4.10 sec. All subjects demonstrated good running consistency as each of these 100-meter run trials were completed within 5% of each individuals' mean run time.

Additionally, the participants ran a similar distance for the two 30-minute maximal run efforts demonstrating an equally relative level of fatigue. Subjects ran 6.92 ± 0.98 km for the SHD condition and 6.94 ± 0.93 km for the BFT condition.

Rate of Tibial Acceleration. No significant interaction between footwear and time on the average rate of peak tibial acceleration at impact occurred ($F[1,10] = 0.107$, $p = 0.751$). There was a significant main effect of footwear on acceleration rate ($F[1,10] = 46.476$, $p < 0.001$), but no significant effect of time ($F[1,10] = 0.034$, $p = 0.857$). The barefoot condition presented greater mean acceleration rates than the shod condition, but neither changed over time (Figure 1).

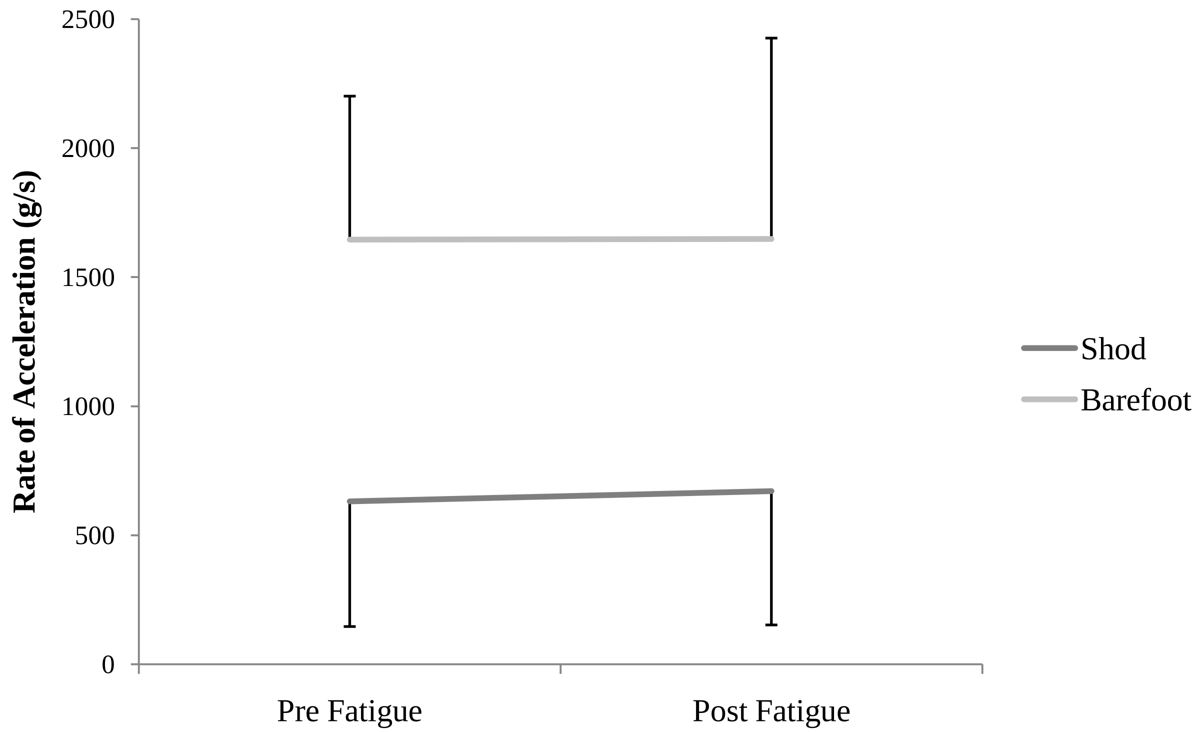

Figure 1. Mean (± SD) rate of tibial acceleration before and after fatiguing protocol for shod and barefoot conditions.

Medial Gastrocnemius iEMG. No significant interaction was found between footwear and time on the average preactivation iEMG of the gastrocnemius (F[1,10] = 0.900, p = 0.365). Additionally, there were no significant main effects of footwear (F[1,10] = 4.506, p = 0.060) or time (F[1,10] = 0.108, p = 0.749) on gastrocnemius iEMG. Mean gastrocnemius iEMG did not change over time or between condition (Figure 2).

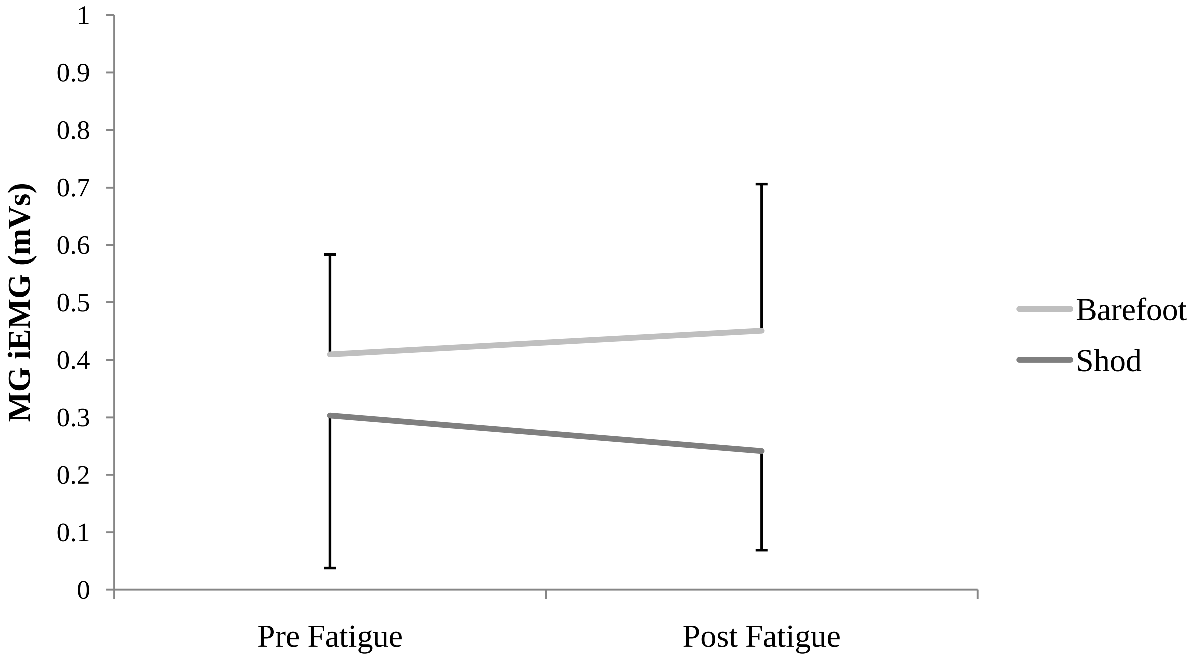

Figure 2. Mean (± SD) medial gastrocnemius iEMG before and after fatiguing protocol for shod and barefoot conditions.

Tibialis Anterior iEMG. A significant interaction was indentified between footwear and time on the average preactivation iEMG of the tibialis anterior ($F[1,10] = 6.339$, $p = 0.031$). The simple effects assessed showed significant differences between footwear within the pretest ($F[1,10] = 47.58$, $p < 0.001$) and the posttest ($F[1,10] = 35.45$, $p < 0.001$); as well as between time within the shod ($F[1,10] = 36.61$, $p < 0.001$) and barefoot ($F[1,10] = 42.90$, $p < 0.001$) conditions. Analysis of these effects showed that tibialis anterior preactivation iEMG increased after fatigue in the SHD condition and decreased after fatigue in the BFT condition (Figure 3).

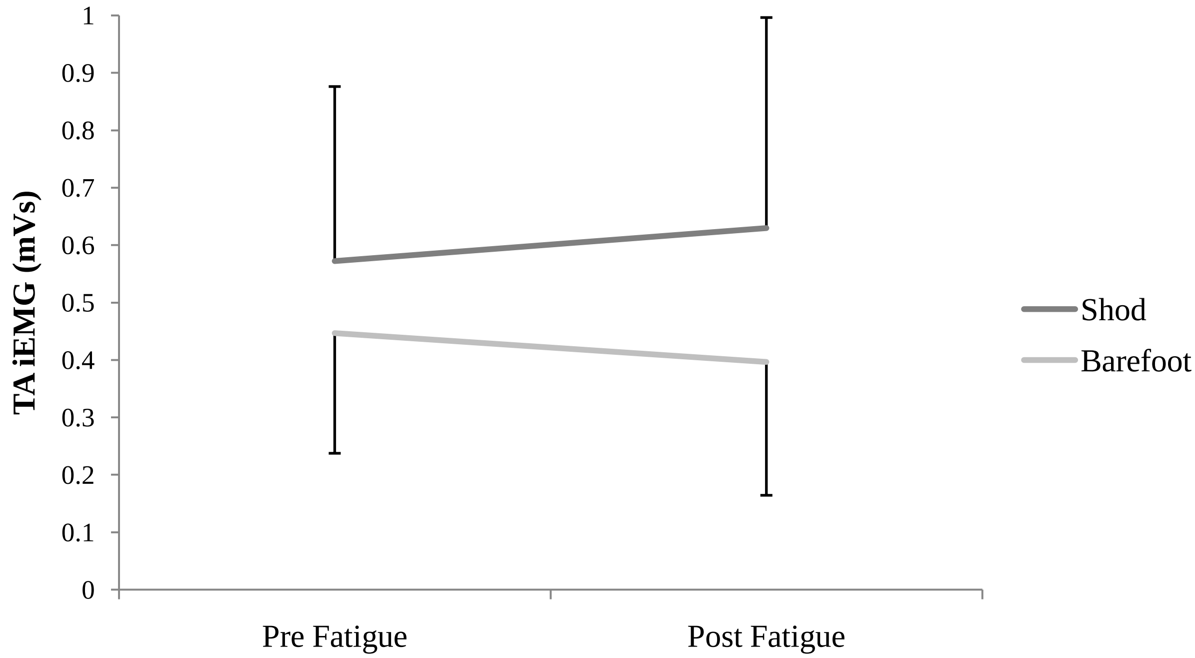

Figure 3. Mean (± SD) tibialis anterior iEMG before and after fatiguing protocol for shod and barefoot conditions.

A percent change in tibialis anterior iEMG from pretest to posttest was also calculated to depict the difference between conditions. A significant difference in percent change of iEMG was observed for this muscle ($t[10] = 2.943$, $p = 0.015$). In the SHD condition TA preactivation increased by 8.66% whereas, in the BFT condition, it decreased by 12.54% (Figure 4).

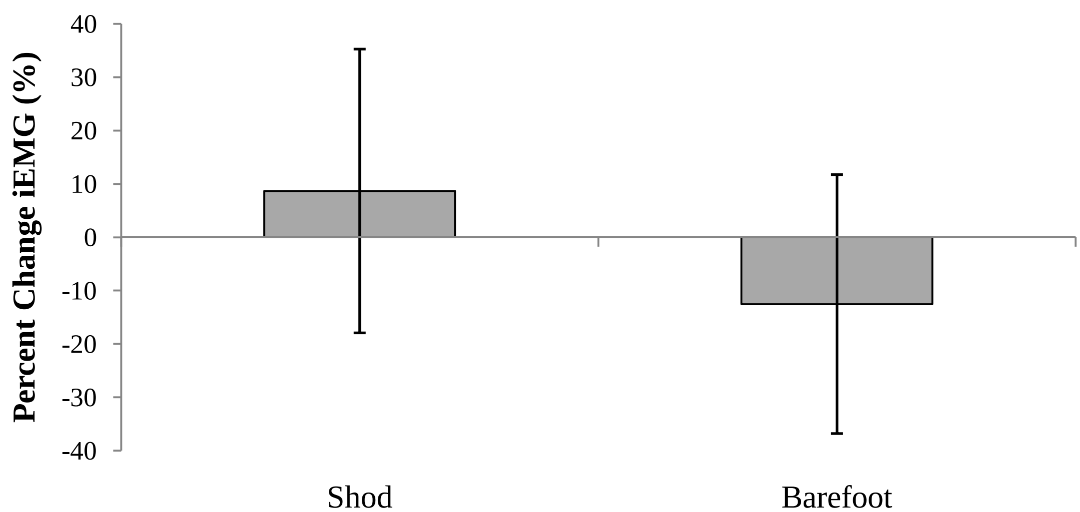

Figure 4. Mean (± SD) percent change in tibialis anterior iEMG following the fatiguing protocol for shod and barefoot conditions.

Discussion

In this study, we set out to investigate whether differences in fatigue response on running mechanics occurred between shod and barefoot running. Habitual shod runners were used and instructed to run 100 meters at a comfortable typical training pace for each of the instrumented running trials. The hypothesis included the concept that different muscle coordinative strategies would be present in barefoot and shod running and that these strategies would be impacted differently with fatigue. Therefore, it was expected to observe an interaction between footwear and time for the preactivation iEMG of both muscles examined. Additionally, it was anticipated to find a main effect of footwear on the rate of tibial acceleration, where the BFT condition would yield higher acceleration rates than the SHD condition. Furthermore, we expected to see an interaction effect on the rate of acceleration, in that greater increases to this measure would be present in the BFT condition than in the SHD condition.

As anticipated, a difference was identified in the average rate of tibial acceleration due to footwear condition. Acceleration rates were higher in BFT than in SHD (1647 g/s compared to 651 g/s for BFT and SHD respectively). No significant effect of time was observed, showing the runners' abilities to maintain the same rate of shock attenuation from pre to post-fatigue. These results only partially supported the hypothesis regarding the rate of shock attenuation. Rate of tibial acceleration has not been previously described in studies of this nature. Rather, only peak tibial acceleration has been assessed. The parameter was chosen for the current study to help examine the rate at which impact shock was experienced, much the same as a loading rate for ground reaction forces.

Also as expected, a significant interaction between footwear and time was found for the average preactivation iEMG of the tibialis anterior. Our results demonstrated that the TA iEMG increased by 8.66% in the SHD condition and decreased by 12.54% in the BFT condition following the fatigue protocol. The study by Nigg et al. (2003) cited a 3.2% difference in TA preactivation as being significant for running conditions between two types of shoes. Although their study did not examine fatigue and utilized a root mean square method of EMG analysis, the methodology of calculation and comparison was very similar to that of the present study (Nigg, et al., 2003). Nigg et al. (2003) assessed the change in preactivation EMG for a single observational running study comparing two shoes of varying midsole compliance. In a similar fashion, two footwear conditions were assessed in the current study as well, but fatigue was also induced which may help to explain the difference between the two studies.

Contrary to the hypothesis, the medial gastrocnemius did not display an interaction between footwear and time for the average preactivation iEMG. No significant main effects were observed for this muscle. These results portray the gastrocnemius acting in a similar manner regarding fatigue response within the two conditions, SHD and BFT.

Many potential methodological limitations may explain why an interaction was not observed for the gastrocnemius. When the means were plotted by condition over time, the lines were distinctly non-parallel (Figure 2), however the large standard deviation of the sample values prevented the finding from having any statistical significance. It seems apparent that the lack of power created from a small sample size (n=11) might have impacted these results.

Furthermore, although many acknowledged researchers have used the medial gastrocnemius to assess EMG in running mechanics (Hanon, et al., 2005; Kellis & Liassou, 2009; Mizrahi, et al., 1997; Nigg, et al., 2003), the gastrocnemius may not have been the appropriate muscle to examine for plantarflexion since it also acts to control motion about the knee. Of six lower extremity muscles assessed, Hanon et al. (2005) found the gastrocnemius to be the only fatigue resistant muscle throughout a multistage graded exercise test. Additional research has indicated the role of gastrocnemius as a primary muscle in power production, whereas the soleus acts more as a postural muscle (Hamill & Knutzen, 2009). For these reasons, it is possible that the gastrocnemius and soleus muscles were responding differently to fatigue between the two conditions. In the current study, instrumenting the soleus would have interfered with the materials used to adhere the accelerometer to the tibia. Unfortunately, attempting to instrument smaller plantarflexors such as the tibialis posterior would have been subject to confounding muscular crosstalk with surface electrodes (De Luca, 1984).

The findings are well supported by the literature and discussed in the preceding review. Elliot and Roberts (1980) noted distinct kinematic changes in running mechanics due to the onset of global fatigue. At the end of a 3000-meter race, runners displayed an increased forward trunk lean and less efficient footstrike placement, resulting in higher shock and braking forces.

Accompanying these changes, higher peak tibial accelerations have been observed in fatigued runners demonstrating an inability to adequately attenuate impact shock (Mercer, Bates, et al., 2003; Mizrahi, et al., 2000). Thigh and leg muscle fatigue have long been suspected in playing a role in increased tibial shock. However, findings by Flynn et al. (2004) and Mizrahi et al. (1997) point to changes in muscular coordinative strategies as the primary factor. Therefore detriments to shock attenuation in running may not be due to specific muscle fatigue, but rather a lack of proper coordination (Flynn, et al., 2004).

Investigations comparing shod and barefoot running mechanics have resulted in different conclusions based on study methodology. Studies examining a small number of strides or using a heelstrike method yield higher vertical ground reaction force (vGRF) loading rates and peak tibial accelerations while running barefoot as compared to shod (De Wit, et al., 2000; McNair &

Marshall, 1994). More recently, Divert et al. (2005) concluded that runners will adapt better to barefoot running if performed over a greater length of time. When measured over 60 consecutive strides, runners displayed lower vGRF peaks during a barefoot condition as compared to shod (Divert, et al., 2005). Furthermore, Lieberman et al. (2010) demonstrated how a forefoot strike pattern in barefoot runners can also reduce vGRF peaks and loading rates as opposed to a heelstrike typically facilitated by shoes.

The current study did not account for footstrike pattern or assess kinematics which may limit our ability to conclusively explain the results; however, we are able to address our primary research question. The findings support the conclusion that barefoot and shod running require different muscular coordination strategies to adequately attenuate impact shock. The coordination strategy employed during shod running accounts for increased tibialis anterior activity with fatigue which may be designed to maintain dorsiflexion prior to contact such that the impact stress can be dispersed through the mobility of the subtalar joint. In barefoot running, the tibialis anterior decreases its activity with fatigue, which may indicate an increased reliance on the plantarflexors to control shock attenuation. This point was alluded to by Lieberman, et al. (2010), such that barefoot runners tend to strike the ground more plantarflexed, whereas shod running is more characteristic of dorsiflexed heelstriking. Global fatigue, as caused by prolonged near-maximal running, affects these coordination strategies differently and ultimately leads to impaired shock attenuation during barefoot running. Previous researchers have speculated that these changes in running mechanics are potential mechanisms for increased injury risk. Specifically, unbalanced activity from the plantarflexor and dorsiflexor muscles may predispose the tibia to increased bending forces which are viewed as unfavorable (Mizrahi, et al., 2000).

It is important to recognize that trained habitually shod runners were used in this study, and that the findings can only be applied to such individuals. In attempt to limit the likelihood of overuse injuries it may be suggested that habitual shod runners intending to begin a barefoot training regimen should do so with caution and limit fatigue within barefoot activities.

Chapter V

Summary, Conclusions, and Recommendations

Summary

Recently, barefoot running has received notable attention from athletes, running syndicates, and acknowledged researchers alike. Predominate theories regard barefoot training as a vehicle to reduce the likelihood, and in some cases eliminate, overuse running injuries. Scientific conclusions encompassing the efficacy and practicality of barefoot running are clouded. Comparisons between shod and barefoot conditions seem to weigh heavily on individual factors such as stride kinematics, footstrike patterns, training status and previous barefoot experience.

The current literature surrounding the use of barefoot running addresses changes in vertical ground reaction force peak and loading rates, plantar pressures, muscle coordination and shock attenuation as compared to shod conditions. However, little attention has been paid to fatigue response between conditions on these measured variables.

The purpose of the present study was to examine how shock attenuation and shank muscle coordination strategies are affected by fatigue while running barefoot and shod. Mizrahi et al. (1997) demonstrated how breakdown of muscle coordination, rather than specific muscle fatigue, has a profound effect on assumed forces while running. The original hypothesis of the current study was that different muscle coordination strategies would be employed while running barefoot and shod. Furthermore, it was hypothesized that global fatigue from a prolonged high-intensity run would have different effects on these strategies based on the presence or absence of footwear. Ultimately, the effect that these changing muscle activation patterns have on the ability to adequately attenuate impact shock was assessed. It was hypothesized that running barefoot would yield higher tibial acceleration rates than running shod, and that following fatigue, these rates would increase to a greater extent while running barefoot.

Ultimately, the question of whether or not a runner can be less prone to injury by supplementing barefoot running has not been completely addressed. Researchers have allowed speculation to this point, but no prospective injury studies involving barefoot running have actually been employed. The current project was designed to provide insight into how running mechanics may respond or adapt to fatigue. This could be valuable information for practitioners to understand regarding the timing and amount that runners can be safely subjected to barefoot training regimens.

Conclusions

The data indicate that runners adopt different shank muscle coordination strategies while running barefoot and shod. In addition, these strategies appear to be affected differently by fatigue in order to adequately attenuate impact shock. Specifically, activation of the tibialis anterior increases with fatigue in a shod condition and decreases with fatigue in a barefoot condition. The gastrocnemius, which initially appeared to have the opposite trend, failed to prove significant perhaps due to the small sample size or the nature of the muscle acting at two joints. The results also show no difference in shock attenuation due to fatigue in either condition; however, barefoot running yielded impaired shock attenuation compared to shod at all times.

Based on the findings by Mizrahi et al. (2000), unbalanced shank muscle coordination may lead to increased bending forces on the tibia while running which could dispose the leg to undue stress. Milner et al. (2006) suggested that impaired shock attenuation may lead to increased injury risk. With these points in mind and paired with the findings of the current study, one may approach barefoot running with caution, especially when an athlete is in a fatigued state.

Recommendations

Future directions for research may include assessing fatigue response of the soleus muscle, as opposed to the gastrocnemius, between barefoot and shod running conditions. Also, it is suggested that prospective injury studies be conducted to unveil whether barefoot training regimens can actually translate to decreased injury risk. In such studies, it is suggested to implement barefoot activities gradually and on compliant surfaces. Caution should be expressed with regards to fatigue with barefoot running.

It may be suggested to remain conservative when implementing barefoot training regimens in habitually shod individuals. Until further information is gathered it may be most appropriate to limit barefoot running to a small percentage of total training volume, and maintain the majority of run training in minimalist shoes such as lightweight trainers or racing flats.

References

Bergmann, G., Kniggendorf, H., Graichen, F., & Rohlmann, A. (1995). Influence of shoes and heel strike on the loading of the hip joint. *J Biomech, 28*(7), 817-827. doi:0021-9290(94)00129-R

Bishop, M., Fiolkowski, P., Conrad, B., Brunt, D., & Horodyski, M. (2006). Athletic footwear, leg stiffness, and running kinematics. *J Athl Train, 41*(4), 387-392.

Butler, R. J., Davis, I. S., & Hamill, J. (2006). Interaction of arch type and footwear on running mechanics. *Am J Sports Med, 34*(12), 1998-2005. doi:10.1177/0363546506290401

Cavanagh, P. R., & Williams, K. R. (1982). The effect of stride length variation on oxygen uptake during distance running. *Med Sci Sports Exerc, 14*(1), 30-35.

Chapman, A. R., Vicenzino, B., Blanch, P., & Hodges, P. W. (2008). Is running less skilled in triathletes than runners matched for running training history? *Med Sci Sports Exerc, 40*(3), 557-565. doi:10.1249/MSS.0b013e31815e727a

De Luca, C. J. (1984). Myoelectrical manifestations of localized muscular fatigue in humans. *Crit Rev Biomed Eng, 11*(4), 251-279.

De Wit, B., De Clercq, D., & Aerts, P. (2000). Biomechanical analysis of the stance phase during barefoot and shod running. *J Biomech, 33*(3), 269-278. doi:S002192909900192X

Divert, C., Mornieux, G., Baur, H., Mayer, F., & Belli, A. (2005). Mechanical comparison of barefoot and shod running. *Int J Sports Med, 26*(7), 593-598. doi:10.1055/s-2004-821327

Dixon, S. J., & McNally, K. (2008). Influence of orthotic devices prescribed using pressure data on lower extremity kinematics and pressures beneath the shoe during running. *Clin Biomech (Bristol, Avon), 23*(5), 593-600. doi:10.1016/j.clinbiomech.2008.01.015

Dugan, S. A., & Bhat, K. P. (2005). Biomechanics and analysis of running gait. *Phys Med Rehabil Clin N Am, 16*(3), 603-621. doi:10.1016/j.pmr.2005.02.007

Elliott, B. C., & Roberts, A. D. (1980). A biomechanical evaluation of the role of fatigue in middle-distance running. *Can J Appl Sport Sci, 5*(4), 203-207.

Flynn, J. M., Holmes, J. D., & Andrews, D. M. (2004). The effect of localized leg muscle fatigue on tibial impact acceleration. *Clin Biomech (Bristol, Avon), 19*(7), 726-732. doi:10.1016/j.clinbiomech.2004.04.015

Hamill, J., & Knutzen, K. (2009). *Biomechanical basis of human movement* (3rd ed.). Baltimore, MD: Lippincott Williams & Wilkins.

Hanon, C., Thepaut-Mathieu, C., & Vandewalle, H. (2005). Determination of muscular fatigue in elite runners. *Eur J Appl Physiol, 94*(1-2), 118-125. doi:10.1007/s00421-004-1276-1

Hirschmuller, A., Baur, H., Muller, S., Helwig, P., Dickhuth, H. H., & Mayer, F. (2009). Clinical effectiveness of customised sport shoe orthoses for overuse injuries in runners- a randomised controlled study. *Br J Sports Med*. doi:10.1136/bjsm.2008.055830

Hreljac, A. (2000). Stride smoothness evaluation of runners and other athletes. *Gait Posture, 11*(3), 199-206. doi:S096663620000045X

Jungers, W. L. (2010). Biomechanics: Barefoot running strikes back. *Nature, 463*(7280), 433-434. doi:10.1038/463433a

Kellis, E., & Liassou, C. (2009). The effect of selective muscle fatigue on sagittal lower limb kinematics and muscle activity during level running. *J Orthop Sports Phys Ther*, *39*(3), 210-220. doi:10.2519/jospt.2009.2859

Kerrigan, D. C., Franz, J. R., Keenan, G. S., Dicharry, J., Della Croce, U., & Wilder, R. P. (2009). The effect of running shoes on lower extremity joint torques. *PM R, 1*(12), 1058-1063. doi:10.1016/j.pmrj.2009.09.011

Knobloch, K., Yoon, U., & Vogt, P. M. (2008). Acute and overuse injuries correlated to hours of training in master running athletes. *Foot Ankle Int, 29*(7), 671-676.

Kurz, M. J., & Stergiou, N. (2004). Does footwear affect ankle coordination strategies? *J Am Podiatr Med Assoc, 94*(1), 53-58.

Lieberman, D. E., Venkadesan, M., Werbel, W. A., Daoud, A. I., D'Andrea, S., Davis, I. S., et al. (2010). Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature, 463*(7280), 531-535. doi:10.1038/nature08723

Ly, Q. H., Alaoui, A., Erlicher, S., & Baly, L. (2010). Towards a footwear design tool: influence of shoe midsole properties and ground stiffness on the impact force during running. *J Biomech, 43*(2), 310-317. doi:10.1016/j.jbiomech.2009.08.029

Maiwald, C., Grau, S., Krauss, I., Mauch, M., Axmann, D., & Horstmann, T. (2008). Reproducibility of plantar pressure distribution data in barefoot running. *J Appl Biomech, 24*(1), 14-23.

McNair, P. J., & Marshall, R. N. (1994). Kinematic and kinetic parameters associated with running in different shoes. *Br J Sports Med, 28*(4), 256-260.

Mercer, J. A., Bates, B. T., Dufek, J. S., & Hreljac, A. (2003). Characteristics of shock attenuation during fatigued running. *J Sports Sci, 21*(11), 911-919.

Mercer, J. A., Devita, P., Derrick, T. R., & Bates, B. T. (2003). Individual effects of stride length and frequency on shock attenuation during running. *Med Sci Sports Exerc, 35*(2), 307-313. doi:10.1249/01.MSS.0000048837.81430.E7

Milner, C. E., Ferber, R., Pollard, C. D., Hamill, J., & Davis, I. S. (2006). Biomechanical factors associated with tibial stress fracture in female runners. *Med Sci Sports Exerc, 38*(2), 323-328. doi:10.1249/01.mss.0000183477.75808.92

Mizrahi, J., Verbitsky, O., & Isakov, E. (2000). Fatigue-related loading imbalance on the shank in running: a possible factor in stress fractures. *Ann Biomed Eng, 28*(4), 463-469.

Mizrahi, J., Voloshin, A., Russek, D., Verbitsky, O., & Isakov, E. (1997). The influence of fatigue on EMG and impact acceleration in running. *Basic Applied Myology, 7*(2), 111-118.

Morio, C., Lake, M. J., Gueguen, N., Rao, G., & Baly, L. (2009). The influence of footwear on foot motion during walking and running. *J Biomech, 42*(13), 2081-2088. doi:10.1016/j.jbiomech.2009.06.015

Nigg, B. M., Stefanyshyn, D., Cole, G., Stergiou, P., & Miller, J. (2003). The effect of material characteristics of shoe soles on muscle activation and energy aspects during running. *J Biomech, 36*(4), 569-575. doi:S0021929002004281

O'Connor, K. M., & Hamill, J. (2005). Frontal plane moments do not accurately reflect ankle dynamics during running. *J Appl Biomech, 21*(1), 85-95.

Pohl, M. B., Messenger, N., & Buckley, J. G. (2007). Forefoot, rearfoot and shank coupling: effect of variations in speed and mode of gait. *Gait Posture, 25*(2), 295-302. doi:10.1016/j.gaitpost.2006.04.012

Pohl, M. B., Rabbito, M., & Ferber, R. (2010). The role of tibialis posterior fatigue on foot kinematics during walking. *J Foot Ankle Res, 3*, 6. doi:10.1186/1757-1146-3-6

Queen, R. M., Abbey, A. N., Wiegerinck, J. I., Yoder, J. C., & Nunley, J. A. (2010). Effect of shoe type on plantar pressure: a gender comparison. *Gait Posture, 31*(1), 18-22. doi:10.1016/j.gaitpost.2009.08.235

Reinking, M. F., Austin, T. M., & Hayes, A. M. (2007). Exercise-related leg pain in collegiate cross-country athletes: extrinsic and intrinsic risk factors. *J Orthop Sports Phys Ther, 37*(11), 670-678. doi:10.2519/jospt.2007.2534

Robbins, S. E., & Gouw, G. J. (1991). Athletic footwear: unsafe due to perceptual illusions. *Med Sci Sports Exerc, 23*(2), 217-224.

Robbins, S. E., & Hanna, A. M. (1987). Running-related injury prevention through barefoot adaptations. *Med Sci Sports Exerc, 19*(2), 148-156.

Rome, K., Hancock, D., & Poratt, D. (2008). Barefoot running and walking: the pros and cons based on current evidence. *N Z Med J, 121*(1272), 109-111.

Shim, J., Acevedo, E. O., Kraemer, R. R., Haltom, R. W., & Tryniecki, J. L. (2003). Kinematic changes at intensities proximal to onset of lactate accumulation. *J Sports Med Phys Fitness, 43*(3), 274-278.

Smoliga, J. M., Myers, J. B., Redfern, M. S., & Lephart, S. M. (2010). Reliability and precision of EMG in leg, torso, and arm muscles during running. *J Electromyogr Kinesiol, 20*(1), e1-9. doi:10.1016/j.jelekin.2009.09.002

Squadrone, R., & Gallozzi, C. (2009). Biomechanical and physiological comparison of barefoot and two shod conditions in experienced barefoot runners. *J Sports Med Phys Fitness, 49*(1), 6-13.

Stacoff, A., Nigg, B. M., Reinschmidt, C., van den Bogert, A. J., & Lundberg, A. (2000). Tibiocalcaneal kinematics of barefoot versus shod running. *J Biomech, 33*(11), 1387-1395. doi:S0021-9290(00)00116-0

Stergiou, N., Bates, B. T., & Kurz, M. J. (2003). Subtalar and knee joint interaction during running at various stride lengths. *J Sports Med Phys Fitness, 43*(3), 319-326.

Tessutti, V., Trombini-Souza, F., Ribeiro, A. P., Nunes, A. L., & Sacco Ide, C. (2010). In-shoe plantar pressure distribution during running on natural grass and asphalt in recreational runners. *J Sci Med Sport, 13*(1), 151-155. doi:10.1016/j.jsams.2008.07.008

van Mechelen, W. (1992). Running injuries. A review of the epidemiological literature. *Sports Med, 14*(5), 320-335.

Vormittag, K., Calonje, R., & Briner, W. W. (2009). Foot and ankle injuries in the barefoot sports. *Curr Sports Med Rep, 8*(5), 262-266. doi:10.1249/JSR.0b013e3181b9e3be

Warburton, M. (2001). Barefoot running. *Sportscience, 5*(3), 1-5.

Wen, D. Y. (2007). Risk factors for overuse injuries in runners. *Curr Sports Med Rep, 6*(5), 307-313.

Wiegerinck, J. I., Boyd, J., Yoder, J. C., Abbey, A. N., Nunley, J. A., & Queen, R. M. (2009). Differences in plantar loading between training shoes and racing flats at a self-selected running speed. *Gait Posture, 29*(3), 514-519. doi:10.1016/j.gaitpost.2008.12.001

Williams, K. R., & Cavanagh, P. R. (1987). Relationship between distance running mechanics, running economy, and performance. *J Appl Physiol, 63*(3), 1236-1245.

Wilson, J. F., & Rochelle, R. D. (2009). Footfall dynamics for racewalkers and runners barefoot on compliant surfaces. *J Biomech, 42*(15), 2472-2478. doi:10.1016/j.jbiomech.2009.07.020

Appendix A.

Informed Consent

Volunteer copy / Investigator copy

WESTERN WASHINGTON UNIVERSITY

An equal opportunity university

Department of Physical Education, Health & Recreation

516 High Street, Bellingham, Washington 98225-9067
(360)650-3105 Fax (360) 650-7447

INFORMED CONSENT FORM
WESTERN WASHINGTON UNIVERSITY

Title of Investigation: Effects of fatigue on muscle activation and shock attenuation during barefoot running.

Investigator: Carl Newton
Department of Physical Education, Health and Recreation
516 High St.
Western Washington University
Bellingham, WA 98225-9067
Phone: (360) 650 – 3449 or (360) 305-7875
newtonc5@students.wwu.edu

Co-Advisors:

Dave Suprak	Kathy Knutzen
Phone: (360) 650- 2586	Phone: (360) 650- 3055
David.Suprak@wwu.edu	Kathy.Knutzen@wwu.edu

This is to certify that I, ____, hereby agree to participate as a volunteer in a scientific investigation as an authorized part of the education and research program of Western Washington University under the supervision of graduate student Carl Newton.

Purpose of the Study:

This study in which I will be participating is designed to investigate the effect of global running fatigue on barefoot and shod running in habitually shod runners.

Procedures to be followed:
I understand that males and females between the ages of 18 to 45 will be invited to participate in this study. I understand that in order to participate in this study:

- I must be over the age of 18 years.
- I must be free of injury to the muscles, bones, or joints of the upper and lower extremity.
- I must have full range of motion of my trunk, shoulder, elbow, wrists, hips, knees, and ankles.
- I must be running on average at least 10 miles per week.
- I must have little to no experience running barefoot and train regularly in one pair of running shoes.
- I must be willing to attend 2 separate testing sessions, each containing a 30-minute fatiguing run.

I understand that the study will require attendance at two sessions (totaling ~2 hours), and that I will be required to complete forms before participating in the first session. The two sessions will be selected in random order as determined by a coin flip and activities will be as follows:

<u>Prior to Testing:</u>

- I will receive a detailed explanation of the study procedures.
- I will read and sign the informed consent form.
- I will read and complete the medical background form and the hold harmless agreement.
- My age, height and weight measurements will be recorded.

<u>Test Session A: (barefoot trials)</u>

- I will have portions of my right lower leg shaved and cleansed with alcohol wipes. Five electrodes will be placed on my leg and connected to a small data logger computer which I will wear around my waist.
- A small light-weight accelerometer will also be placed on my leg and secured with prewrap and coban elastic tape. I understand that the coban will be fit snug to reduce movement of the accelerometer, but adjusted to my individual tolerance and comfort.
- I will perform a 5 minute warm up jogging lightly around a running track in my typical training shoes.
- I will remove my shoes and socks, and run barefoot for 100 meters ($^1/_4$ lap) at a comfortable self-selected pace while data is collected.

- The equipment will be removed, I will replace my shoes and proceed to run a maximal effort 30-minute run around the track. I will be allowed to monitor my running time and pace myself as evenly as possible. The goal will to be to achieve as many laps as possible in 30 minutes.
- Immediately upon completion of the 30-minute run, the equipment will be replaced, I will remove my shoes and socks, and run barefoot for 100 meters ($^1/_4$ lap) at the same comfortable pace which data was collected prior to the 30-minute run. This pace will be monitored by a stopwatch, and I will attempt to run as evenly as possible.
- The sites of the electrodes will be marked with a pen and I will be encouraged not to completely wash away the markings until after the second session, such that electrodes can be replaced in the same precise location. I understand that these markings will easily wash away after the second session.
- Once I have completed the testing session, I understand that all the equipment will be removed and I will be encouraged to cool down and stretch to individual satisfaction.

<u>Test Session B: (shod trials)</u>

- I will have portions of my right lower leg shaved and cleansed with alcohol wipes. Five electrodes will be placed on my leg and connected to a small data logger computer which I will wear around my waist.
- A small light-weight accelerometer will also be placed on my leg and secured with prewrap and coban elastic tape. I understand that the coban will be fit snug to reduce movement of the accelerometer, but adjusted to my individual tolerance and comfort.
- I will perform a 5 minute warm up jogging lightly around a running track in my typical training shoes.
- Without removing my shoes, I will run shod for 100 meters ($^1/_4$ lap) at a comfortable self-selected pace while data is collected.
- The equipment will be removed and I will proceed to run a maximal effort 30-minute run around the track. I will be allowed to monitor my running time and pace myself as evenly as possible. The goal will to be to achieve as many laps as possible in 30 minutes.
- Immediately upon completion of the 30-minute run, the equipment will be replaced, I will run shod for 100 meters ($^1/_4$ lap) at the same comfortable pace which data was collected prior to the 30-minute run. This pace will be monitored by a stopwatch, and I will attempt to run as evenly as possible.

- Once I have completed the testing session, I understand that all the equipment will be removed and I will be encouraged to cool down and stretch to individual satisfaction.

Discomforts and Risks:

I understand that the electrodes placed on my skin are part of a passive system that record activity from my muscles. They do not send any signal to myself, and I will not receive any electrical stimulus, shocks or discomforts.
I understand that, as with any exercise program, there are risks of injury due to accidents during the exercise activities. Additionally, I realize that there may be minimal risk, such as discomfort or pain as a result of injury to involved musculature, joints or connective tissue. These are risks associated with any physical activity. Although some running segments will be performed at a high intensity, I understand that I will dictate my own pace and it should not be any greater than I would experience in a race situation. If I experience pain, I am aware that I may withdraw from participation in this study at any time, without penalty.
If I feel I cannot or should not perform any of the barefoot or shod running segments, I should not participate in this study.

Benefits to Me:

I understand that there are no direct benefits to me as a result of participating in this study, however, the results may help me analyze my ability to attenuate impact shock or interpret my muscle coordinative strategies while running fatigued.

Potential Benefits to Society:

By participating in this study I will be contributing to research that aims to advance our understanding of how fatigue effects alterations in barefoot running mechanics. If alterations in running mechanics related to fatigue are found to be more detrimental while running barefoot, it may help to identify the practicality and efficacy of barefoot running in terms of injury prevention. The results of this study may help to set the stage for future work in prescribing longer term barefoot running studies.

Statement of Confidentiality:

I understand that any data or my answers to questions will remain confidential with regard to my identity. Only the investigator and his assistants will have access to my identity and to information that can be associated with my identity.

In the event of publication of this research, no personally identifying information will be disclosed.

The investigation and my part in the investigation have been defined and fully explained to me by Carl Newton or his assistant and I understand his/her explanation. The procedures of this investigation and a description of any risks and discomfort have been discussed in detail with me and I understand that a copy of the signed consent form will be provided to me.

Right to Ask Questions:

I have been given an opportunity to ask whatever questions I may have had and all such questions have been answered to my satisfaction.

I understand that I am free to deny any answers to specific items or questions in interviews or questionnaires. If I have any questions about this study, I can contact Carl Newton at the contact information listed on the front page of this consent form.

I understand that for additional information about my rights as a research participant, I may contact the WWU HSRC Administrator, at:

Janai Symons
HRSC Administrator
Research and Sponsored Programs
Old Main Building 530
Western Washington University
Bellingham, WA 98225-9038
(360) 650-3082
janai.symons@wwu.edu

Event of injury:

I understand that emergency medical care will be summoned in the event of injury resulting from this study. In the event of adverse effects related to this study, I understand that I shall contact the office listed above. I also understand that I am not waiving any rights that I may have against WWU for injury resulting from negligence of the University or investigators.

Voluntary Participation:

I understand that my participation in this study is voluntary, and that I may withdraw from this study at any time by notifying the investigator. I also understand that my participation may be terminated by the investigator if I do not fit any of the pre-determined subject categories or if he or she feels that my personal well-being is in question.

This is to certify that I am over the age of 18 years, and I consent to and give permission for my participation as a volunteer in this program of investigation. I understand that I will receive a signed copy of this consent form. I have read this form, and understand the content of this consent form.

Volunteer Date

I, the undersigned, have fully explained the investigation to the above subject.

Investigator Date

Audiotaping, Videotaping, and Photography

By initialing on the lines below, I am indicating that I give the research team permission to (please initial all that apply):

______Photograph, audiotape and/or videotape my participation in this study.

______Use photographs, audiotape or videotape recordings of me when they present this research in educational and professional venues, *even if I am personally identifiable*.

______Use photographs, audiotape or videotape recordings of me when they present this research in educational and professional venues, *only as long as I am not personally identifiable*.

Appendix B.

Health History Questionnaire

Department of Physical Education, Healthy & Recreation
Western Washington University

Health History

Name:______________________________ Phone:__________________

Address:________________________ City:____________ Zip:_________

Date of birth:____________ Age:______ Height:______ Weight:______

1. Do you currently have, or have had in the previous 3 months, any injuries or medical conditions that have prevented you from participating in typical training practices for more than one week?

    i. If yes, please list.

2. Are you currently receiving or have received any medical treatment, physical therapy or alternative treatment for any condition discussed above? **Yes or No** (please circle one)

    i. If yes, please explain.

3. Is there any other condition not mentioned here that might affect your ability to exercise, or be aggravated by exercise? **Yes or No** (please circle one)

    i. If yes, please describe.

4. Are you an experienced runner? How many years of experience do you have?

5. Are you currently running? How often?

6. Do you have any experience running barefoot? **Yes or No** (please circle one)

    i. If yes, please explain.

Appendix C.

Data Collection Form

Name:_____________________________ Subject Number:__________

Shod Condition *(Heads)*

Date of trial:____________ Time:______ Height:______ Weight:______

Pre-fatigue 100m time:_______________

Accelerometer file name:___________________

EMG file name:___________________________

30-minute run distance:______________

Post-fatigue 100m time:______________

Accelerometer file name:___________________

EMG file name:___________________________

Barefoot Condition *(Tails)*

Date of trial:____________ Time:______ Height:______ Weight:______

Pre-fatigue 100m time:_______________

Accelerometer file name:___________________

EMG file name:___________________________

30-minute run distance:______________

Post-fatigue 100m time:______________

Accelerometer file name:___________________

EMG file name:___________________________

Appendix D.

Example Data Set

	time (s)	
SHD pre	20.84	1.61%
SHD pst	20.82	1.70%
BFT pre	22.12	4.44%
BFT pst	20.94	1.13%
mean	21.18	1.059

SUBJECT 10

conversion to Gs

(N-1822)/177

Height	1.74
Weight	71.7
Age	26
Shod distance	7.815
Barefoot distance	7.620

shod pre					
acc tz	2.916				
emg tz	1.388				
	time	peak	rate	TA iEMG	MG iEMG
start	11.887				
50% pt	22.307				
pk 1	22.377	7.791	313.84	1.121	0.153
pk 2	23.018	8.034	311.82	1.560	0.079
pk 3	23.654	8.475	362.17	1.355	0.189
pk 4	24.293	7.802	302.86	1.797	0.173
pk 5	24.934	7.215	253.51	1.545	0.117
pk 6	25.570	8.006	346.26	1.398	0.180
pk 7	26.209	7.559	280.30	1.217	0.176
pk 8	26.855	8.288	305.57	0.816	0.107
	mean	7.896	309.54	1.351	0.147

shod post					
acc tz	3.859				
emg tz	1.449				
	time	peak	rate	TA iEMG	MG iEMG
start	10.813				
50% pt	21.223				
pk 1	21.322	6.960	420.39	1.657	0.111
pk 2	21.965	7.062	253.68	1.559	0.066
pk 3	22.607	7.424	279.14	2.300	0.085
pk 4	23.254	7.514	295.70	1.655	0.070
pk 5	23.895	7.102	248.51	1.406	0.070
pk 6	24.543	6.910	250.86	1.209	0.098
pk 7	25.195	6.638	250.51	1.077	0.081
pk 8	25.848	5.972	217.48	1.188	0.064
	mean	6.948	277.03	1.506	0.081

bare pre					
acc tz	7.064				
emg tz	1.415				
	time	peak	rate	TA iEMG	MG iEMG
start	11.119				
50% pt	22.179				
pk 1	22.248	11.763	975.37	0.727	1.144
pk 2	22.850	12.842	2343.20	0.826	0.814
pk 3	23.465	12.056	1190.97	1.049	0.708
pk 4	24.072	11.593	1196.04	0.771	0.713
pk 5	24.666	12.192	1315.05	0.807	0.659
pk 6	25.283	12.842	1439.91	0.509	0.733
pk 7	25.900	10.684	963.19	0.616	0.522
pk 8	26.520	12.102	1430.03	0.849	0.581
	mean	12.009	1356.72	0.769	0.734

bare post					
acc tz	0.906				
emg tz	1.370				
	time	peak	rate	TA iEMG	MG iEMG
start	10.588				
50% pt	21.058				
pk 1	21.373	12.017	1218.64	0.641	0.755
pk 2	21.953	9.243	1111.82	0.842	0.578
pk 3	22.527	5.401	358.10	0.773	0.940
pk 4	23.098	12.842	1625.78	0.729	0.579
pk 5	23.678	8.684	765.16	0.656	0.687
pk 6	24.250	12.492	1586.24	0.832	0.560
pk 7	24.822	12.842	1560.90	0.436	0.434
pk 8	25.398	9.757	1293.10	0.735	0.508
	mean	10.410	1189.97	0.705	0.630

Appendix E.

All Subjects' Data

Rate of Acceleration				
Subject	**ShodPre**	**ShodPost**	**BarePre**	**BarePost**
1	326.44	338.04	1775.20	1995.98
2	295.39	302.25	971.05	1059.01
3	454.09	287.01	1277.71	689.92
4	691.94	481.75	1330.20	668.37
5	655.70	1802.90	2128.68	2919.96
6	2000.40	1385.82	2316.75	2252.41
7	416.67	347.36	802.14	832.36
8	698.91	989.34	2499.56	2170.43
10	309.54	277.03	1356.72	1189.97
11	363.21	366.19	2058.98	2279.56
12	725.43	805.01	1581.17	2069.87
mean	**630.70**	**671.15**	**1645.29**	**1647.98**
s.d.	484.85	519.08	555.84	778.05

Tibialis Anterior iEMG						
Subject	**ShodPre**	**ShodPost**	**BarePre**	**BarePost**	**Shod % Change**	**Bare % Change**
1	0.725	0.824	0.725	0.827	13.53	14.16
2	0.618	0.540	0.476	0.391	-12.58	-17.75
3	0.277	0.257	0.089	0.075	-7.08	-15.89
4	0.650	0.592	0.384	0.415	-8.93	8.15
5	0.469	0.501	0.506	0.351	6.83	-30.74
6	0.459	0.369	0.372	0.232	-19.55	-37.56
7	0.503	0.594	0.520	0.248	18.16	-52.28
8	0.512	0.915	0.368	0.488	78.70	32.46
10	1.351	1.506	0.769	0.705	11.50	-8.29
11	0.164	0.159	0.146	0.113	-3.43	-22.70
12	0.565	0.667	0.560	0.518	18.06	-7.49
mean	**0.572**	**0.630**	**0.447**	**0.397**	**8.66**	**-12.54**
s.d.	0.304	0.367	0.209	0.232	26.59	24.26

Medial Gastrocnemius iEMG						
Subject	**ShodPre**	**ShodPost**	**BarePre**	**BarePost**	**Shod % change**	**Bare % Change**
1	0.195	0.266	0.315	0.442	36.64%	40.36%
2	0.310	0.214	0.194	0.099	-30.87%	-49.20%
3	0.036	0.066	0.536	0.384	83.57%	-28.38%
4	0.413	0.337	0.388	0.532	-18.40%	37.26%
5	0.337	0.370	0.537	0.583	9.74%	8.51%
6	0.351	0.118	0.176	0.039	-66.48%	-77.96%
7	0.394	0.330	0.415	0.489	-16.18%	17.91%
8	0.990	0.634	0.224	0.987	-36.00%	341.45%
10	0.147	0.081	0.734	0.630	-44.96%	-14.19%
11	0.053	0.066	0.545	0.398	24.62%	-27.08%
12	0.108	0.171	0.441	0.374	57.54%	-15.28%
mean	**0.303**	**0.241**	**0.410**	**0.451**	**-0.07%**	**21.22%**
s.d.	0.265	0.172	0.174	0.255	46.51%	112.04%