2012

The Effect of Extended Durations of Walking in Different Footwear on Measures of Muscle Activity

Kacie Brooke Childers

THE EFFECT OF EXTENDED DURATIONS OF WALKING IN DIFFERENT FOOTWEAR ON MEASURES OF MUSCLE ACTIVITY

by
Kacie Childers

A thesis submitted to the faculty of The University of Mississippi in partial fulfillment of the requirements of the Sally McDonnell Barksdale Honors College.

Oxford
May 2012

Approved by

Advisor: Dr. John Garner

Reader: Dr. Christopher Black

Reader: Dr. Teresa Carithers

ACKNOWLEDGEMENTS

This thesis would not have been possible without the help of Dr. John Garner and Harish Chander. Thank you for your advice, encouragement, and dedication to helping me succeed.

I extend my thanks to Dr. Christopher Black and Dr. Teresa Carithers for serving as my second and third readers.

I owe gratitude to the Sally McDonnell Barksdale Honors College, the department of Exercise Science, and the University of Mississippi for providing me with a rich and challenging undergraduate educational experience that will benefit me for a lifetime.

ABSTRACT

KACIE CHILDERS: The Effect of Extended Durations of Walking in Different Footwear on Measures of Muscle Fatigue (Under the direction of Dr. John Garner)

In 2010, nearly 3.1 million nonfatal workplace injuries and illnesses were reported among private industry employers resulting in an incidence rate of 3.5 cases per 100 full time workers, according to the Bureau of Labor Statistics. Of the fatal occupational injuries in 2010, 14% were attributed to falls. Forty-five percent of falls have been attributed to inappropriate footwear, and occupationally induced muscle fatigue has been identified as a contributing factor to falls. Different footwear types have been shown to effect muscle fatigue levels during walking and standing. In an occupational setting, workers generally endure relatively small workloads for extended durations. The purpose of this study was to examine the amount of muscle activation of the ankle musculature over an extended duration of walking and standing in two types of commonly worn occupational footwear. Fourteen healthy male adults with no history of orthopedic, musculoskeletal, cardiovascular, neurological or vestibular abnormalities completed the study. The experimental session included an extended duration of walking (4 hours) with muscle activation of the ankle musculature during a maximal voluntary contraction measured at 30 minute intervals. The footwear included work boots that met ANSI standards and low top shoes. A minimum of 72 hours of rest between conditions was provided. Repeated measures analyses of variance determined there were no significant differences between the time points for footwear or the time-shoe interaction. Significance was found with shoe type with the low top shoe eliciting greater activation for both muscles. Shaft height of the shoe may have been the significant influencing factor affecting muscle activity of the lower extremities. The work boots with the higher

shaft showed more potential for resisting fatigue over extended durations. The results of this study may be beneficial in the aid of efficient design of future footwear to be worn in occupational settings.

TABLE OF CONTENTS

CHAPTER 1

INTRODUCTION

According to the Bureau of Labor Statistics, nearly 3.1 million nonfatal workplace injuries and illnesses were reported among private industry employers in 2010, resulting in an incidence rate of 3.5 cases per 100 full-time workers (BLS 2011). Of the fatal occupational injuries reported in 2010, 14% were attributed to falls and 17% occurred in the construction sector (BLS 2011). Most people between ages 22 and 65 spend approximately 40% of their waking hours at the workplace, and there is some risk of injury or illness attending virtually every job held by every working American (Leigh, Markowitz and Fahs 1997). A high prevalence of occupational injury has been noted in the mining industry. Between 2003 and 2007, 14.4% of the injuries and accidents of underground miners and 22.7% of the surface mine accidents that were recorded in the Mine Safety and Health Administration database were related to slips or falls of the worker (Merryweather, Yoo and Bloswick 2011).

Slips and falls are one of the major causes of serious injury and death in the United States (Fingerhut, Cox and Warner 1998), not only for miners but for several occupations such as railroad workers, construction workers, and many others. Decrements in balance control have been considered an important factor in the probability of a fall, and often falls are a proximal cause of injury and disability in contemporary industrial settings (Lin, et al. 2008). Occupationally induced localized muscle fatigue has been identified as an intrinsic factor contributing to slips and falls. Muscle fatigue has been defined as a decrease in muscle activity over time (Arend-Nielsen and Mills 1988, Lepers, Maffiuletti, et al., Neuromuscular fatigue during a long-duration cycling exercise 2002). One third of the workforce in the U.S. holds a job demanding a considerable amount of strength. Strength requires muscles to produce force, and over time fatigue may occur at work (Swaen, et al. 2003). An increased postural sway has been observed after the onset of fatigue, which may be associated with a higher risk of falling (Parijat and Lockart 2008).

Footwear may be a direct cause of a fall. In fact, the force interactions between the shoe and floor are probably the most critical biomechanical parameters in slips and falls (Redfern, et al. 2001). Around 45% of falls reported in older adults have been attributed to improper footwear. One study of people who had suffered a fall-related hip fracture reported that at the time of the injury 75% were wearing improper footwear (J. Menant, et al. 2008).

In the workplace, maintenance of upright posture and balance is essential to the workers' safe and efficient conductance of required tasks. To maintain upright balance in quiet standing, a person must keep their center of gravity (COG) inside the base of support (BOS) created by their foot placement (Kincl, et al. 2002). During gait, however, the goal is to move the body outside of the BOS while preventing a fall. The position of a person's COG is constantly changing as sensory input from the visual, somatosensory, and vestibular systems alters motor function to accommodate the effect of gravity. The postural muscles must continually compensate to keep the COG within the BOS, which potentially produces postural sway.

Although a relatively small amount of muscle activation is required to maintain erect standing posture in bipeds, the motor control system is essential to postural control (Corbeil, et al. 2002, Gefen, et al. 2002). When muscles essential to postural control become fatigued, the risk of falls and/or injury increases and the ability of those muscles to maintain posture and balance is jeopardized. The onset of muscle fatigue therefore requires more effort from the postural muscles in order to maintain erect posture (Corbeil, et al. 2002). Recent studies examining the detrimental effects of muscle fatigue on balance mechanisms have investigated the effects of a relatively high level of fatigue (greater than 50% maximal voluntary contraction (MVC)) and greater than 33% of maximum aerobic capacity (Yaggie and McGregor 2002, Vuillerme, et al. 2002). However, workloads experienced in occupational settings are usually lower than those induced by exercise. Also, the worker is generally exposed to the workload over an extended duration of time as opposed to a shorter bout of exercise. Fatigue in occupational settings has been shown to occur over an extended duration of time at a low rate of less than 15% MVC and less than 33% of maximal aerobic capacity (Davidson, Madigan and Nussbaum 2004).

Fatigue of the ankle musculature has been shown to increase the frequency of body sway (Corbeil, et al. 2002, Lepers, Bigard, et al. 1997, Nardone, Tarantola and Giordano, et al. 1997). A study by Corbeil et al. looked at the effects of inducing muscle fatigue of the ankle plantarflexors on upright standing and postural stability. The results suggested that fatigue placed higher demands on the postural control system by increasing the frequency of actions needed to regulate an upright stance as compared to a condition without fatigue (Corbeil, et al. 2002).

Some characteristics of shoes may have an effect on levels of muscle fatigue. A stiff boot shaft has been shown to block the ankle joint motion and cause fatigue at more proximal joints such as the knee and the hip (Bohm and Hosl 2010). A shoe with a heavier mass may also elicit greater fatigue levels at the knee and the hip as compared with a lighter shoe (H. Chander, et al. 2010).

Previous studies have shown that muscle fatigue effects balance over time (Behm and St-Pierre 1997; Lepers, Maffiuletti, et al. 2002; Pline, Madigan and Nussbaum 2006). Research has also shown that footwear has an effect on muscle fatigue levels (Bohm and Hosl 2010, H. Chander, et al. 2010). With this research showing that footwear has an effect on muscle fatigue levels and muscle fatigue effects risk of falling over time, the effects of different types of footwear on muscle activity over extended durations have not been analyzed. The purpose of the current study is to examine the effect of footwear on muscle activation while walking for extended durations.

Purpose

The purpose of this study was to examine the amount of muscle activation of the ankle musculature over an extended duration of walking and standing in two different types of commonly worn occupational footwear.

Hypotheses

H_{0_1}: There will be no difference in muscle activation in the ankle over time, irrespective of footwear type.

H_{A_1}: The muscle activity of the ankle musculature will decrease over time.

In a study conducted by Behm and St-Pierre comparing long and short duration muscle fatigue which was induced with an isometric submaximal fatigue protocol, the long duration fatigue group showed a significantly greater decrease in muscle activation than the short duration fatigue group (Behm and St-Pierre 1997). A study by Lepers et al. examining fatigue during a long duration cycling exercise found a significant reduction in muscle activation towards the end of the five hour cycling period (Lepers, Maffiuletti, et al. 2002). Pline, et al. observed the effects of the time over which fatigue was induced and the magnitude of fatigue on increases in postural sway and found that as fatigue was induced over longer durations there were larger increases in sway velocity and sway area (Pline, Madigan and Nussbaum 2006). Postural sway increases with muscle fatigue at the ankle (Yaggie and McGregor 2002). Based on these conclusions from previous studies, a decreased activation of the ankle musculature was expected.

H_{0_2}: There will be no difference in the measure of muscle activation in the work boot (WB) and low top flat sole slip resistant shoe (LT) conditions.

H_{A_2}: There will be a significant difference in muscle activation between the WB and LT conditions.

The work boot was characterized by a high boot shaft, greater mass, and higher heel height whereas the LT was characterized by a lower shaft, smaller mass, and lower heel height. A high boot shaft, which functions to provide support and stability at the ankle joint, may compromise the ankle range of motion and therefore the ability of the ankle to generate enough power for forward propulsion (Bohm and Hosl 2010; Hijmans et al. 2007). With a lower boot shaft, the LT was expected to allow for more activation of the ankle musculature. Therefore, over time, it is expected that the WB has the greater potential to resist fatigue for longer durations. It has been shown that fatigue increases the frequency of actions necessary to maintain an upright stance as compared to a condition without fatigue (Corbeil, et al. 2002).

Based on this previous research, a significant difference in muscle activation between the WB and LT conditions was expected. Specifically, higher muscle activation was expected to be seen with the LT as compared with the WB.

CHAPTER II

REVIEW OF LITERATURE

Muscle Fatigue

The purpose of this experiment was to examine the effects of walking in different types of footwear on a hard surface for an extended duration of time on muscle fatigue. Muscle fatigue may be defined as the decrease in the force-generating capacity or power of a muscle or muscle group that occurs regardless of the task performed and results in a reduced ability to perform work (Arend-Nielsen and Mills 1988; Lepers, Maffiuletti, et al. 2002).

In this experiment, muscle fatigue of the lower extremities resulting from walking and standing for extended durations in different types of footwear was measured by electromyographical activity levels. Electromyography (EMG) is commonly used to evaluate muscle activation. Muscle fatigue is characterized by a decrease in the force output of a muscle as well as reduced speed in the muscle contractile properties (Bigland-Ritchie, Johansson, et al. 1983). Fatigue of sustained maximal voluntary contractions is often coupled with a decline in surface EMG. This suggests that loss of force in voluntary contractions may result from inadequate muscle activation as well as failure of the muscle's contractile mechanism (Bigland-Ritchie, Johansson, et al. 1983).

The use of three paramaters to quantify changes in EMG during fatigue has traditionally been employed. These include muscle fiber conduction velocity, mean power frequency of the EMG spectrum, and summated, integrated or some other measure of overall EMG amplitude (Arend-Nielsen and Mills 1988). The myoelectric signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle (De Luca 1979). During a sustained, isometric contraction with the presence of localized muscle fatigue the signal shows a shift in the frequency content of the signal toward the low end, an increase in the amplitude, and decreased force output (M. Redfern 1992). As the frequency content of the signal shifts to lower frequencies, more energy is transferred through the tissues to the electrodes, resulting in a greater amplitude of the recorded signal (M. Redfern 1992).

Foot stability in stance can be measured from abnormal deviations in the center of pressure (COP) trajectory (Gefen, et al. 2002, Han, Paik and Im 1999)and localized muscle fatigue may be quantified by use of EMG as the spectral density shifts towards lower frequencies (Gefen, et al. 2002). For an independent motor unit discharge, the power spectral density of the surface EMG signal is the summation of the power spectral densities of the motor unit action potentials generated by the active motor units multiplied by the spectra of the point processes that describe the motor unit firing patterns. At higher frequencies, the EMG spectrum is the summation of the spectra of the surface potentials of the active motor units. The spectrum depends on the muscle unit action potential shape, the conduction velocity with which the muscle unit action potential propagates, and the distance of the motor unit from the recording electrodes. A greater distance of the motor unit from the recording electrodes results in a lower power at high frequencies (Farina, Fosci and Merletti 2001).

Many physiological explanations have been proposed to account for the spectral shifts and amplitude changes of the EMG during fatigue, including motor unit recruitment (Edwards and Lippold 1956), motor unit synchronization (Bigland-Ritchie, Donovan and Roussos 1981), firing rate and interpulse intervals (M. Redfern 1992), and motor unit and action potential shape (Mills 1982). Motor unit recruitment is believed to occur as a muscle becomes fatigued as a response to reduced muscle contractility. The predominant causes of spectral shifts in EMG are thought to result from factors such as firing rate, variation in interpulse intervals, and changes in the shapes of the motor unit action potentials (M. Redfern 1992).

Such changes in EMG recordings of muscle activity over extended periods show a decrease in muscle activation from the muscle as more time passes, which indicates muscle fatigue. Since muscle fatigue is associated with disturbance of balance and increased postural sway, muscle fatigue is directly related to an increased risk of falling and injury. Therefore, it is important for industrial workers to minimize fatigue in the workplace in order to minimize the risk of falls and injuries on the job.

The cause of muscle fatigue is not completely clear, and there is research to support fatigue originating from a combination of both central and peripheral factors (Zwarts, Bleijenberg and van Engelen 2008).That is to say that much research on muscle fatigue focuses on determining if fatigue originates from the exercising muscle (Merton 1954; Bigland-Ritchie, Furbush and Woods 1986)or the nervous system (Martin, et al.

2010; Gribble and Hertel 2004). Central fatigue occurs when neural activation of a muscle insufficiently stimulates muscle units to contract to their fullest potential. Peripheral fatigue refers to a phenomenon in which the nervous system supplies adequate stimulation to the muscle units, yet the muscle units are unable to generate a maximal response despite the maximal neural activation. Therefore, the terms "central" and "peripheral" in reference to sites of fatigue do not necessarily refer to the anatomical meaning of the words.

Several studies conclude that peripheral control is the primary factor contributing to muscle fatigue (Merton 1954; Bigland-Ritchie, Furbush and Woods 1986). One such study by P.A. Merton supported this claim by finding that no significant difference in tension development was seen when comparing a voluntary maximal contraction with an electrically induced maximal contraction, and when fatigue occurred by voluntary contraction electrical stimulation of the motor nerve was unable to restore tension. Therefore, voluntary effort must be continually exerted after the onset of muscle fatigue to activate the muscle fully and the decrease in tension must be due to the peripheral factors (Merton 1954). When a voluntary contraction of a muscle is implemented, the muscle exerts maximal force for only a few seconds before fatigue sets in. After a few moments, the effort exerting the voluntary contraction does not decrease but the force generated begins to slope. This supports that idea that muscle fatigue is due in some part to the periphery. Peripheral nervous system level muscle fatigue is contributed to a muscle's failure to react to a neural signal or the failure of a muscle to react to neural excitation. Once a muscle is fatigued, electrical stimulation cannot fully restore the contraction potential.

Central fatigue refers to an exercise-induced reduction in maximal voluntary contraction force, which is not accompanied by the same reduction in maximal evocable force (Vøllestad 1997). An article by Asmussen and Mazin titled "A Central Nervous Component in Local Muscular Fatigue" explains that any activity used as a diversion between bouts of physical activity causes a decrease in the effects of muscle fatigue as compared with bouts of activity with no diverting activities performed in between. The CNS as the originating site of fatigue was also supported by an experiment in which muscle fatigue was induced with the eyes closed and tension in the muscles was restored simply by opening the eyes. Fatigue at the central level is seen to induce a failure of excitation of the motor neurons due to a reduction of motor drive from the central nervous system (Bigland-Ritchie, Johansson, et al. 1983; Asmussen and Mazin 1978). These changes in neuromuscular control may be due to the slow conduction of afferent signals from the fatigue altered state of the muscle will lead to slowed propagation of efferent signals, thereby hindering the ability to effectively carry out compensatory movements (Gribble and Hertel 2004).

One study conducted by Jane A. Kent-Braun concluded that central factors contribute approximately 20 percent of muscle fatigue developed, while peripheral factors contribute the other 80 percent. These approximations fluctuate depending on which muscle is being activated. In a maximal voluntary contraction of the biceps brachii, there was a decline of muscle fiber conduction velocity predominantly due to an increase in peripheral fatigue during the first half of contraction that remained constant during the latter half. In contrast, central fatigue was seen to induce a greater force during the latter half of the contraction (Schillings, et al. 2003). The primary peripheral factors contributing to fatigue include metabolic inhibition of the contractile process and failure of excitation-contraction coupling (Kent-Braun 1999). The central nervous system is responsible for driving motorneurons, increased tremor of the exercising limb, recruitment of muscles that were not initially involved in the task, and subjective increase in effort. However, the relative contribution of central factors to the overall level of fatigue is undetermined due to the changing of their influence based on the characteristics of the task at hand. Task variables include the muscle activation pattern, the type of muscle group involved, the type of muscle contraction, and most importantly the intensity and duration of the activity (Martin, et al. 2010).

Although the precise causes and control of muscle fatigue are unclear, it is accepted that a person's fitness status, dietary regimen, muscle fiber type, intensity and duration of exercise are all contributing components to the process (Lepers, Maffiuletti, et al. 2002). A review by Roger M. Enoka and Douglas G. Stuart, "Neurobiology of muscle fatigue", describes task dependency, force-fatigability relationship, muscle wisdom, and sense of effort as major contributors to muscle fatigue (Enoka and Stuart 1992).

Task dependency is the consideration of the details of specific tasks that help determine the underlying mechanisms and sites associated with fatigue. Some task variables include the neural strategy, subject motivation level, activity intensity and duration, speed of contraction, and the extent to which the activity is sustained continuously. The specific mechanisms that cause fatigue may include the CNS drive to motor neurons, the activated muscles and motor units, neuromuscular propagation, excitation-contraction coupling, metabolic substrate availability, the contractile apparatus, and muscle blood flow (Enoka and Stuart 1992).

As a general rule, with some exceptions, the greater the force exerted by a muscle or motor unit, the more the muscle will fatigue. The force-fatigability relationship considers these two variables and the scale to which normal fatigue versus force would follow. It has been discovered that the relationship between force and fatigability may be altered by varying the rate and pattern of the electrical stimulation (Enoka and Stuart 1992).

When presented with a certain task, a person may report an effort-related fatigue even without any impairment of the ability to exert force. A subject may judge a task as too hard simply by the effort required to exert a force rather than the actual magnitude of the force required. This sense of effort is directly related to motor performance in that an increased effort and force failure are associated with a decrease in performance (Enoka and Stuart 1992).

During long bouts of exercise, metabolic changes in the recruited muscle fibers may decrease the force-generating capacity of the skeletal muscle by changing a part of the excitation-contraction coupling process (Lepers, Maffiuletti, et al. 2002). In fatigued muscle, the amplitude of the action potential is smaller compared with that of a muscle unaffected by fatigue (Stephenson, Lamb and Stephenson 1998).When comparing fatigue induced during the excitation-contraction coupling process of fast-twitch and slow-twitch muscle fibers, slow twitch fibers show a higher resistance to fatigue than fast-twitch (Stephenson, Lamb and Stephenson 1998).

Prolonged aerobic exercise will gradually deplete glycogen stores which alters skeletal muscle metabolism. Peripheral fatigue would be expected to occur progressively during this type of exercise. Muscle glycogen depletion is not the exclusive cause of fatigue during prolonged exercise, however. A study that examined the effects of prolonged cycling on neuromuscular parameters found that over a period of five hours, contractile properties of the muscle are significantly altered after the first hour and excitability and central drive are more greatly altered after the first hour to the end of the exercise bout (Lepers, Maffiuletti, et al. 2002).

Muscle fatigue is often contributed to insufficient energy supply due to either a limiting supply of substrate to the tricarboxylic acid cycle or limitations in the tricarboxylic acid cycle activity due to reduced intermediates. Fatigue may be brought about by the inability to generate enough ATP at a sufficient rate to sustain activity level. After prolonged activity to fatigue, the amount of free ADP in the muscle increases significantly from resting levels. The diffusion of ADP is restricted in muscle, and so these significant increases may therefore limit power output during exercise (Sahlin, Tonkonogi and Soderlund 2002).

Fatigue leading to decreases in force output is thought to be influenced by a number of other factors besides insufficient energy supplies. Whether the muscle contraction is voluntary or involuntary, whether the contraction is static or dynamic, sustained or intermittent, maximal or submaximal are all possible contributing factors to the extent of muscle fatigue (Behm and St-Pierre 1998). Muscles with higher percentages of fast-twitch fibers have been seen to fatigue more quickly than muscles composed mostly of slow-twitch fibers (Behm and St-Pierre 1997; Colliander, Dudley and Tetsch 1988; Kupa, et al. 1995). In a study comparing long and short duration muscle fatigue, the long duration fatigue group had a significantly greater decrease in muscle activation than the short duration group. However, there was no difference in the impairment of maximal voluntary contraction force after thirty seconds of recovery. The results of the study concluded that both duration and muscle composition play a role in the rate and extent of muscle fatigue (Behm and St-Pierre 1997).

During a submaximal muscle contraction, an individual is able to recruit more motor units to counteract the reduction in force output that occurs as the muscles fatigue. However, during maximal isometric activity, the individual is unable to increase motor unit recruitment and alterations in firing strategy to individual motor units must be employed in order to alter force production (St Clair Gibson, Lambert and Noakes 2001). In order to optimize a combination of force and duration of the force, there may be a decrease in the firing rate to the recruited muscle fibers (St Clair Gibson, Lambert and Noakes 2001).

Muscle fatigue has been linked with a detrimental effect on neuromuscular control, which may be quantified by measuring postural control. It is believed that muscle fatigue may inhibit the kinesthetic and proprioceptive properties of joints by increasing the threshold of muscle spindle discharge, disrupting afferent feedback, and ultimately resulting in altered conscious joint awareness which affects postural control (Gribble and Hertel, 2004). Muscle fatigue, which is generally considered an internal perturbation, contributes to the displacement of body posture away from equilibrium by destabilizing the body's Center of Mass (COM) (Nardone, Tarantola and Giordano, et al. 1997). The body systems responsible for minimizing the destabilizing effects of fatigue are the proprioceptive, visual, and vestibular systems. When one of these systems is compromised, body sway and muscle activity increase in order to maintain balance

(Corbeil, et al. 2002; Lepers, Bigard, et al. 1997). During gait, sufficient ankle power is needed for forward propulsion in order to obtain appropriate walking velocities (Requaio, et al. 2005) .

In upright standing, the ability to evaluate joint position, movement direction, and speed are critical to maintaining balance (Vuillerme, et al. 2002). It has been proposed that ankle proprioception is critical for the establishment of an internal reference allowing for stabilization of the body with respect to an external gravitational reference (Di Fabio, et al. 1990). A common model is that humans behave as in inverted pendulum with the ankle response being sufficient to counteract minor perturbations that occur during normal stance (Kuo 1993).

Mechanically, the muscle is unable to compensate for torques caused by external perturbations which lead to stance destabilization. Balance is actively controlled by the CNS, which recruits the necessary postural muscles as they are needed (Nardone, Tarantola and Giordano, et al. 1997). A reduced ankle power may impair stability during both gait and quiet standing, which may in turn result in compensatory changes at the knee and hip joints (Bohm and Hosl 2010). The shift from an ankle to a hip strategy due to reduced ankle power output after fatigue may be responsible for applying more stress to the postural control systems because of the increased mass involved in balance control with a hip strategy as compared with ankle strategy as well as an increase in trunk acceleration (Adlerton and Moritz 2003).

When standing is affected by an external perturbation, such as backwards translation of a moving support surface, humans generally respond by moving in the sagittal plane. For a small disturbance, the tendency is to keep the knees, hips, and neck fairly straight and predominantly moving about the ankles (Kuo 1993). This is known as the ankle strategy. For a larger disturbance, humans tend to invoke a motion which coordinates flexion or extension of the hips with smaller concurrent extension or flexion of the ankles and keeping the other joints fairly straight (Kuo 1993). This is known as the hip strategy. The hip strategy is recruited if the ankle strategy is not sufficient to regain balance. In an ankle strategy, the body moves as one stiff segment that may be visualized as in inverted pendulum. The hip strategy is seen as a multi-chain model hinged at the hips (Adlerton and Moritz 2003).

Muscle fatigue at ankle level is known to affect and impair postural control. The deterioration of postural control may be due to the inability to generate or maintain the required force output at the ankle joint and/or a decrease in the function of the proprioceptive system (Vuillerme, et al. 2002). Fatigue places higher demands on the postural control system by increasing the number of actions that must be taken to maintain upright stance (Corbeil, et al. 2002). An increase in postural sway due to localized muscle fatigue is indicative of impaired postural control, which is generally associated with increased fall rates. This is especially true in occupational environments in which fatigue is common and falls may be fatal, because localized muscle fatigue is often induced at a relatively low workload over an extended amount of time (Pline, Madigan and Nussbaum 2006). The maintenance of balance for workers is important so the tasks may be safely performed. A higher fall rate, decreased balance, and decreased task performance are associated with fatigue and are hazardous to workers (Kincl, et al.

2002). Muscular fatigue and postural control deficits may be predispositions to musculoskeletal injury (Gribble and Hertel 2004).

Muscle fatigue due to environmental and job related stresses is a major concern in the workplace, and especially localized muscle fatigue. Some symptoms associated with localized muscle fatigue include loss of force production capabilities, localized discomfort and pain (Chaffin 1973). Localized muscle fatigue may occur at low levels of muscle exertion, even as low as 10% of MVC (M. Redfern 1992). In the occupational setting, muscle exertion levels are often greater than 10% MVC and localized muscle fatigue often manifests and leads to increased postural sway and consequently an increased risk of injury on the job.

The effects of muscle fatigue on balance and posture during exercise and strenuous activity have been assessed by several different researchers (Nardone, Tarantola and Giordano, et al. 1997; Caron 2003; Yaggie and McGregor 2002). A study was conducted by Cham and Redfern which examined the effect of different types of flooring on standing comfort and fatigue for extended durations (Cham and Redfern 2001). Ten healthy participants with no history of lower-extremity or back problems were subjected to four hours of standing on seven different types of flooring conditions while performing computer tasks. Participants wore the same brand and style of socks and hard-sole shoes throughout the testing. The results of the study suggest that flooring properties do affect low-back and lower-leg discomfort and fatigue, but that significant results may only be detected after three hours of standing (Cham and Redfern 2001). Previous studies measured the effects of relatively high rates of fatigue on balance (>50% MVC and/or over >33% maximum aerobic capacity) over a short period of time (<10 minutes) (Vuillerme, et al. 2002;. Yaggie and McGregor 2002). In the workplace, a smaller intensity of fatigue may be experienced over a greater amount of time (Davidson, Madigan and Nussbaum 2004). Even though maintenance of upright stance does not require great effort, it is still a well coordinated task that may be impaired by minor losses in postural muscle force. Some work related tasks require great physical effort for balance control, in which case fatigue would accumulate rapidly. Other work related tasks may accumulate fatigue slowly due to static loading of the postural muscles, which results in decreased muscular force capacity. With a decrease in muscular output, there is an expected decrease in work capability and an internal perturbation to the motor control system with the consequence of impaired motor coordination and possibly the postural control system (Nardone, Tarantola and Galante, et al. 1998).

Footwear

When humans stand, only the feet are in contact with the ground directly. The feet function as sense organs and receptors that play an important role in the maintenance of standing posture. The feet support the body and adjust the center of pressure in order to prevent falls (Hosoda, et al. 1998). Footwear is intended to help maintain proper standing posture. Work boots are designed to accommodate the safety measures needed to work in an industrial setting. The requirements for work boots as determined by the American National Standards Institute (ANSI) include steel toes or metatarsal guards that provide toe protection from impact and compression injuries, oil resistant soles, 28.5 cm laced shaft heights that extend above the ankle joint, and distinct heels (OSHA 1999).

The coefficient of friction (COF) between the shoe and walking surface must provide sufficient resistance to counteract the forward resulting force at the point of contact in order to prevent the occurrence of a slip. It has been reported that mid sole roughness had a positive influence on COF (Abeysekera and Gao 2001). With the focus on meeting safety standards, the work boots may lack the supports found in other shoes needed for balance maintenance and normal gait patterns. Athletic shoes, for example, provide rear foot control, cushioning, shock distribution, and heel stabilization (Divert, et al. 2005). One characteristic commonly seen in orthopedic, sport, hiking or military boots is a high boot shaft. This is worn in order to provide support and stability at the ankle joint, but may compromise the ankle ROM and therefore the ankles' ability to generate power for propulsion (Bohm and Hosl 2010). Work boots tend to have characteristics that enhance support, protection, stability, and structure. For example, a slip-resistant shoe serves to provide traction on dry and wet surfaces. A steel toe is considered the best protection from falling objects or punctures to the foot. It is important for workers to have footwear that maximizes safety as well as decreases the risk of injury related to muscle fatigue and increased postural sway.

There are several features of the shoe that may affect balance and risk of falling. In fact, approximately 45% of falls reported in older people have been attributed to inappropriate footwear and a study of people who had suffered a fall-related hip fracture reported that 75% were wearing poor footwear at the time of the injury (J. C. Menant, S. D. Perry, et al. 2008). Some of the features of shoes that affect balance include sole hardness, heel height, heel collar height, and tread pattern. Manipulation of many of these variables and conditions in which the shoes are worn has been the topic of several researchers interested in their effect on dynamic balance and muscle fatigue (Perry, Radtke and Goodwin 2007; Bohm and Hosl 2010; J. C. Menant, S. D. Perry, et al. 2008). In a study by Menant, et al. examining the effects of shoe features on acute dynamic stability, eleven young adults and fifteen older adults wore six different shoe types and their balance control was assessed using a series of sensorimotor tests. The subjects also performed three walking trials on a level linoleum-covered floor as well as twelve trials over an uneven walkway in each shoe condition. The standard shoe design was an Oxford-style laced shoe with a suede leather upper, ethylene vinyl acetate sole material of average hardness and thickness, low heel collar, square heel, and standard shape smooth sole. The other five shoes were altered to change only one feature: sole hardness, heel height, heel collar height, or sole pattern. It was found that walking on an uneven surface increases the proximity of the center of motion to the edge of the base of support in both younger and older subjects which is an indication that uneven surfaces impose a balance control challenge (J. C. Menant, J. R. Steele and H. Menz, et al. 2008). It was also found that increased shoe heel height and sole softness promotes a more conservative walking pattern and impaired medial-lateral balance control and increased sole hardness, tread sole, and a raised collar height did not improve walking stability in neither the younger nor older subjects. The study concluded that for older people, a low collar and a sole of standard hardness might provide optimal dynamic balance when walking on even and uneven surfaces. (J. C. Menant, J. R. Steele and H. Menz, et al. 2008).

To examine the influence of longer durations, the effect of different types of flooring on standing comfort and fatigue was examined by Redfern and Cham (Cham and

Redfern 2001). Participants stood for four hours on each of seven different flooring conditions while performing computer tasks. After two hours, the floor type had a significant effect on subjective ratings such as lower-leg and lower-back discomfort as well as objective ratings including COP weight shift and lower-extremity skin temperature. In general, the study found that floor mats characterized by increased elasticity, decreased energy absorption, and increased stiffness resulted in less discomfort and fatigue (Cham and Redfern 2001). However, results were only significant after at least two hours of standing time. Further, this study speculated that longer durations of exposure may increase fatigue and discomfort measures above those seen after only two hours.

In a study examining the effects of footwear on standing posture control, Hosoda, et al. assessed dynamic balance among 33 female and 10 male students ages 18 through 23 with various footwear including slippers, with and without clog thongs; leather-soled sandals; and Japanese socks. In the condition of footwear with clog thongs, postural response latencies to horizontal movement of the platform were decreased when compared with that of the slippers without clog thongs condition (Hosoda, et al. 1998). Latency shows the period, in milliseconds, from when the platform starts to move until subjects start to adjust their postures. This study concluded that footwear with thin soles, low heels and clog thongs are the most stable and best for the prevention of unbalanced standing posture.

One occupation with which falling on the job is a major risk for workers is the mining industry. Between 2003 and 2007, 14.4% of the injuries/accidents of underground miners and 22.7% of the surface mine accidents recorded in the Mine Safety and Health Administration database were related to slips or falls of the worker (Merryweather, Yoo and Bloswick 2011). Trips are likely to occur when the swing foot contacts the ground, or when the stance foot prematurely contacts the ground before initial support. Important parameters related to a trip-induced fall include minimum foot clearance, minimum toe clearance and variability, and foot velocity at minimum foot clearance. Irregular surfaces promote an increased muscle activity observed as an additional cocontraction required during gait on irregular surfaces via electromyography, which suggests that fatigue plays an important role in trip-induced falls (Merryweather, Yoo and Bloswick 2011). The cocontraction of both synergistic and antagonistic muscles may have a dramatic impact on implications of muscular activation and force production. (Cham and Redfern, Effect of Flooring on Standing Comfort and Fatigue 2001). With the use of EMGs, Falconer and Winter developed an isometric model that estimated the relative cocontraction between the antagonistic muscles, soleus, and tibialis anterior, acting about the ankle during gait. This model showed that significant levels of cocontraction occur about the ankle during gait and the effects should not be overlooked (Cham and Redfern 2001; Falconer and Winter 1985).

The maintenance of stability when walking requires a person to control their total body center of mass within an ever changing and moving base of support. The human foot is responsible for providing sensory information to the CNS during static and dynamic tasks (Nurse, et al. 2005). Therefore, since footwear is the only barrier between the foot and contact with the ground, the design and type of shoe is important in gait and posture (Bohm and Hosl 2010).

Efficient walking performance depends on the transfer of mechanical power output produced by the musculoskeletal system through footwear. The viscoelastic characteristics of the midsole or the weight of the shoe may influence energy expenditure (Bohm and Hosl 2010), either attenuating or increasing fatigue. Ankle sprains of the lateral ligaments are common injuries during walking and other activities, and a high boot shaft functions biomechanically to restrict excessive range of motion (ROM). For the shaft to provide effective protection laterally, it requires a circular embracing of the ankle which may result in a compromised functional ROM of the ankle. A reduced ROM may lead to a reduction of ankle power which is required for obtaining appropriate walking velocities. It was further theorized that boot shaft stiffness may inhibit the ability of the ankle joint to adapt to uneven surfaces. (Hijmans, et al. 2007). Consequently, propulsion power may be compensated for by changes at the knee and hip joints in order to maintain walking speed and stability (Bohm and Hosl 2010).

The influence of footwear midsole material hardness on dynamic balance control during gait termination was assessed by Perry, Radtke and Goodwin using a barefoot condition as the control (Perry, Radtke and Goodwin 2007). The group tested 12 healthy young females age 20-23 years with the primary outcome measures being COP excursion, and the average vertical force loading rates. The group found that variations of the midsole material and even the presence of a midsole impair the dynamic balance control system (Perry, Radtke and Goodwin 2007). As midsole hardness decreased, there was a decrease in the medial-lateral range of COM movement with respect to the BOS which demonstrates a possible constraint being imposed upon the balance control system. Also, the restrictive nature of the softer midsole material is indicated by a reduction in the maximum COM-COP difference. The soft, standard, and hard midsole material constrains the range of COM over the BOS, which may be required because of the softer interface between the foot and the environment. A softer interface offers less mechanical support that is needed to create reactive forces needed to counteract a range of COM movement as compared to the barefoot condition. The adaptation of the body to reduce the range of COM movement could be detrimental if a balance perturbation is experienced, because an individual wearing footwear with a soft midsole will have an impaired ability to respond effectively (Perry, Radtke and Goodwin 2007).

There has been a substantial amount of research on the effects of muscle fatigue on postural destabilization over a short period of time (Vuillerme, et al. 2002; Yaggie and McGregor 2002; H. Chander, et al. 2010). These effects may include an increase in postural sway (Vuillerme, et al. 2002), instability at the ankle joint (Yaggie and McGregor 2002), and an increase in sway velocity (H. Chander, et al. 2010). One study examined muscle fatigue and postural control deficits in twelve professional firefighters participating in a fire simulation activity. Results showed that there was an overall increase in sway velocity between the pre and post tests, indicating a diminished postural control for both conditions (H. Chander, et al. 2010). However, the leather boots had a significantly smaller increase in sway velocity as compared with the rubber boots condition, suggesting that the heavier rubber boot may cause greater levels of fatigue which impairs balance and control measures.

Wearing shoes has been shown to change the way in which the foot, leg and ankle are used during gait (Divert, et al. 2005; Wolf, et al. 2008). Majumdar et al examined the temporal spatial parameters of gait with barefoot, bathroom slippers and military boots and found that there was a significant increase in step and stride lengths, swing phase and single support time and a decrease in cadence, stance phase and double support time while walking with military boots as compared to barefoot conditions (Majumdar, et al. 2006). The ankle and knee, as well as the joints of the foot and the associated ligaments, tendons, capsules and muscles are subjected to different patterns of stresses in a shod versus barefoot condition . Changed patterns of neural feedback from proprioceptors associated with the ankle, knee, and foot complexes modify the pattern of gait, and this may be observed in limb segment displacements (Majumdar, et al. 2006). From this study it is evident that alterations of sensory information may promote unstable gait patterns and abnormal muscular activation.

When running in shoes, the shoes primarily function to protect the foot against shocks that occur when the heel strikes the ground. Therefore running shoes modify the dynamic characteristics of the contact phase. Slight modifications of mechanical characteristics between different types of shoes may be easily and automatically compensated by a person's neuromuscular control (Divert, et al. 2005). The protections implemented by the various types of shoe designs may require the sacrifice of other beneficial characteristics of the shoe such as a high boot shaft leading to decreased range of motion at the ankle joint (Bohm and Hosl 2010) or a soft sole resulting in impaired balance (J. C. Menant, J. R. Steele and H. Menz, et al. 2008), possibly resulting in postural destabilization. However, walking barefoot has not been proven to be a safe alternative to wearing shoes, as barefoot walking minimizes external protection and shock reduction (J. C. Menant, J. R. Steele and H. B. Menz, et al. 2009; Majumdar, et al. 2006; Divert et al. 2005).

There have been several studies assessing different types of footwear and the effects of the altered sensory input on different types of surfaces and risk of falling (J. C. Menant, J. R. Steele and H. B. Menz, et al. 2009; Koepsell, et al. 2004, Tencer, et al. 2004) but there have been few studies examining the impact of different types of footwear on postural balance with extended durations in relation to an occupational setting. Specifically, the aim of the current study was to examine the effects of extended durations in different footwear on fatigue of the plantarflexor and dorsiflexor muscle groups.

CHAPTER III

METHODS

The purpose of this study was to examine the effects of wearing two different types of footwear for extended durations on a hard surface on muscle activation and muscle fatigue by use of EMG measurements. The activity levels of selected muscles of the lower extremities (plantarflexors and dorsiflexors) due to the prolonged exposure of walking on a hard surface were analyzed and EMG activity was recorded.

Participants

The participants in the study were fourteen healthy, college aged, adult males who were recruited based on an anthropometric blocked assignment. As per the Institutional Review Board regulation, written informed consent was acquired. Participants were excluded based on the following criteria: cardiovascular, musculoskeletal, pulmonary, orthopedic, and neurological abnormalities, which included but were not limited to disease of the vestibular system and any complications with walking or standing which might inhibit successful testing session completion and normal gait and/or postural control. Table 1 provides relevant participant demographics.

Subject Demographics	Mean ± SD
Age (years)	23.6 ± 1.2
Weight (kg)	89.2 ± 14.6
Height (cm)	181 ± 5.3

Table 1. Subject demographics
SD=standard deviation

Instrumentation

The Noraxon TeleMyo 2044T G2 (Noraxon U.S.A., Inc. Scottsdale, AZ) was used to record electromyography (EMG) data in the Applied Biomechanics Laboratory (ABL) in the Department of Health, Exercise Science and Recreation at the University of Mississippi. The TeleMyo device consisted of a transmitter that accommodated up to sixteen channels for EMG leads. A hardware band pass filter of 20-250 Hz and notch filter at 60 Hz was applied. The sampling frequency was set to 1500 Hz. The transmitter was connected to a computer by way of the Noraxon PC interface, and all EMG data were recorded on the computer with the program MyoResearch XP.

EMG data were recorded from the muscles in the right leg, including the Tibialis Anterior (TA) and the Medial Head of Gastrocnemius (MG). Proper electrode placement was confirmed using a sub-maximal exertion test, followed by an MVC test.

Experimental Conditions

Participants were tested in two different conditions of footwear: the Steel-Toed Work Boots (WB) and Low-Top Flat Sole Slip-Resistant Shoes (LT). The WB were equipped with steel toes or metatarsal guards which protect the toe from compression and impact injuries, oil-resistant soles, a 28.5 centimeter laced shaft which extended beyond the ankle joint, and distinctive heels. These boots met ANSI-Z41-1991 standards as per regulations of OSHA for footwear safety and protection (Occupational Safety and Health Administration. U.S. Department of Labor). The LT is commonly worn in a varied population for which there is no prescribed ANSI standard (OSHA 1999) the footwear must meet. The average footwear was size 11, and table 2 lists the characteristics of this footwear.

Shoe	LT	WB
Mass (kg)	0.4	0.9
Boot Shaft Height (cm)	9.5	18.5
Heel Sole Width (cm)	8.5	9.6
Forefoot Sole Width (cm)	10.5	12
Heel Height (cm)	2.0	3.2

Table 2. Footwear Characteristics

Experimental Testing Procedures

The testing procedures for each subject included a study design with repeated measures, with duration of exposure to a hard firm surface as a nine level independent variable. The time intervals included: pre-test, 30, 60, 90, 120, 150, 180, 210, and 240 minutes of exposure wearing both types of footwear.

Each subject was marked with surface electrodes placed 2 centimeters apart over the bellies of selected muscles of the right lower extremity, including the medial gastrocnemius (MG) to measure muscle activation during maximal plantarflexion and the tibialis anterior (TA) to measure muscle activation during maximal dorsiflexion. The subjects executed an isometric MVC prior to the start of the walking session as a pre-test measure (Pre) and then again every 30 minutes for the duration of 4 hours. The same protocol was followed for both the WB and LT conditions. The participants were instructed to walk at their own pace on the hard surface during the times between MVCs. A minimum of 72 hours of rest from testing was given between testing conditions. Each type of footwear and socks were provided to participants before each testing session for control purposes.

Data Processing

Data was collected for the MVC of the plantarflexors and dorsiflexors for 3 trials for each muscle group at each 30 minute time interval (Pre to minute 240). The data was filtered with a band-pass filter between 20-250 Hz. After filtering, the data was then rectified. RMS was calculated by taking the square root of the sum of squares. The following formula was used to calculate RMS:

$$RMS = \sqrt{\frac{1}{n}\sum_{i=0}^{n}\left(mV_i - mV_{avg}\right)^2}$$

Statistical Analysis

In order to establish if any differences in muscle activation occurred over time between both types of footwear (WB and LT), a repeated measures analyses of variance (RMANOVA) was completed. In order to determine if differences occurred within the exposure time and/or the shoe types, muscle activation dependent variables were evaluated using a 2 x 9 (Shoe [WB v. LT] x Extended duration of walking intervals [Pre, 30, 60, 90, 120, 150, 180, 210, and 240]). If the Mauchly's test of sphericity was violated, a Greenhouse-Geisser correction was utilized to determine significance. A Bonferroni post-hoc analysis was performed if significance was found in order to locate the significance. Significance was set at an alpha level of $p=0.05$ for all analyses performed; the SPSS 20 statistical software package was used for all statistical analyses.

CHAPTER IV

RESULTS

Muscle Activation of the Plantarflexors and Dorsiflexors

A pairwise comparison with a Bonferroni correction determined that there was no significant difference with the time-shoe interaction and no statistically significant differences among the time points for either footwear condition (WB and LT) with time as the main effect. However, a significant difference with shoe type as the main effect with activation of both the dorsiflexors (p=0.032) and plantarflexors (p=0.087). For both muscle groups, the LT showed a greater muscle activation compared to the WB condition. Table 1 and table 2 show the root mean square (RMS) of the muscle activity of the dorsiflexors and plantarflexors over time.

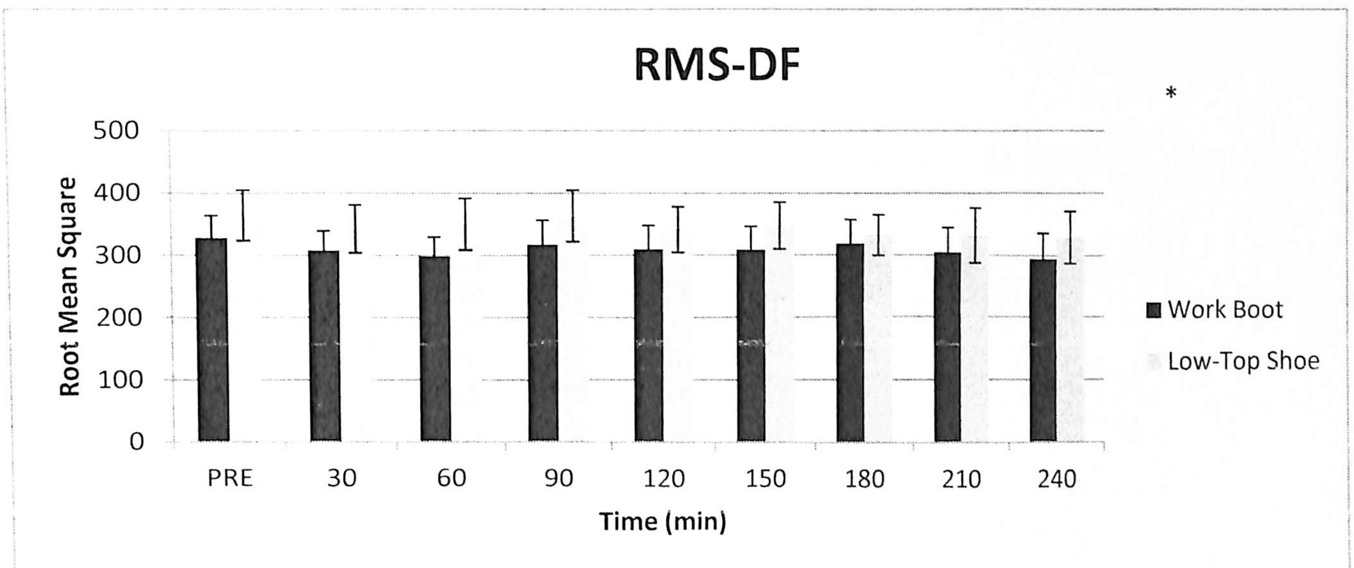

Figure 1. Root Mean Square of Muscle Activation of the Dorsiflexors Over the 9 Time Points of Interest. * Indicates significance between the two footwear types, with standard error bars.

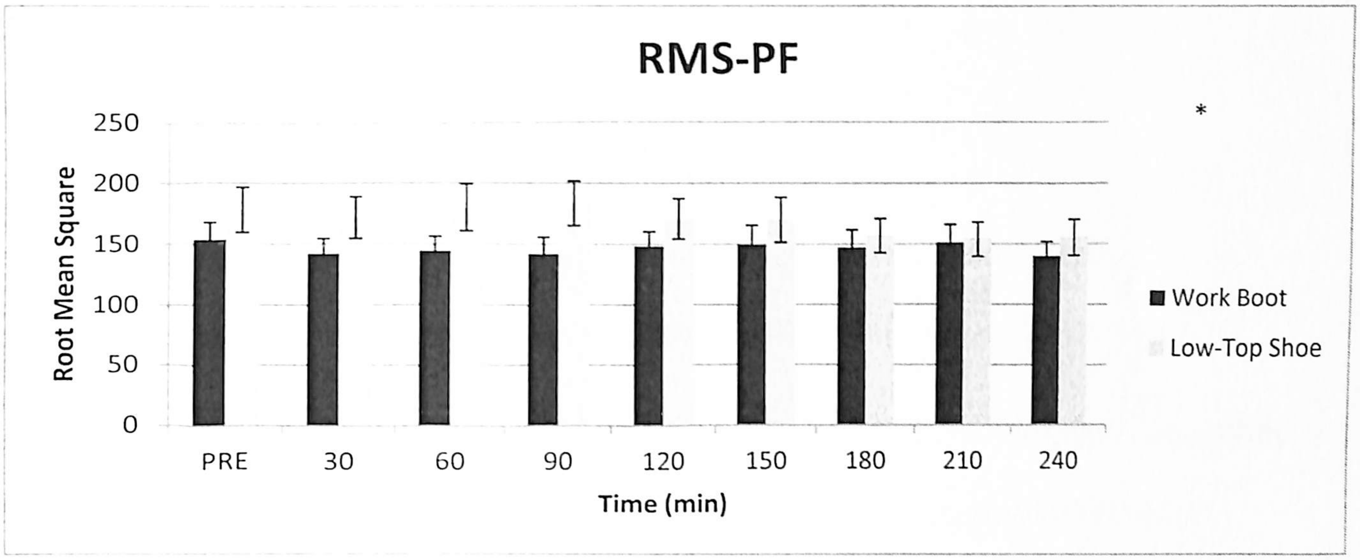

Figure 2. Root Mean Square of Muscle Activation of the Plantarflexors Over the 9 Time Points of Interest. * Indicates significance between the two footwear types, with standard error bars.

CHAPTER V

DISCUSSION

The purpose of this study was to investigate the effects of walking on a hard firm surface for extended durations in two different types of commonly used industrial footwear on muscle fatigue. The specific objective of the study was to analyze muscle activity of the lower leg after exposure to extended duration of gait and stance in different types of footwear. The results showed that there was neither a significant difference in muscle activation for either footwear condition with time as the main effect nor a significant difference with the time-shoe interaction. The results did show a statistically significant difference with shoe type as the main effect. The discussion addresses both muscle activation over time and muscle activation between footwear types.

Muscle Activation between footwear types

For both the plantarflexors and dorsiflexors the LT showed greater muscle activation as compared to the WB condition. The LT condition was characterized by a lighter mass (0.4 kg), lower boot shaft (9.5 cm), and a lower heel height (2 cm) as compared with the WB condition with a relatively heavier mass (0.9 kg), higher boot shaft (18.5 cm), and greater heel height (3.2 cm).

The mass of a shoe may have a significant effect on muscle activity of the lower extremities. A study on the influence of firefighter boot type on postural measures tested twelve professional firefighters in two shoe conditions, one a leather boot weighing 2.44 kg and the other a rubber boot weighing 2.93 kg. The results suggested that the heavier boots elicited greater fatigue as compared to the lighter leather boots (H. Chander, et al. 2010). It has been shown that for every gram of increase in mass of the footwear, an increase in energy expenditure of 0.7-1.0% occurs (Jones, et al. 27, Hamill and Bensel 1996). This may be a potential causative mechanism with fatigue between footwear types. However, the effects of footwear mass on muscle fatigue have been studied only relative to the knee and hip joints. Therefore, mass is not expected to be a contributing factor to differences in muscle activity levels of the ankle musculature between shoe types.

Shoe shaft height may have a significant effect on fatigue levels of the ankle musculature. A high boot shaft may be worn to provide stability and support at the ankle joint in order to prevent excessive inversion. In turn, ROM at the ankle joint may be diminished and consequently the ability of the ankle joint to generate propulsion power declines (Bohm and Hosl 2010). With a high boot shaft, muscle activity at the ankle joint is minimized and fatigue of those muscles is expected to be minimal. This is consistent with our results, as a boot with a higher shaft showed potential for resisting fatigue for longer durations. While a high boot shaft may minimize fatigue at the ankle joint, restriction of the ankle may cause a shift to a knee or hip strategy (Bohm and Hosl 2010). In terms of balance decrements, fatigue of the hip musculature results in a more significantly reduced control of medial/lateral sway of COP as compared with fatigue of the ankle musculature (Gribble and Hertel 2004). This suggests that fatigue of the hip joint could potentially lead to more falls than fatigue of the ankle joint.

Heel height may also impact muscle activation levels. As heel height increases, the ankle is forced into plantarflexion causing continuous activation of the plantarflexors even in quiet standing. Therefore, a greater heel height is expected to result in greater levels of muscle activity, which is thought to contribute to decreased balance performances over time (Parijat and Lockart 2008). This notion is supported by other studies that found increased postural sway and decreased balance performance in shoes with an elevated heel height (J. C. Menant, et al. 2008, Lord and Bashford 1996). These findings are consistent with the results of the current studying in which the WB with the greater heel height showed relatively less muscle activation compared to the lower heel height condition.

Muscle activation over time

Our results showed no significant difference in muscle activation with time as the main effect. This is consistent with Cham and Redfern's study on the effect of flooring on standing comfort and fatigue. They found no statistically significant changes in the median frequency of the EMG power spectrum with time (Cham and Redfern 2001). Cham and Redfern attribute the results to the possibility that the procedure of collecting EMG data during a specific percentage of an MVC may cause fatigue in itself and this may mask standing fatigue effects, as well as skin impedance changes over a long period of time which may make EMG signals unreliable (Cham and Redfern 2001). They also considered that spectral shifts in EMG are most commonly seen for dynamic tasks that are characterized by high muscle contraction levels over short durations of time. In their study, the intensity of the workload may not have been sufficient to cause changes in EMG that could be attributed to fatigue (Cham and Redfern 2001). This is consistent for the results of the current study as well. The results do not support the hypothesis that muscle activity of the plantarflexors and dorsiflexors will decrease over time.

The results of the study are inconsistent with previous studies in which muscle activity was shown to increase over time (Behm and St-Pierre 1997; Lepers, Maffiuletti, et al. 2002; Pline, Madigan and Nussbaum 2006). One possible contributing factor is the time over which muscle activation was elicited. Behm and St-Pierre considered long duration fatigue over a span of 19 minutes 30 seconds during which fatigue was induced by a submaximal fatigue protocol (Behm and St-Pierre1997). In an occupational setting, muscle activity most often occurs over extended durations of several hours. It has been shown that an increased time of muscle activity leading to fatigue results in greater decreases in muscle activation (Behm and St-Pierre 1997). The studies by Pline, et al. and Lepers, et al. both used exercise protocols (back extensions and cycling, respectively) to induce fatigue (Lepers, Maffiuletti, et al. 2002; Pline, Madigan and Nussbaum 2006). With extended duration standing and walking, muscle exertion is significantly lower as compared with exercise activity. This is the major difference among the present study and other previous studies that focused on muscle activity. Also, the subjects participating in the study by Lepers, et al. consisted of endurance-trained cyclists or triathletes (Lepers, Maffiuletti, et al. 2002). The population of the present study was recruited on a voluntary basis with no previous physical training requirements. These factors may account for some differences seen in our results as compared to other previous studies.

Conclusions

The results of this study may be beneficial in the aid of efficient design of future footwear to be worn in occupational settings. Cham and Redfern designed a study in which the floor type was manipulated (Cham and Redfern, Effect of Flooring on Standing Comfort and Fatigue 2001). However, manipulation of footwear is much more efficient and realistic in an occupational setting as compared with reconstruction of entire work sites. In future studies, it may be beneficial to examine other variables of the shoe such as mid sole hardness and rear foot motion and these effects on muscle activation and postural stability. The results of the present study may help clarify the contributing factors of some shoe characteristics to muscle fatigue levels in an occupational setting for extended amounts of time.

BIBLIOGRAPHY

Abeysekera, John, and Chuansi Gao. "The identification of factors in the systematic evaluation of slip prevention on icy surfaces." *International Journal of Industrial Ergonomics*, 2001: 303-313.

Adlerton, Anna-Karin, and Ulrich Moritz. "Forceplate and accelerometer measures for evaluating the effect of muscle fatigue on postural control during one-legged stance." *Physiotherapy Research International*, 2003: 187-199.

Arend-Nielsen, Lars, and Kerry R. Mills. "Muscle fibre conduction velocity, mean power frequency, mean EMG voltage and force during submaximal fatiguing contractions of human quadriceps." *European Journal of Applied Physiology*, 1988: 58: 20-25.

Asmussen, E., and B. Mazin. "A Central Nervous Component in Local Muscle Fatigue." *Journal of Applied Physiology*, 1978: 9-15.

Behm, D.G., and D.M.M. St-Pierre. "Effects of fatigue duration and muscle type on voluntary and evoked contractile properties." *Journal of Applied Physiology*, 1997: 1654-1661.

Behm, D.G., and D.M.M. St-Pierre. "Effects of fatigue duration and muscle type on voluntary and evoked contractile properties." (Journal of Applied Physiology) 82 (1997): 1654-1661.

Behm, David G., and Diane M.M. St-Pierre. "Fatigue Mechanisms in Trained and Untrained Plantar Flexors." *Journal of Strength and Conditioning Research*, 1998: 166-172.

Bigland-Ritchie, B., E.F. Donovan, and C.S. Roussos. "Conduction velocity and EMG power spectrum changes in fatigue of sustained maximal efforts." *J. of Applied Physiology*, 1981: 51: 1300-1305.

Bigland-Ritchie, B., F. Furbush, and J.J. Woods. "Fatigue of intermittent submaximal voluntary contractions: central and peripheral factors." *Journal of Applied Physiology*, 1986: 421-429.

Bigland-Ritchie, B., R. Johansson, O.C.J. Lippold, and J.J. Woods. "Contractile Speed and EMG Changes During Fatigue of Sustained Maximal Voluntary Contractions." *Journal of Neurophysiology*, 1983: 313-324.

BLS. "Workplace Injuries and Illnesses, 2010." *Bureau of Labor Statistics*, October 20, 2011: 1.

Bohm, Harald, and Matthias Hosl. "Effect of boot shaft stiffness on stability joint energy and muscular co-contraction during walking on an uneven surface." *Journal of Biomechanics*, 2010: 2467-2472.

Bohm, Harald, and Matthias Hosl. "Effect of boot shaft stiffness on stability joint energy and muscular co-contraction during walking on uneven surface." (Journal of Biomechanics) 43, no. 2467-2472 (2010).

Caron, O. "Effects of local fatigue of the lower limbs on postural control and postural stability in standing posture." *Neuroscience Letters*, 2003: (340) 83-86.

Chaffin, DB. "Localized muscle fatigue: definition and measurement." *Journal of Occupational Medicine*, 1973: 15: 346-354.

Cham, Rakie, and Mark S. Redfern. "Effect of Flooring on Standing Comfort and Fatigue." *Human Factors: The Journal of the Human Factors and Ergonomics Society*, 2001: 43: 381-391.

Cham, Rakie, and Mark S. Redfern. "Effect of Flooring on Standing Comfort and Fatigue." *Human Factors: The Journal of the Human Factors and Ergonomics Society*, 2001: 381-391.

Chambers, April J., and Rakie Cham. "Slip-related muscle activation patterns in the stance leg during walking." *Gait and Posture*, 2006: 565-572.

Chander, H., C. Wade, J.C. Garner, R. Garten, and E. Acevedo. "The influence of firefighter boot type on posture measures." 2010.

Chander, Harish, Chip Wade, John C. Garner, Ryan Garten, and Ed Acevedo. "The influence of firefighter boot type on postural measures." 2010.

Colliander, E.B., G.A. Dudley, and P.A. Tetsch. "Skeletal muscle fiber type composition and performance during repeated bouts of maximal, concentric contractions." *Eur. J. Appl. Physiol. Occup. Physiol.*, 1988: 81-86.

Corbeil, Philippe, Jean-Sebastien Blouin, Francois Begin, Vincent Nougier, and Normand Teasdale. "Perturbation of the postural control system induced by muscular fatigue." (Gait and Posture) 18, no. 92-100 (2002).

Davidson, B.S., M.L. Madigan, and M.A. Nussbaum. "Effects of lumbar extensor fatigue and fatigue rate on the postural sway." *European Journal of Applied Physiology,* 2004: (93) 183-189.

De Luca, Carlo J. "Physiology and Mathematics of Myoelectric Signals." *IEEE Transactions on Biomedical Engineering*, 1979: 313-326.

Di Fabio, R.P., M.B. Badke, A. McEvoy, and A. and Breunig. "Influence of local sensory afference in the calibration of human balance responses." *Experimental Brain Research*, 1990: 80: 591-599.

Divert, C., G. Mornieux, H. Baur, F. Mayer, and A. Belli. "Mechanical Comparison of Barefoot and Shod Running." *International Journal of Sports Medicine* , 2005: 593-598.

Edwards, R.G., and O.C.J. Lippold. "The Relation between Force and Integrated Electrical Activity in Fatigued Muscle." *J. Physiol.*, 1956: 677-681.

Enoka, Roger M, and Douglas G. Stuart. "Neurobiology of muscle fatigue." *Journal of Applied Physiology*, 1992: 1631-1648.

Falconer, K, and DA Winter. "Quantitative assessment of cocontraction at the ankle during walking." *Electromyogr Clin Neurophysiol*, 1985: 25: 135-149.

Farina, Dario, Mauro Fosci, and Roberto Merletti. "Motor unit recruitment strategies investigated by surface EMG variables." *Journal of Applied Physiology*, 2001: 235-247.

Fingerhut, Lois A., Christine S. Cox, and Margaret Warner. "International Comparative Analysis of Injury Mortality: Findings From the ICE on Injury Statistics." (National Center for Health Statistics) 303 (1998).

Gefen, Amit, M. Megido-Ravid, Y. Itzchak, and M. Arcan. "Analysis of muscular fatigue and foot stability during high-heeled gait." *Gait and Posture*, 2002: 56-63.

Gribble, Phillip A., and Jay Hertel. "Effect of hip and ankle muscle fatigue on unipedal postural control." *Journal of Electromyography and Kinesiology*, 2004: 641-646.

Gribble, Phillip A., and Jay Hertel. "Effect of hip and ankle muscle fatigue on unipedal postural control." (Journal of Electromyography and Kinesiology) 14, no. 6 (2004).

Gribble, Phillip A., and Jay Hertel. "Effect of Lower-Extremity Muscle Fatigue on Postural Control." *Arch Phys Med Rehabi*, 2004: 85: 589-92.

Hamill, Joseph, and Carolyn K. Bensel. *Biomechanical analysis of military boots: phase III.* Science and Technology Directorate, Natick: United States Army Natick, 1996.

Han, Tai Ryoon, Nam Jong Paik, and Mi Sik Im. "Quantification of the path of center of pressure (COP) using an F-scan in-shoe transducer." (Gait and Posture) 10, no. 3 (1999).

Hijmans, J.M., J.H. Geertzen, P.U. Dijkstra, and K. Postema. "A systematic review of the effects of shoes and other ankle or foot appliances on balance in older people and people with peripheral nervous system disorders." *Gait and Posture*, 2007: 316-323.

Hosoda, Masataka, et al. "The effects of footwear on standing posture control." *J. Phys. Ther. Sci.*, 1998: 47-51.

Jones, B.H., M. M. Toner, W. L. Daniels, and J.J. Knapik. "The energy cost and heart-rate response of trained and untrained subjects walking and running in shoes and boots." *Ergonomics*, 27: 895-902.

Kent-Braun, Jane A. "Central and peripheral contributions to muscle fatigue in humans during sustained maximal effort." *European Journal of Applied Physiology*, 1999: 80:57-63.

Kincl, Laurel D., Amit Bhattacharya, Paul A. Succop, and Scott C. Clark. "Postural Sway Measurements: A Potential Safety Monitoring Technique for Workers Wearing Personal Protective Equipment." *Applied Occupational and Environmental Hygiene*, 2002: 256-266.

Koepsell, Thomas D., et al. "Footwear Style and Risk of Falls in Older Adults." *Journal of the American Geriatrics Society*, 2004: 1495-1501.

Kuo, Arthur D. "An Optimal Control Model for Analyzing Human Postural Balance." *IEEE Transactions of Biomedical Engineering*, 1993: (42) 87-101.

Kupa, E.J., S.H. Roy, S.C. Kandarian, and de Luca. C.J. "Effects of muscle fiber type and size on EMG median frequency and conduction velocity." *Journal of Applied Physiology*, 1995: 23-32.

Leigh, J. Paul, Steven B. Markowitz, and Marianne Fahs. "Occupational Injury and Illness in the United States: Estimates of Costs, Morbidity, and Mortality." *Archives of Internal Medicine*, 1997: 1557-1568.

Lepers, Romuald, Andre X. Bigard, Jean-Pierre Diard, Jean-Francois Gouteyron, and Charles Y. Guezennec. "Posture control after prolonged exercise." (European Journal of Applied Physiology) 76, no. 55-61 (1997).

Lepers, Romuald, Nicola A. Maffiuletti, Ludovic Rochette, Julien Brugniaux, and Guillaume Y. Millet. "Neuromuscular fatigue during a long-duration cycling exercise." (Journal of Applied Physiology) 92, no. 1487-1493 (2002).

Lepers, Romuald, Nicola A. Maffiuletti, Ludovic Rochette, Julien Brugniaux, and Guillaume Y. Millet. "Neuromuscular fatigue during a long-duration cycling exercise." (J Appl Physiol) 92, no. 1487-1493 (2002).

Lin, Dingding, Hyang Seol, Maury A. Nussbaum, and Michael L. Madigan. "Reliability of COP-based postural sway measures and age-related differences." *Gait and Posture*, 2008: 337-342.

Lord, S.R., and G.M. Bashford. "Shoe characteristics and balance in older women." (J Am Geriatr Soc) 44, no. 4 (1996).

Majumdar, Dhurjati, Pratul K. Banerjee, Deepti Majumdar, Madhusudan Pal, Rakesh Kumar, and William Selvamurthy. "Temporal spatial parameters of gait wth barefoot, bathroom slippers and military boots." *Indian J Physiol Parmacol*, 2006: 33-40.

Maki, BE, and WE McIlroy. "Postural control in the older adult." *Clinics in Geriatric Medicine*, 1996: 635-658.

Maki, Brian E., et al. "Interventions to Promote More Effective Balance Recovery Reactioons in Industrial Settings: New Perspectives on Footwear and Handrails." *Industrial Health*, 2008: 40-50.

Martin, Vincent, et al. "Central and peripheral contributions to neuromuscular fatigue induced by a 24-h treadmill run." *Journal of Applied Physiology*, 2010: 1224-1233.

Menant, J, S Perry, J Steele, H Menz, B Munro, and S Lord. "Effects of Shoe Characteristics on Dynamic Stability When Walking on Even and Uneven Surfaces in Young and Older People." *Phys Med Rehabil;89:1970-6.*, 2008: 1970-1976.

Menant, Jasmine C,, Julie R. Steele, Hylton B. Menz, Bridget J. Munro, and Stephen R. Lord. "Effects of Footwear Features on Balance and Stepping in Older People." (Gerontology) 54 (2008).

Menant, Jasmine C., Julie R. Steele, Hylton B. Menz, Bridget J. Munro, and Stephen R. Lord. "Effects of walking surfaces and footwear on temoro-spatial gait parameters in young and older people." *Gait and Posture*, 2009: 392-397.

Menant, Jasmine C., Julie R. Steele, Hylton Menz, Bridget J. Munro, and Stephen R. Lord. "Effects of Footwear Features on Balance and Stepping in Older People." *Gerontology*, 2008: 18-23.

Menant, Jasmine C., Stephen D. Perry, Julie R. Steele, Hylton B. Menz, Bridget J. Munro, and Stephen R. Lord. "Effects of Shoe Characteristics on Dynamic Stability When Walking of Even and Uneven Surfaces in Young and Older People." *Arch Phys Med Rehabil vol. 89*, 2008: 1970-1976.

Merryweather, Andrew, Byungju Yoo, and Donald Bloswick. "Gait Characteristics Associated with Trip-Induced Falls on Level and Sloped Irregular Surfaces." *Minerals*, 2011: 109-121.

Merton, P.A. "Voluntary Strength and Fatigue." *The Journal of Physiology*, 1954: 553-564.

Mills, K.R. "Power Spectral Analysis of Electromyogram and Compound Muscle Action Potential during Muscle Fatigue and Recovery." *J. Physiol.* , 1982: 401-409.

Nardone, Antonio, Jessica Tarantola, Andrea Giordano, and Marco Schieppati. "Fatigue effects on body balance." (Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control) 104, no. 4 (1997).

Nardone, Antonio, Jessica Tarantola, Massimo Galante, and Marco Schieppati. "Time Course of Stabilometric Changes After a Strenuous Treadmill Exercise." *Arch Phys Med Rehabil*, 1998: (79) 920-924.

Nurse, Matthew A., Manuel Hulliger, James M. Wakeling, Benno, M. Nigg, and Darren J. Stefanyshyn. "Changing the texture of footwear can alter gait patterns." *Journal of Electromyography and Kinesiology*, 2005: 496-506.

OSHA. "Personal Protective Equipment." *Occupational Safety and Health Administration.* 1999. http://www.osha.gov/Publications/osha3151.pdf.

Parijat, Prakriti, and Thurmon E. Lockart. "Effects of quadriceps fatigue on the biomechanics of gait and propensity." 28, no. 4 (2008).

Perry, Stephen D., Alison Radtke, and Chris R. Goodwin. "Influence of footwear midsole material hardness on dynamic balance control during unexpected gait termination." *Gait and Posture*, 2007: 94-98.

Pline, Kevin M., Michael L. Madigan, and Maury A. Nussbaum. "Influence of fatigue time and level on incerases in postural sway." (Ergonomics) 49, no. 15 (2006).

Pline, Kevin M., Michael L. Madigan, and Maury A. Nussbaum. "Influence of fatigue time and level on increases in postural sway." (Ergonomics) 49 (2006).

Redfern, Mark. "Functional Muscle: Effects on Electromyographic Output." In *Selected Topics in Surface Electromyography for use in the Occupational Setting: Expert Perspectives*, 179. U.S. Department of Health and Human Services, 1992.

Redfern, Mark S., et al. "Biomechanics of Slips." *Ergonomics*, 2001: 1138-1166.

Requaio, L.F., S. Nadeau, M.H. Milot, D. Gravel, D. Bourbonnais, and D. Gagnon. "Quantification of level of effort at the plantarflexors and hip extensors and flexor muscles in healthy subjects walking at different cadences." *J. Electromyogr. Kinesiol.*, 2005: 393-405.

Sahlin, K., M. Tonkonogi, and K. Soderlund. "Energy supply and muscle fatigue in humans." *Acta Physiologica*, 2002: 261-266.

Schillings, Maartje L., Wouter Hoefsloot, Dick F. Stegeman, and Machiel J. Zwarts. "Relative contributions of central and peripheral factors to fatigue during a maximal sustained effort." *European Journal of Applied Physiology*, 2003: 562-568.

St Clair Gibson, Alan, Michael I. Lambert, and Timothy D. Noakes. "Neural Control of Force Output During Maximal and Submaximal Exercise." *Sports Med*, 2001: 637-650.

Stephenson, D.G., G.D. Lamb, and G.M.M Stephenson. "Events of the excitation–contraction–relaxation (E–C–R) cycle in fast- and slow-twitch mammalian muscle fibres relevant to muscle fatigue." *Acta Physiologica Scandinavica* , 1998: 229.

Swaen, G.M.H., L.G.P.M van Amelsvoort, U. Bultmann, and I.J. Kant. "Fatigue as a risk factor for being injured in an occupational accident: results from the Maastricht Cohort Study." (Occup Environ Med) 60 (2003).

Tencer, Allan F., et al. "Biomechanical Properties of Shoes and Risk of Falls in Older Adults." *Journal of the American Geriatrics Society*, 2004: 1840-1846.

Vøllestad, Nina K. "Measurement of human muscle fatigue." *Journal of Neuroscience Methods*, 1997: 219-227.

Vuillerme, Nicolas, Frederic Danion, Nicolas Forestier, and Vincent Nougier. "Postural sway under muscle vibration and muscle fatigue in humans." *Neuroscience Letters*, 2002: 131-135.

Wade, Chip, Ryan Garten, Scott Breloff, and Ed Acevedo. "A comparison of lower extremity fatigue between leather and rubber boots in professional firefighters." n.d.

Winter, D.A. "Human balance and posture control during standing and walking." (Gait and Posture) 3, no. 193-214 (1995).

Wolf, Sebastian, Jan Simon, Dimitrios Patikas, Waltraud Schuster, Petra Armbrust, and Leonhard Doderlein. "Foot motion in children shoes-A comparison of barefoot walking with shod walking in conventional and flexible shoes." *Gait and Posture*, 2008: 51-59.

Yaggie, James A., and Stephen J. McGregor. "Effects of Isokinetic Ankle Fatigue on the Maintenance of Balance and Postural Limits." (Arch Phys Med Rehabil) 83 (2002).

Zwarts, M.J., G. Bleijenberg, and B.G.M. van Engelen. "Clinical neurophysiology of fatigue." *Clinical Neurophysiology*, 2008: 2-10.