August 2016

Examining the Effects of Body Weight Support and Speed on Physiological Measures and Lower Leg Muscular Activity

Michael Thomas Soucy
University of Nevada, Las Vegas

EXAMINING THE EFFECTS OF BODY WEIGHT SUPPORT AND SPEED ON PHYSIOLOGICAL MEASURES AND LOWER LEG MUSCULAR ACTIVITY

By

Michael Thomas Soucy

Bachelor of Science in Exercise Science
California Lutheran University
2014

A thesis submitted in partial fulfillment
of the requirement for the

Master of Science – Kinesiology

Department of Kinesiology and Nutrition Sciences
School of Allied Health Sciences
Division of Health Sciences
The Graduate College

University of Nevada, Las Vegas
August 2016

UNLV | GRADUATE COLLEGE

Thesis Approval

The Graduate College
The University of Nevada, Las Vegas

July 13, 2016

This thesis prepared by

Michael Thomas Soucy

entitled

Examining the Effects of Body Weight Support and Speed on Physiological Measures and Lower Leg Muscular Activity

is approved in partial fulfillment of the requirements for the degree of

Master of Science - Kinesiology
Department of Kinesiology and Nutrition Sciences

John Mercer, Ph.D.
Examination Committee Chair

Julia Freedman Silvernail, Ph.D.
Examination Committee Member

James Navalta, Ph.D.
Examination Committee Member

Jennifer Pharr, Ph.D.
Graduate College Faculty Representative

Kathryn Hausbeck Korgan, Ph.D.
Graduate College Interim Dean

ABSTRACT

Examining the Effects of Body Weight Support and Speed on Physiological Measures and Lower Leg Muscle Activity

By

Michael Thomas Soucy

Dr. John Mercer, Examination Committee Chair
Professor, Associate Dean of Allied Health Sciences
University of Nevada, Las Vegas

The purpose of this study was to determine if body weight support or speed influences specific physiological and biomechanical parameters while running at a preferred pace, or variations of preferred pace. Nine participants (age: 28.56 ± 7.88 years, height: 1.68 ± 0.08 cm, mass: 65.70 ± 7.64 kg) who were running a minimum of 10 miles per week (14.67 ± 4.92 miles) were recruited for participation. Participants were asked to sign an institutionally approved informed consent form upon arrival to the lab. After determining preferred running speed at each body weight support condition (no support, 10%, and 20% support), participants were instrumented with four Delsys EMG leads, a PCB one-dimensional accelerometer, and a K4B2 portable metabolic gas analysis system to measure muscle activity, tibial acceleration, and metabolic variables at various points throughout each running condition. Muscle activity was recorded for the rectus femoris, semitendinosus, tibialis anterior, and medial gastrocnemius muscles. Participants were asked to run a total of nine conditions, running with no body weight support, 10% support, and 20% support at a preferred running pace, as well as +10% and -10% of that preferred speed. Average muscle activity, average tibial acceleration, and average VO_2 and VCO_2 values were determined for each condition. None of the dependent variables

were influenced by the interaction of speed and body weight support ($p>0.05$). EMG of the four lower extremity muscles was not influenced independently by body weight support ($p>0.05$) or speed ($p>0.05$). VO_2 was influenced by body weight support ($p<0.001$), reporting VO_2 during running at 0% support was significantly higher than VO_2 at both 10% body weight support ($p<0.05$) and 20% body weight support ($p<0.05$). Tibial acceleration was not influenced by body weight support ($p>0.05$) but was influenced by speed ($p<0.05$). Specifically, tibial acceleration was significantly higher during the +10% of preferred speed condition compared with both the preferred speed condition ($p<0.05$) and the -10% of preferred speed condition ($p<0.05$). It is concluded that oxygen consumption was affected by both body weight support and speed; yet, lower extremity muscle activity was not. It should be understood that although it may be possible to achieve similar levels of muscle activation while running at 10% and 20% support, the metabolic demand is continually less and less as body weight support is added.

Acknowledgements

First and foremost I would like to thank my Graduate Advisor, Dr. John Mercer. Dr. Mercer has provided me with the tools and guidance to be successful throughout our time working together, and none of this would have been possible without Dr. mercer's support.

I would also like to thank my committee members, Dr. Julia Freedman-Silvernail, Dr. James Navalta, and Dr. Jennifer Pharr. You have all taken the time to assist with any and everything that was needed to complete this project. My education and eventually my career will be heavily impacted by the support I received from this group, and I am continually grateful for this.

Finally, I would be remise if I did not thank my mother. Throughout my entire life my mom has supported me to get me to this stage, and she is truly the driving force behind the researcher and the man that I have become. Thank you for always being a rock for me.

Table of Contents

List of Figures

Chapter 1

Introduction

For decades, clinicians and researchers have sought out alternative rehabilitation methods for individuals suffering from clinical ailments, as well as recreational and professional athletes who are dealing with lower leg overuse injuries, or who simply want to find another training modality. In the most recent publication, the Centers for Disease Control and Prevention (CDC) estimated that nearly 800,000 people in the United States alone suffer a stroke (CDC 2015). It has been reported that among stroke patients, over 50% of the population that survives the acute stages will require a period of rehabilitation in order to regain locomotive abilities (Barbeau & Visintin 2003). Along those same lines, the National Institute of Health (NIH) reported that over 330,000 hip replacements are performed in the United States each year (NIH 2013). Additionally, taking into account that anywhere from 27-70% of recreational runners may experience an injury in a given year (Caspersen et al. 1984), and that the number of marathon finishers alone jumped to 541,000 in 2014 (Running USA 2014), it is easy to see just how many people are affected by some lower body ailment. Thus, it is imperative to find effective rehabilitation tools that can return people to functional levels in a timely manner.

A rehabilitation technique that has been hypothesized to be effective in treating these types of ailments is the use of body weight supported (BWS) locomotion (Finch, Barbeau, Bertrand 1991; Eastlack et al. 2005; Hesse at al. 2003). BWS locomotion was first introduced as a rehabilitation technique roughly thirty years ago (Finch & Barbeau 1985). Through out that time span, there have been many different tools and techniques used to support body weight. These mechanisms include crutches, walkers, parallel bars,

deep-water running, harnesses, and most recently, the lower-body positive pressure (LBPP) treadmill. Although the clinical application of body-weight supported locomotion has proved useful for rehabilitation (Grabowski 2010, Visintin et al. 1998), there are still some inherent problems with the different devices. Eastlack et al. (2005) laid out a few of the issues that are present with the use of these devices: crutches can increase oxygen consumption by up to 70% for a given speed due to the usage of the upper extremity, deep-water running cannot produce normal gait mechanics, and walkers, parallel bars, and harnesses all have the tendency to alter gait and joint mechanics. The introduction of the LBPP treadmill provided a means for utilization of BWS locomotion with theoretically minimal changes to gait kinematics.

The use of BWS locomotion in a rehabilitation setting has raised questions about its practicality, especially in relation to muscle activity and the cardiovascular fitness benefits of locomotion at varying levels of BWS. It has been reported that muscular activity of lower extremity muscles is decreased as body weight is reduced (Liebenberg et al. 2011; Mercer, Applequist, and Masumoto 2013). These reductions in muscle activity have been linear, however, the reduction in muscle activity does not occur at the same rate as the reduction in body weight (i.e. a reduction of 20% BW does not guarantee a 20% reduction in muscle activity). Similar results have been found for the metabolic cost of locomotion at varying levels of BWS: net metabolic rate decreased moderately, but in less than direct proportion to the decreases in body weight (Grabowski, Farley, & Kram 2005; Teunissen, Grabowski, & Kram 2007).

It is important to note that in these studies, all subjects were asked to run at a controlled speed, or multiple controlled speeds. As of now, there are no publications that

investigate how levels of muscle activity and metabolic cost are altered while running with BWS at a preferred running pace. This is important to understand because each person is different, and as a result each has the potential to run at a different pace. In order to fully understand the effect of body weight support and speed on these dependent variables, it is imperative that each individual is able to run in as close to normal conditions as possible. This includes allowing each individual to pick a speed that is as close to his or her preferred pace as possible.

Purpose of the Study

The primary purpose of this study is to determine if body weight support influences specific physiological and biomechanical parameters while running at a preferred pace. A secondary purpose was to determine if these parameters were influenced by the interaction of body weight and speed.

Chapter 2

Review of Literature

The primary purpose of this study is to determine if body weight support influences specific physiological and biomechanical parameters while running at a preferred pace. A secondary purpose was to determine if these parameters were influenced by the interaction of body weight and speed.

The review of the literature on this topic will first show how and why body weight supported locomotion has been used in a clinical rehabilitation setting. Following a review of the classical applications, this investigation will discuss injury occurrence during endurance running and how body weight supported locomotion may aid in rehabilitating from those injuries. From there, the document will examine how independently manipulating body weight support and speed can effect metabolic variables and lower limb muscle activity during running. To conclude, the final section will focus on the acceleration of the lower leg during running.

Rehabilitation

In 2015, the Centers for Disease Control and Prevention (CDC) reported that more than 795,000 people have a stroke each year, with just fewer than 130,000 of these people dying as a result (CDC, 2015). This leaves approximately 665,000 living stroke patients each year, and it has been reported that over 50% of this population will require at least some period of rehabilitation in order to regain locomotive abilities (Barbeau & Visintin, 2003). Another population of people that requires rehabilitation to regain locomotive abilities are those who undergo hip replacements. According to the National Institute of Health (NIH), there are over 330,000 hip replacements performed each year

in the United States (NIH, 2013). It is clear to see that many people rely on rehabilitative efforts to return some level of functioning, and body weight supported locomotion has been a major asset in the rehabilitative process.

Body weight supported locomotion was first introduced as a rehabilitation technique roughly thirty years ago (Finch & Barbeau, 1985). Early methods for providing body weight support included the use of crutches, walkers, parallel bars, aqua therapy, waist belts, or harness supported treadmill exercises. While all of these devices theoretically provided body weight support to the user, they also all compromised rehabilitation to some extent based on “logistic, medical, or safety issues” (Eastlack et al., 2005). Crutches, parallel bars, and walkers all alter normal gait mechanics, in addition to requiring considerable amounts of upper body strength. Patients who use crutches have been seen to increase oxygen consumption by up to 70% due in large part to the upper extremity involvement (Eastlack et al., 2005).

Additionally, these authors point out that while aqua therapy has the potential to reduce weight bearing loads, the drag forces created by the water have the potential to alter gait, muscle activation patterns, and coordination (Eastlack et al., 2005). A form of aqua therapy, deep water running, eliminates the foot impact with the ground entirely and is therefore believed to reduce injury (or re-injury) risk (Masumoto, Delion, Mercer, 2009). However it is important to note that there are “distinct kinematic differences” (Masumoto, Delion, Mercer, 2009) between the deep water running style and “normal” over ground locomotion, and as such the two forms of locomotion are not completely synonymous.

Many studies have used some variation of an overhead harness (Finch, Barbeau, Arsenault, 1991; Visintin et al, 1998; Teunissen, Farley, Kram, 2004), as a method of applying an assistive vertical force to the person. A problem associated with all types of harness systems is the tendency to cause discomfort and impede circulation, and as such may not be an adequate tool for long-term rehabilitation (Grabowski & Kram, 2008). A problem specific to using harnesses attached to a fixed point is the possibility of inadvertently providing assistive horizontal forces to the person while walking (Grabowski, Farley, Kram, 2005), which could further complicate the return to normal body weight locomotion. Utilizing an attachment to a rolling trolley that can move forwards and backwards as the person adjusts forwards and backwards on a treadmill can help this particular problem. A separate issue pertaining to the use of a spring-loaded harness system is the inability to maintain a constant lengthening (and therefore a constant force application) of the spring. This has the potential to alter the vertical forces acting on a runner on a stance phase by stance phase basis, and lead to inconsistent amounts of body weight being supported.

Despite the inherent problems associated with each method of providing body weight support, researchers and clinicians have reported that the use of assistive devices has led to better outcomes for patients than traditional rehabilitation measures (Barbeau & Visintin, 2003; Hesse et al., 2003; Visintin et al., 1998). For example, stroke patients who underwent rehabilitation programs that included body weight support were seen to have increased walking speed, walking endurance, functional balance, and lower-limb motor recovery compared with patients who rehabilitated by walking without body weight support (Barbeau & Visintin, 2003; Visintin et al 1998). Three-month follow up

tests indicated that the body weight support group maintained higher scores for all measures (Barbeau & Visintin, 2003).

Patients that have undergone a lower extremity surgery have also benefitted from the use of body weight supported locomotion during rehabilitation (Hesse et al., 2005; Eastlack et al., 2005). When examining the effectiveness of body weight support for rehabilitating hip replacements, patients who were in the body weight support group scored higher on the Harris Hip Score (a test used to determine a person's pain and functional ability in the time after a hip replacement) directly after, three months after, and twelve months after surgery (Hesse et al., 2003). Similar findings were seen in a group of patients who had undergone knee surgery, as they experienced pain reductions up to 80% while using lower body positive pressure exercise after surgery (Eastlack et al., 2005). The work completed by these researchers suggests that body weight supported rehabilitation is more successful than traditional rehabilitation in a clinical setting, and as such there may be applications for its use in recovery from endurance running and training injuries.

To address these limitations of traditional body weight support, a more recent technique to provide body weight support during locomotion has been the use of lower body positive pressure. Having the lower body in an enclosure within which the pressure is increased increases the pressure applied to the lower body. The increased pressure causes an upward directed force applied to the waist region causing body weight support.

Overuse Running Injuries and Rehabilitation

There has been a dramatic increase in the number of people becoming involved in recreational running over the last twenty years, as the sport has seen a 300% growth from

1990 to 2003 (Running USA, 2015). In 2014 alone over eighteen million people completed a U.S. running event, which included the 5K, 10K, half marathon, and full marathon (Running USA, 2015). The 25 to 44 year old age group accounted for half of all participants, which makes this group of particular interest to running researchers. Not included in this statistic is the total number of recreational runners who reported running at least fifty days per year, and that number reached nearly thirty million in 2013 (Running USA, 2014). Of course, an increase in the number of injured runners has accompanied the increase in participation.

Over the course of one year somewhere between 27-70% of recreational runners will experience some type of overuse injury (Liebenberg et al., 2011). That risk increases potentially up as high as 90% for individuals who are training for a marathon, with the highest risk seen in those individuals who are running over 40 km/week (Fredericson & Misra, 2007). There have been multiple hypotheses presented as to what could be the primary cause of lower leg overuse injuries during running. One such hypothesis is that overuse injuries occur due to excessive mechanical loading of the knee joint (Chumanov et al., 2012), as would be the case when this structure is exposed to a high number of repetitive forces. These forces could be individually smaller in magnitude than the acute injury threshold, yet produce a combined fatigue affect over long periods of time based on their cumulative loading effects (Hreljac, Marshall, & Hume, 2000). Hamill et al. (1983) measured forces from foot contact with the ground and determined each to have a magnitude somewhere in between two and four times a person's body weight. Considering data shows that there can be up to 2,500 collisions between the foot and the

ground during one thirty-minute run (Mercer et al., 2005) it would seem that these multiple impacts could play a role in overuse injury development.

There are several potential risk factors when talking about overuse running injury that go beyond the forces acting on the lower extremity. However it is thought that the ability to reduce reaction forces through supporting body weight can potentially reduce the joint forces acting on the lower limb, specifically at the knee joint (Patil et al., 2013). The intention of lower body positive pressure locomotion is to reduce forces acting on the body as a whole, and on the joints individually. As such this investigation will mainly address the ability to reduce the magnitude of the reaction force to potentially lower injury risk, knowing that there may be other associated risk factors.

Body Weight Support, Speed, and Ground Reaction Forces

Classical applications of body weight supported locomotion have been focused on reducing the overall ground reaction force, as well as lower limb joint forces, in order to decrease a patient's pain and return more quickly to functional locomotion (Hesse et al., 2003). Yet there has been much less research conducted to determine what supporting body weight does to other variables that play a role in gait. The effects that supporting body weight, speed, and ground reaction forces have on each other will be discussed here.

The relationship between body weight support and speed has remained unexamined for the most part, as previous studies have chosen to use a single control speed (Teunissen, Grabowski, & Kram, 2007; Chang & Kram, 1999) or a range of control speeds (Grabowski & Kram, 2008; Grabowski, 2010) when utilizing body weight support. From a clinical rehabilitation aspect, it has been reported that stroke patients

who rehabilitate at speeds that are more similar to a normal walking pace are more effective in improving self-selected over ground walking velocity than training at speeds at or below a patient's typical walking speed (Sullivan, Knowlton, & Dobkin, 2002). Investigations where one patient group was subjected to body weight supported rehabilitation showed similar findings in that the body weight support group was able to walk at faster speeds compared with the "normal" rehabilitation group after 6 weeks (Eastlack et al., 2005), as well as after a 3-month follow up (Visintin et al., 1998). In work done on healthy subjects, researchers have reported that as body weight support is increased, the preferred speed chosen by subjects increases as well (Masumoto et al., abstract). Specifically, these researchers have reported that preferred speed while running with no body weight support was slower than the preferred speeds at all levels of body weight support. Interestingly, the relationship between preferred speed and body weight support was linear but not directly proportional.

Both body weight support and speed of movement affect ground reaction force production during locomotion (Grabowski & Kram, 2008; Hamill et al, 1983; Mercer et al., 2005). In theory, while supporting a person's body weight, the force that they put onto the ground will decrease, and therefore decrease the force that is acting back on them. Grabowski (2010) demonstrated this practically, reporting that both the impact and active peak vertical GRF were decreased as more body weight was supported. Other investigators have reported that the vertical impact peak GRF and active peak GRF decrease linearly as body weight support was increased (Grabowski & Kram, 2008).

Speed changes can also affect the magnitude of the GRF's. In general, as velocity of locomotion increases, peak vertical (Grabowski, 2010) and horizontal ground reaction

forces increase as well (Hamill et al., 1983). Similarly, vertical impact peak GRF's (Mercer et al., 2005) and active peak GRF's have been reported to increase linearly with increases in speed (Grabowski & Kram, 2008). It is understandable that as more body weight is supported, as well as when speed is decreased, the magnitude of both the horizontal and vertical GRF's would decrease. However it may be that supporting body weight plays a larger role in GRF magnitude than speed manipulation does, as walking faster with more body weight support resulted in reduced peak vertical GRF compared with walking with less body weight support (Grabowski, 2010). If other variables related to gait could remain unchanged while providing body weight support and increasing speed of movement, this may have implications for future rehabilitation.

Electromyography During Locomotion

Lower extremity muscle activity plays a large role in locomotion, and both the magnitude and pattern of activity can be affected by multiple different factors including stride frequency and speed (Hof et al., 2002; Ivanenko, Poppele, & Lacquaniti, 2004; Chumanov et al., 2012). Increases in stride frequency influences muscle activation during locomotion primarily in the late swing phase, potentially in pre-activation anticipation mode for the impending foot-ground contact (Chumanov et al., 2012). A limitation of this work is that speed was controlled. To address this limitation, the proposed study will allow participants to run at a preferred stride frequency during all trials. This approach will allow for determining if changes in muscle activation are attributable to the varying levels of locomotion speed in each condition.

During walking at varying levels of speeds, Hof et al. (2002) demonstrated that there was a linear relationship between increases in speed and magnitude of muscle

activity in fourteen lower extremity muscles. Similar findings have been reported in other walking studies: for example. Capellini et al. (2006) saw an increase in average muscle activity accompany increases in speed for both proximal and distal lower extremity muscles. Additionally it was reported that patterns of activity in both groups of muscles were often different at different speeds. This is in agreement with Ivanenko, Poppele, and Lacquaniti (2004) who reported that a given muscle would express different patterns based on the stepping velocity of the subject. Going one step further Hof et al. (2002) stated that average EMG profiles of leg muscles could be predicted based on a number of speed dependent prediction equations. The existence of these basic normal EMG profiles during walking at given speeds could have implications in identifying normal versus abnormal gait patterns in clinical settings (Hof et al., 2002). If EMG profiles during walking are truly a function of magnitude of speed, than the same could potentially hold true for EMG profiles during running.

Muscles of the lower extremity muscle groups tend to act similarly in pattern, but with some changes in magnitude during running compared with activity during walking. Gazendam and Hof (2007) compared muscle activity during both running and walking over a range of speeds, and the pair found many similarities between the two modes of locomotion. Muscles from the gluteus, quadriceps, and hip flexor groups all showed nearly identical patterns, with the gluteus and quadriceps muscles experiencing larger magnitudes, and the hip flexor groups experiencing smaller magnitudes compared with walking. The hamstrings group and Tibialis anterior both show similar, although slightly altered, profiles compared with walking. These findings were corroborated by Capellini et al. (2006) when it was shown that speed changes were associated mainly with the

intensity, and only slightly with the timing, of the muscle activation. These differences in muscle activation in walking and running were found to be occurring only during the stance phase, as the activation patterns during swing phase were generally the same. Looking at purely running, Kyrolainen, Komi, & Belli (1999) showed that the magnitude of muscle activity increased with increases in speed in the gluteus maximus, vastus lateralis, biceps femoris, and gastrocnemius muscles. The ability to manipulate speed to elicit different magnitudes of muscle activity should continue to be a factor when looking at body-weight supported locomotion.

Considerably less is known about the activity of lower extremity muscles during locomotion with partially supported body weight. During walking trials at 0%, 20%, and 40% body weight support, Colby, Kirkendall, & Bruzga (1999) reported no differences in muscle activity of the hamstring or gastrocnemius groups, and only saw differences in the quadriceps group at 40% body weight support (27.8% less than 0% BWS). It is noted that although these results are non-significant muscle activity tended to decrease with body weight support, and as such could be significantly lower at higher levels of support. Klarner et al. (2010) ran a similar design with trials occurring every 20% reduction in support, but opted to test all the way to full support. Despite testing the additional support levels the results remained the same, although the average EMG did still tend to decrease with body weight support.

When examining running with body weight support, there appears to be a clearer pattern of how muscle activity is influenced by body weight support. Comparing 0% support with 90%, 80%, 70%, and 60% body weight support Liebenberg et al. (2011) reported average EMG as well as root mean square EMG to be significantly lower for the

biceps femoris, rectus femoris, gastrocnemius, and tibialis anterior. The decreases in activity seen in each of the muscles were linear, however not in direct proportion to the increases in body weight support. This finding was echoed in another study that tested activity levels in the same muscles between 50%-80% of body weight support. Mercer et al. (2013) showed that these muscles were less active by 36%, 43%, 51%, and 52% when running with 50%, 60%, 70%, and 80% body weight support, respectively. The activity levels for the biceps femoris in this investigation were not significantly different across support conditions, the lone difference in findings between the two investigations. This may be due to the existence of a “ceiling effect” in muscle activity as there would still be some activation simply from moving the legs in a running motion with 100% body weight support (Mercer et al. 2013). A close comparison to full support is deep water running, however results indicate that only the tibialis anterior and gastrocnemius activity was greater in treadmill running compared with deep water running, while the rectus femoris and biceps femoris were not different (Masumoto, DeLion, & Mercer 2009). Apart from the magnitudes of activity being different, the patterns of activity appeared to be different for at least the tibialis anterior and gastrocnemius between the two modes of locomotion. This could be attributed to the differences in kinematics between the two styles (Masumoto, DeLion, & Mercer 2009).

Findings related to lower extremity muscle activation during locomotion have been in general agreement, especially during running. EMG activity is shown to increase with increases in speed, and shown to decrease with increases in body weight support Hof et al., 2002; Kyrolainen, Komi, & Belli, 1999; Liebenberg et al., 2011). This may

suggest an ability to use increased running velocities with increased levels of body weight support to attain similar muscle function.

Metabolic Activity During Body Weight Supported Locomotion

Both speed and effective body weight changes play a role in determining the metabolic cost of locomotion while utilizing body weight support. Studies that have investigated this phenomenon have utilized either a lower body positive pressure (LBPP) treadmill or a harness system in conjunction with a normal treadmill (Farley & McMahon, 1992; Chang & Kram, 1999; Grabowski & Kram, 2008; Hoffman & Donaghe, 2011). The use of an LBPP treadmill or harness system can be attributed to the point made earlier about the increase in VO_2 while using devices such as crutches and walkers. Research in this area first focused on the effects supporting body weight had on the metabolic cost of locomotion (Farley & McMahon, 1992), and has progressed into determining the cost of generating force to support body weight as well as the cost of performing work to move the center of mass forward (Chang & Kram, 1999;Grabowski, Farley, & Kram, 2005).

As body weight support is increased during locomotion, the net metabolic rate decreases as well (Grabowski & Kram, 2008; Farley & McMahon, 1992; Teunissen, Grabowski, & Kram, 2007). The decrease seen in metabolic cost requirement is linear to the corresponding reductions in effective body weight, but there has been some debate as to whether this decrease is in direct proportion or not. For example, Farley and McMahon (1992) reported that at 25% and 50% body weight support, the metabolic cost of running decreased by 25% and 50% as well. At 75% body weight support, their subjects saw a 72% reduction in metabolic cost. These values are much closer to showing decreases in

direct proportion compared with other studies. Teunissen, Grabowski, & Kram (2007) looked at the same levels of body weight support utilizing both “rolling trolley” and “fixed pulley” methods of providing body weight support. In this investigation metabolic cost reductions were shown to be 19%, 38%, and 55% for the rolling trolley trials, and 19%, 43%, and 59% for the fixed pulley trials. Values found by Grabowski (2010) were also shown to decrease in less than direct proportion, as the percentage of net metabolic power was only decreased by 45% at 75% body weight support.

Apart from determining the overall effect of body weight support on metabolic cost, researchers have become increasingly interested in determining how much of this cost is going towards supporting body weight, and how much is being used to accelerate mass forward. An estimated 28% of net metabolic cost of normal walking is used to support normal body weight (Grabowski, Farley, & Kram, 2005). This estimate is much lower than for normal running, as metabolic cost of supporting body weight can comprise up to 74% of the total net metabolic cost during running (Teunissen, Grabowski, & Kram, 2007). The drastic change in metabolic cost between walking and running can be attributed in at least some part to speed increases, and the influence of speed on metabolic cost of locomotion will be discussed. There is substantial metabolic cost to accelerating the body forward as well, with one group estimating that up to 45% of the net metabolic cost of walking (Grabowski, Farley, & Kram, 2005).

Chang & Kram (1999) showed that a 6% of total body weight resistive horizontal force resulted in a 30.2% increase in average metabolic rate compared with a non-resistive force trial. A 6% of total body weight assistive horizontal force resulted in a

22.8% decrease in average metabolic cost, while a 15% of total body weight assistive force resulted in a 32.5% decrease in average metabolic cost.

The speed of movement plays a large role in the metabolic cost of locomotion (Farley & McMahon, 1992; Grabowski & Kram, 2008; Teunissen, Grabowski, & Kram, 2007). An important note from the methodology of each of these studies is the use of a control speed, or a range of control speeds, for all of their subjects. Controlling speed may unintentionally constrain running characteristics, which could have the potential to change the metabolic cost required to complete the task.

The inherent problems with traditional body weight support devices have been discussed previously, and how to provide body weight support has remained a methodological question. With the introduction of the lower-body positive pressure treadmill in 2005 (Alter-G), researchers and clinicians have been able to provide body weight support without causing discomfort or altering running kinematics. Measurements taken while subjects were standing in an LBPP device show no changes in diastolic blood pressure at either level, an increase in systolic blood pressure, and a decrease in heart rate only at 50%, but not at 25% body weight support compared with a standing trial with 0% body weight support (Hoffman & Donaghe, 2011). This demonstrates that physiological changes that occur during less than 50% body weight supported exercise occur as a result of the exercise, and not due to confounding factors from the device itself. Additionally, it has been reported that LBPP exercise has no effect on blood pressure, and only decreases heart rate when running at levels equal to or greater than 70% of body weight support (Hargens et al., 1999). One unintended circumstance of using the LBPP treadmill is that participants are able to lean back into the device and receive an assistive force while

using it (Grabowski, 2010); however, this is negligible when compared with the issues surrounding use of other supportive devices.

Tibial Acceleration and Shock Attenuation

To date there has been a minimal amount of published literature that has investigated tibial acceleration levels at different levels of body weight support. Information about the magnitude of tibial acceleration and shock attenuation experienced during normal over ground or treadmill running, in conjunction with what little is known during body weight supported running, should provide insight enough to produce a hypothesis regarding how this measure will be affected by fluctuations in speed and body weight support.

Each time a person contacts the ground with their foot a shock wave is transmitted from the foot, through the body, and eventually reaches the head (Hamill, Derrick, & Holt, 1995). The ability of the body to reduce the magnitude of this shock as it travels from the foot to the head has been deemed shock attenuation. The body relies on several different mechanisms to help attenuate these forces, including bone, cartilage, synovial fluids, soft tissues, joint kinematics, and muscular activity (Lafortune, Lake, & Hennig, 1996). Of all these mechanisms, the musculo-skeletal system is the best equipped to actively attenuate shock based on its ability to adjust joint stiffness and joint kinematics to place segments in positions that can more easily attenuate shock (Hammill, Derrick, & Holt, 1995).

Research looking into shock attenuation has focused on how leg and head acceleration are affected by three main variables: speed, stride length, and fatigue. It has been documented that speed plays an important role in the acceleration at the tibia as well

as how much this is attenuated moving up the body (Mercer & Chona, 2015; Mercer et al., 2005). When measuring changes in these variables while running at speeds from 50% to 100% of total maximum speed, Mercer et al. (2002) reported roughly a 60% increase in shock attenuation, which was an approximate attenuation increase of 20% per every 1 m/s in speed increase. Additionally leg acceleration increased 4.8 gravity units (g's) in magnitude, which translated to a 42% increase in acceleration per each 1 m/s increase in speed. Although the head acceleration was reported to increase with running speeds as well, the increase of 0.5 g's was relatively small compared with the acceleration seen in the leg. This may be due to the fact that the head needs to maintain constant acceleration levels in order to keep a stable visual field during running (Hamill et al., 1995). Other studies have found similar results, with Mercer et al. (2005) reporting leg impact acceleration increases of 24% per every 1 m/s increase. This increase in leg acceleration, rather than an increase in head acceleration, is the mechanism behind the increase in impact attenuation (Mercer et al., 2005).

An interesting caveat from these studies that have manipulated speed is the fact that they allow participants to run at one (preferred) stride length. While controlling for speed and measuring at ±10% and ±20% of preferred stride length Derrick, Hammill, & Caldwell (1998) reported a decrease in tibial acceleration (5.6 g's) as well as head acceleration (0.8 g's) as stride length was reduced. This resulted in an overall decrease in attenuation of 7.7 decibels between the longest and shortest stride frequencies. A similarly designed study used frequency analysis to determine a power spectral density (PSD) for leg and head accelerations. Hamill, Derrick, & Holt (1995) reported differences in PSD for leg acceleration only between the +20% and -20% PSF condition,

while finding no differences in head acceleration PSD. It is important to note that although increases in both speed and stride length have yielded increases in leg accelerations, a linear relationship has been shown between speed and stride length, where increasing one value increases the other (Mercer et al., 2005; Mercer et al., 2002). If runners chose gait patterns based on minimizing impact force or attenuation, manipulating either speed or stride length would produce a parabolic relationship in which preferred speed/stride length would show the lowest levels of leg acceleration. As this is not the case, it is not clear which of the two factors plays the bigger role in leg acceleration and shock attenuation.

The role of fatigue in leg acceleration and the body's ability to attenuate shock cannot be overlooked. Verbitsky et al. (1998) used a group of non-fatigued runners to control against a group of fatigued runners and was able to report an increase of nearly 9 g's in acceleration of the leg between the first and thirtieth minutes of the protocol in the fatigued group. There were no differences reported in the non-fatigued group for tibial acceleration magnitude across time. This idea was somewhat corroborated while testing subjects as their own control in a pre and post-fatigue study. Mercer et al. (2003) reported an average decrease in attenuation of roughly 12% for subjects after the fatigue protocol, yet reported no differences in PSD of the leg or the head during the runs. This was attributed to the combined changes of PSD in both the leg and head, with neither value actually showing statistical significance between the pre and post trials (Mercer et al., 2003).

The effect of body weight support during running on leg acceleration has yet to be determined. The one study that has looked into this was Mercer & Chona (2015) who had

participants run with 0%, 60%, 70%, and 80% body weight support, while running at 100%, 110%, 120%, and 130% of preferred speed. These researchers reported a difference in leg acceleration with no body weight support compared with the 60%, 70%, and 80% conditions, but did not see differences between body weight support conditions. Additionally it was reported that regardless of body weight support, leg impact acceleration increased as velocity increased (Mercer & Chona, 2015).

Based on what is reported in the current published literature, it would appear that there is a linear relationship between speed and leg acceleration during running (Mercer et al., 2002; Mercer & Chona, 2015). However, what is not clear is how body weight support may affect the magnitude of tibial acceleration. The non-significant differences reported by Mercer & Chona (2015) may be attributable to the large amount of body weight support provided. It remains to be seen how leg acceleration is affected with only 10% and 20% of body weight being supported.

Summary

Published literature that has investigated the effects of supporting body weight on gait and recovery in a clinical setting has been overall positive regarding its effectiveness in rehabilitating patients. More recently, researchers have looked to explore how supporting body weight affects variables such as ground reaction force, metabolic cost, muscle activity, and leg acceleration in order to determine if the use of body weight supported exercise could be used during training or for rehabilitation from an acute injury. Researchers investigating running have reported that muscle activity of key lower extremity muscles is increased as running velocity is increased and as body weight support is decreased. Yet the entirety of this literature has focused on testing individuals

at a control speed or a range of control speeds, which may inherently affect gait kinematics. In studies that have looked at ranges of control speeds it has been shown that speed affects both the muscle activation and metabolic cost of running (Mercer et al., 2013; Grabowski, 2010; Grabowski & Kram, 2008) occurring during both normal locomotion and body-weight supported locomotion. Furthermore, few researchers have investigated the effects that supporting body weight can have on the acceleration of the lower leg, which may have implications for injury and rehabilitation. Therefore, it is the attempt of the current study to determine how these variables are affected while running at variations of preferred speed and increasing levels of body weight support.

Chapter 3

Methodology

Participants

A total of 9 healthy participants were recruited to participate in this investigation (age: 28.56 ± 7.88 years, height: 1.68 ± 0.08 cm, mass: 65.70 ± 7.64 kg). In order to be included as a subject for this study, participants needed to be over the age of 18, free of any injury that would interfere with the ability to run, and not pregnant. Additionally, subjects were only recruited if they were deemed to be recreationally active runners (running at least 10 miles/week). The average amount of mileage ran per week by all subjects was 14.67 ± 4.92 miles. This minimum criterion was set to ensure that participants were able to complete the protocol without becoming overly fatigued. IRB approval was obtained prior to the beginning of this investigation, and all participants were required to sign an informed consent form acknowledging their participation in this study.

Instruments

In order to provide body weight support (BWS), a lower body positive pressure treadmill was utilized (Alter-gravity P200 treadmill, Fremont CA). A one-dimensional accelerometer (PCB Piezotronics 352C68, Depew NY) was attached to the anterior, distal portion of each subject's shank. Muscle activity data were recorded using an electromyography EMG system (Delsys Trigno Wireless EMG, Natick, MA). Skin preparation and positioning were done in accordance with SENIAM (surface electromyography for the non-invasive assessment of muscles) recommendations. In order to record metabolic data, subjects were instrumented with a heart rate transmitter

(Polar T31, Lake Success, NY) as well as a portable pulmonary gas exchange measurement system (Cosmed K4B2, Chicago, IL).

Procedures

Participants were asked to come into the laboratory on one day for testing. After providing informed written consent, height, mass, age, and running mileage/week were recorded. Once anthropometric measurements were recorded, subjects were asked to put on the Alter-g shorts and step into the Alter-g treadmill. Participants were allowed to go through a three-minute warm-up, spending one minute running at each body weight support level utilized in this protocol. Each subject was asked to warm-up at different speeds at the varying levels of body weight support, so that he or she was able to become comfortable running with body weight support in the Alter-G.

Figure 1: Alter-g custom made shorts for use with the LBPP treadmill.

Upon completion of the warm-up, each participant was asked to determine a preferred speed at each level of body weight support. In total, there was three body weight support conditions: 0%, 10%, and 20% of total BW was supported. All

participants determined preferred speed at each level, starting with the 0% BWS condition before moving to the 10% and 20% BWS conditions. Preferred speed was explained as the speed that he or she might choose when going out for a thirty-minute training run. The actual speed display was blinded to each participant in order to avoid targeting of any speed. Each participant started the treadmill and was able to manipulate the speed until he or she reached what would be considered the preferred speed. Once preferred speed was reached, the treadmill was stopped, allowed to come to a full stop, and then a second preferred speed measurement was taken. A total of three trials at each BWS condition were performed, and an average of these three trials was used as the preferred speed of the participant at each BWS condition. Additionally, the researcher calculated a +10% and -10% preferred speed value for each subject at each level of effective BW, with these values rounded to the nearest tenth of miles/hour.

Following preferred speed measurements, each participant exited the LBPP treadmill and was instrumented with four Delsys EMG leads, a one-dimensional PCB Piezotronics accelerometer, a Polar heart rate transmitter, and the Cosmed K4B2 portable pulmonary gas exchange measurement system. The muscles of interest for this investigation were the Rectus Femoris, Semitendinosus, Tibialis Anterior, and Medial Gastrocnemius. The accelerometer was placed on the anterior, distal portion of the shank and was used to determine average tibial acceleration values across five peaks during each condition. After EMG and accelerometer instrumentation was complete each participant was instructed to re-enter the Alter-g treadmill, at which point he or she was instrumented with the mask for the portable pulmonary gas exchange measurement system, as well as the heart rate transmitter

As part of the actual testing protocol, there were three speed conditions (preferred speed, +10%, and -10% of preferred speed) as well as three body-weight support conditions (0%, 10%, 20%). This was a total of 9 conditions that each subject was asked to complete. Each condition was timed so that participants ran at each condition for seven minutes, with a maximum of three minutes of rest in between each trial. Subjects were allowed to rest for shorter if they so chose. To avoid any order effect, the order of the BWS conditions was counterbalanced so that half of the participants (odd numbered) began at 0% support and worked downwards, while the other half (even numbered) began at 20% support and worked upwards. However, the order of the speed conditions was always fastest (+10%) to slowest (-10%). The breakdown for the conditions for each group is as follows:

Table 1: Summary of Experimental Trials

Odd Subjects			Even Subjects		
Condition Order	% BWS	Speed	Condition Order	% BWS	Speed
1	0%	(+) 10%	1	20%	(+) 10%
2	0%	Preferred	2	20%	Preferred
3	0%	(-) 10%	3	20%	(-) 10%
4	10%	(+) 10%	4	10%	(+) 10%
5	10%	Preferred	5	10%	Preferred
6	10%	(-) 10%	6	10%	(-) 10%
7	20%	(+) 10%	7	0%	(+) 10%
8	20%	Preferred	8	0%	Preferred
9	20%	(-) 10%	9	0%	(-) 10%

Data Reduction

VO_2, VCO_2, and HR data were measured during minutes 5 through 7 of the 7-minute trial. VO_2 and VCO_2 were then averaged across the three-minute time period for each trial. Muscle activity of the Rectus Femoris, Semitendinosus, Medial Gastrocnemius, and Tibialis Anterior was recorded for 30 seconds during minute one of

the trial, as well as for 30 seconds during minute six of the trial. The DC bias was removed from the EMG data before undergoing a full-wave rectification. From there, average EMG values during 20 seconds of the 30-second period were calculated for each muscle group. These average EMG values were normalized to a standard running condition (0% body weight support, running at preferred speed), and percent differences away from the normalization trial were used for analysis. Acceleration data were collected during minute two of each condition. A custom MATLAB program (Version R2016a; Mathworks, Natick, MA) was written to identify five consecutive leg impact peak accelerations per condition, and the average magnitude of these five leg impacts was used for analysis.

Data Analysis

The independent variables for this study were body weight percentage and speed. The dependent variables in this study were VO_2, VCO_2, tibial acceleration, and EMG of the four muscles previously discussed. For each dependent variable, a three (BWS condition) by three (speed condition) repeated measures factorial ANOVA with Bonferroni adjustments for multiple comparisons was used to determine statistical significance, with the alpha level set to $\alpha=0.05$.

Chapter 4

Results

EMG of the rectus femoris, semitendinosus, tibialis anterior, and gastrocnemius was not influenced by an interaction of body weight support and speed ($p>0.05$ for all muscles). Furthermore, EMG of the four muscles was not influenced independently by the body weight support ($p>0.05$; Table 2) or speed ($p>0.05$; Table 3).

Table 2: Average EMG of the four lower extremity muscles for each body weight support condition. No differences reported for any muscle across body weight support conditions ($p>0.05$). Values represent percent difference mean ± standard error away from the normalization condition.

Body Weight Support Condition	Rectus Femoris	Semitendinosus	Tibialis Anterior	Medial Gastrocnemius
0%	-16.11 ± 4.93	2.31 ± 5.29	11.52 ± 3.23	4.46 ± 2.17
10%	45.05 ± 34.03	1.19 ± 8.59	16.94 ± 13.20	-5.33 ± 5.69
20%	27.76 ± 44.24	-2.94 ± 14.93	2.45 ± 13.14	2.78 ± 14.03

Table 3: Average EMG of the four lower extremity muscles for each speed condition. No differences reported for any muscle across speed conditions ($p>0.05$). Values represent percent difference mean ± standard error away from the normalization condition.

Speed Condition	Rectus Femoris	Semitendinosus	Tibialis Anterior	Medial Gastrocnemius
+ 10% of PS	7.29 ± 23.72	16.69 ± 15.46	14.11 ± 11.72	5.65 ± 7.01
PS	38.01 ± 27.61	-4.40 ± 6.43	9.17 ± 7.48	0.91 ± 4.83
- 10% of PS	11.41 ± 28.52	-11.75 ± 6.99	7.643 ± 9.28	-4.65 ± 6.19

VO_2 was not influenced by an interaction of speed and body weight support ($p>0.05$). VO_2 was influenced by body weight support ($p<0.001$; Figure 2). Post hoc test determined that VO_2 during running at 0% body weight support was significantly higher than VO_2 at both 10% body weight support ($p<0.05$) and 20% body weight support ($p<0.05$).

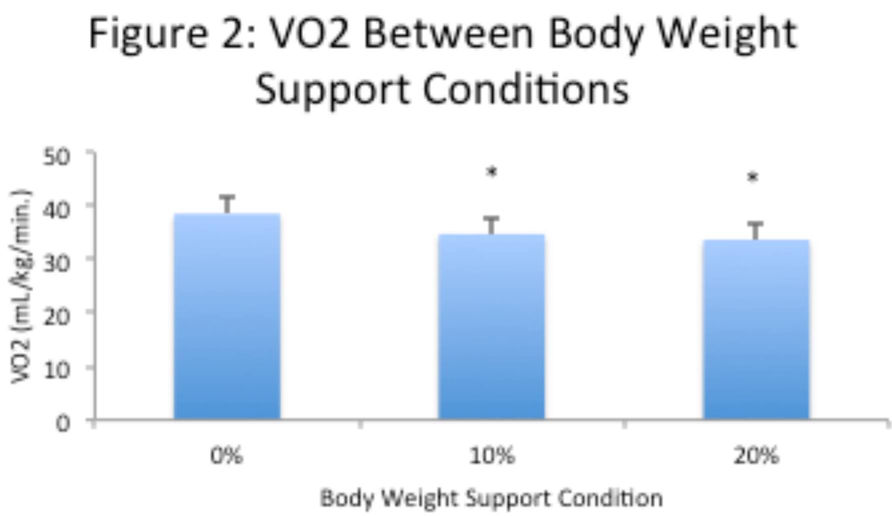

Figure 2: Average VO2 values separated by body weight support. Values for VO2 are given in units of mL/kg/min. * Indicates significant differences from 0% support condition (p<0.05).

VO_2 was influenced by speed ($p<0.001$; Figure 3). Specifically, VO_2 was significantly higher during running at the +10% of preferred speed condition compared with both the preferred speed condition ($p<0.05$) and the –10% of preferred speed condition ($p<0.05$). Additionally, VO_2 was significantly higher in the PS condition compared with the –10% of preferred speed condition ($p<0.05$)

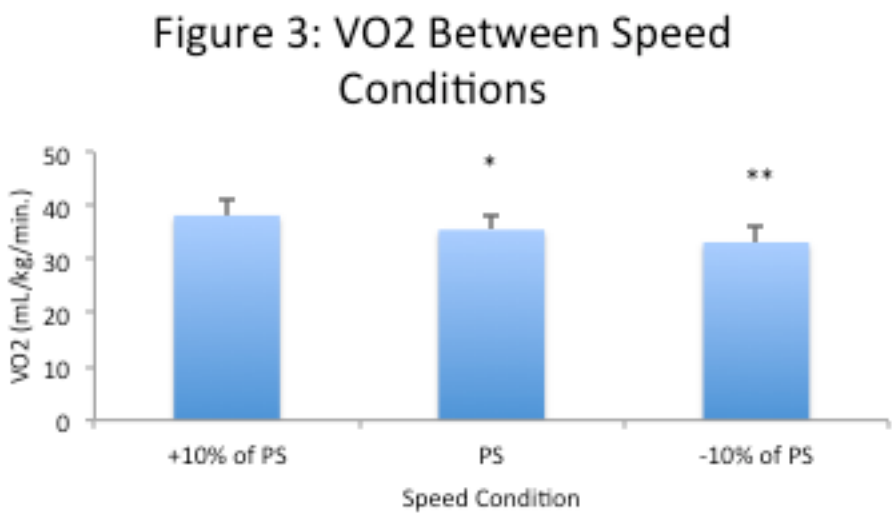

Figure 3: Average VO2 values separated by speed. Values for VO2 are given in units of mL/kg/min. * Indicates significant difference from +10% of PS condition ($p<0.05$). ** Indicates significant difference from PS condition ($p<0.05$).

VCO2 was not influenced by the interaction of speed and body weight support ($p>0.05$). VCO_2 was not influenced by the body weight support ($p>0.05$; Figure 4).

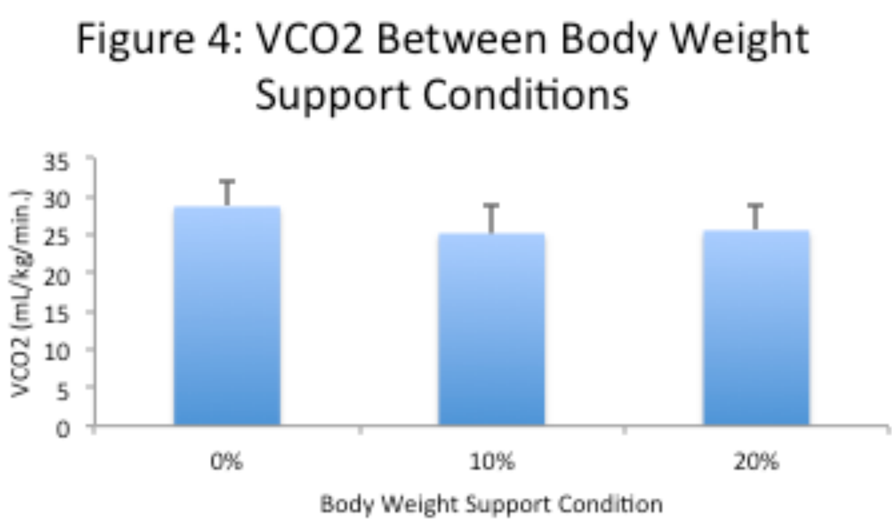

Figure 4: Average VCO2 values separated by body weight support. Values for VCO2 are given in units of mL/kg/min. No significant differences in VCO2 across body weight support conditions ($p>0.05$)

VCO_2 was influenced by the speed ($p<0.05$; Figure 5). Specifically, VCO_2 was lower for the –10% of preferred speed condition compared with both the +10% of preferred speed condition ($p<0.05$) and the preferred speed condition ($p<0.05$).

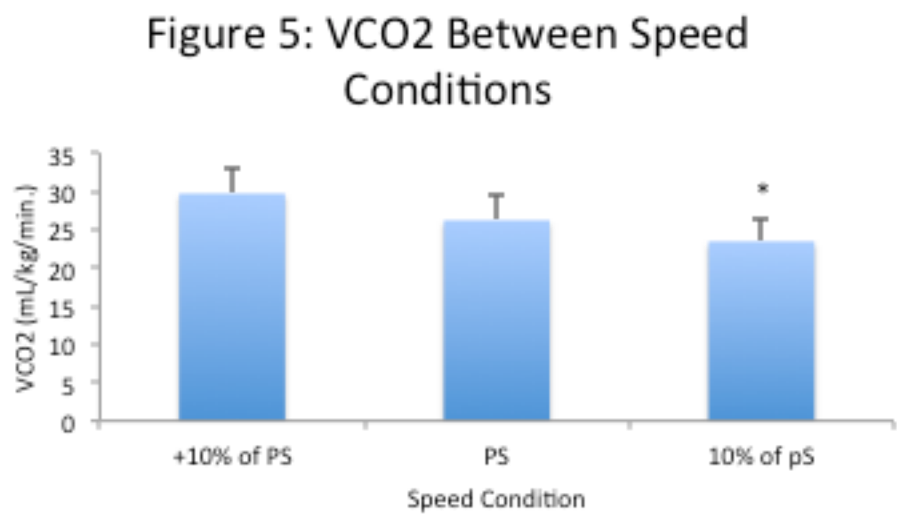

Figure 5: Average VCO2 values separated by speed. Values for VCO2 are given in units of mL/kg/min. * Indicates significant differences from both the +10% of PS and PS conditions ($p<0.05$).

Tibial acceleration was not influenced by an interaction of body weight support and speed ($p>0.05$). Tibial acceleration was not influenced by the body weight support ($p>0.05$; Figure 6) but was influenced by speed ($p<0.05$; Figure 7). Specifically, tibial acceleration was significantly higher during the +10% of preferred speed condition compared with both the preferred speed condition ($p<0.05$) and the -10% of preferred speed condition ($p<0.05$). Figures 6 and 7 demonstrate tibial acceleration values when grouping conditions by body weight support, as well as when grouping conditions by speed.

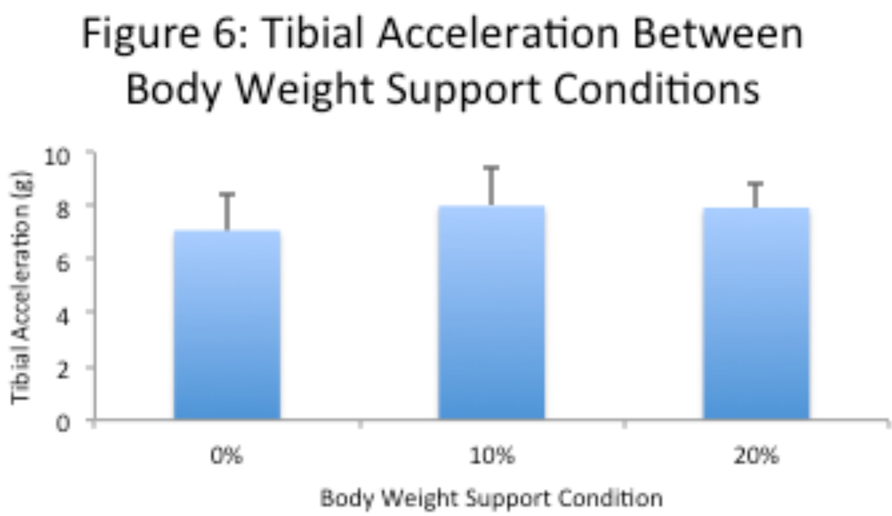

Figure 6: Average tibial acceleration values separated by body weight support. Acceleration magnitudes shown in units of gravity. No significant differences in tibial acceleration across body weight support conditions ($p>0.05$).

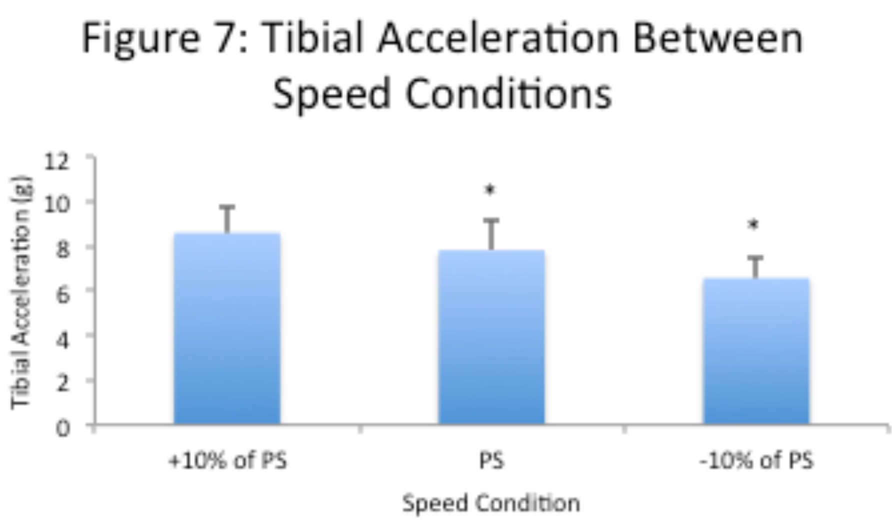

Figure 7: Average tibial acceleration values separated by speed. Acceleration magnitudes shown in units of gravity. * Indicates significant differences from +10% of PS condition ($p<0.05$).

Chapter 5

Discussion

The most important observation of this experiment was that average muscle activity was unchanged despite changes in body weight support and speed. Yet when examining the average oxygen intake there was a significant decrease for both the 10% and 20% body weight support conditions compared with the 0% condition. It was hypothesized that there would be significant interactions between body weight support and speed for all dependent variables. This hypothesis was refuted as a result of these findings. Interestingly, the observation that there was a decrease in oxygen consumption as body weight support increased and speed decreased without a concurrent change in muscular activity of the lower extremity drives the need for more work in this area.

The observation of no change in muscle activity across body weight support conditions is in agreement with studies that have looked at body weight supported walking (Colby, Kirkendall, & Bruzga, 1999; Eastlack et al., 2005). Colby, Kirkendall, & Bruzga (1999) reported no differences in average muscular activity of the quadriceps, hamstrings, or gastrocnemius muscles at 20% body weight support. However, it was reported that there was a significant difference for the quadriceps muscle activity while walking with 40% body weight support. It may be that there is a threshold of body weight support that illicit gait changes that can be detected by muscle activity changes.

Interestingly, the muscle activity response during walking with body weight support seems different than during running. Liebenberg et al. (2011) reported significant decreases in muscular activity as body weight support was increased from 0% to 40% support in 10% increments. Additionally it was reported that muscle activity

increased as speeds increased. In another study, Mercer et al. (2013) reported decreases in average muscle activity for the rectus femoris, tibialis anterior, and medial gastrocnemius at levels of 50%-80% body weight support as compared to the 0% body weight support condition. The biceps femoris muscle was reportedly not influenced by body weight support. All four muscles were influenced by speed.

In the present study, muscle activity did not differ between no body weight support and 10% and 20% of body weight support. This seems to be in disagreement with previous research (Liebenberg et al., 2011; Mercer et al., 2013). However, the disagreement may be due to the magnitude of body weight support tested. Mercer et al. (2013) utilized body weight support levels that were much higher compared to the levels used here, and as such may have seen different outcomes based on higher levels of support. Liebenberg et al. (2011) did not compare muscle activity between the 0% support condition and other support conditions, and as such the question remains about how the body (in particular here, the muscle function) reacts to having 10% of body weight supported. It may be that muscle activity changes over a large range of body weight support, but smaller incremental changes in body weight support do not have a measurable influence on muscle activity.

The difference between studies may also be related to the speeds tested. As part of the two study designs utilized by Mercer et al. (2013) and Liebenberg et al. (2011), participants ran at preferred speed as well as +15% and +25% of preferred speed. It is possible that the magnitude of these running speeds being above preferred speed could elicit higher muscular activity responses, rather than simply a range of speeds. In the present study, participants were tested at speeds 10% higher than preferred speed, at

preferred speed, and 10% lower than preferred speed. The 10% increase performed in the current investigation may simply not have been a high enough magnitude increase to see the same responses.

The different observations between studies may also be related to the type of subject recruited for the different studies. Mercer et al. (2013) reported average running mileage of approximately 4.6 miles/week, while the current investigation reported an average of 14.7 miles/week. The difference in mileage/week may indicate the current study having more trained runners, who may be more prepared to adjust with external perturbations. It is presently not known how running experience influences response to running with body weight support.

The metabolic data presented in this investigation are in agreement with previous literature, which have reported linear decreases in metabolic cost with body weight support increases (Farley & McMahon, 1992; Teunissen, Grabowski, & Kram, 2007; Grabowski & Kram, 2008). It is important to note that although these studies all reported linear decreases, the ratio of metabolic cost decrease to body weight support increase has not maintained a one-to-one ratio. These observations have ranged in their percentage of oxygen cost reduction for a given level of body weight support. Farley and McMahon (1992) reported a nearly one-to-one reduction in net oxygen cost of 28% at 25% body weight support, while Grabowski (2010) reported only a 45% decrease at the same support level). The current study reported reductions in oxygen cost of 9.8% and 12.6% when comparing support levels of 10% and 20% support to the 0% support condition, respectively.

Tibial acceleration data presented here is in general agreement with previous research (Mercer et al., 2002; Mercer et al., 2005; Hamill et al., 1995) when comparing for different speeds, but not necessarily during body weight support conditions (Mercer & Chona, 2015). Much of the literature already published has investigated the effect of speed on leg acceleration without body weight support. When examining this phenomenon, Mercer at al. (2002) reported increases in tibial acceleration as speed increased, where there was roughly a 42% increase in acceleration for every 1 m/s increase in velocity. The current results indicate an increase in tibial acceleration with running speed; however, the increase was observed to only be 21% for every 1 m/s increase in running speed. This is more in line with Mercer et al. (2005) who reported increases in leg acceleration of 24% for every 1 m/s increase in speed. The discrepancy in reported values of the two studies for percent increases in tibial acceleration with increases in running velocity may be attributable to the methodology behind determining levels of running velocity. Mercer et al. (2002) determined a maximal running velocity for each participant and used 10% increments to reduce the running speeds for their conditions (ran at 50%, 60%, 70%, 80%, 90%, and 100% of maximal speed). Mercer et al. (2005) manipulated running velocity by first having participants determine a preferred speed, and then had these participants either increase or decrease this velocity to garner a wide range of running velocities. It could be true that a variation around preferred speed, as opposed to a variation from maximal speed, may affect the amount of acceleration experienced by the leg.

Mercer & Chona (2015) is the only group to date with published research regarding tibial acceleration with body weight support, and as such will be used for

comparative purposes. They reported decreased tibial acceleration during 60%, 70%, and 80% support conditions compared with the 0% support condition, but not in between the support conditions. Tibial acceleration being similar across different support conditions is in agreement with the data presented for the 10% and 20% support conditions, however, does not explain why the tibial acceleration in those conditions is not lower than during the 0% support condition.

There could have been some potential confounding factors to this investigation. A main concern going into data collection was participant fatigue during the experiment. In order to minimize the influence of fatigue, all participants were informed that they were allowed three minutes of rest in between each condition, and the order of body weight support conditions was counter balanced. During testing, each participant had muscular activity data recorded during the first and last minutes of each trial. A follow up analysis to that presented in the results was conducted that included a comparison of EMG between these time periods within each condition. It was determined that there was no significant difference in muscle activity for any of the muscles between time periods within a condition. Furthermore, anecdotally, no subjects indicated they needed longer than a three-minute resting period, with most opting to begin on the next condition before the three-minute period had passed.

Another potential confounding factor was the impact of sweat on EMG signals. The Alter-G treadmill utilizes a vacuum-sealed neoprene bag in order to apply continual positive pressure on the lower body, and the air inside of the bag has the potential to become warm during use. The use of kinesio-tape for application of the EMG electrodes allowed researchers to use an adhesive that was designed with the intent to resist sweat

during activity. For most subjects the use of this adhesive was enough to keep the sweat from building up on the lower body. Some subjects did sweat a considerable amount, and care was taken between each condition to wipe down the areas surrounding the electrode attachment sites.

It was considered that the Alter-G treadmill may have provided unintentional assistive forces from the harness apparatus. The effect of assistive forces on metabolic cost has been documented (Chang & Kram 1999). However, the magnitudes of assistive forces created by an Alter-G have not been recorded. More research is necessary into this area to determine the magnitude of the assistive forces being provided by the harness apparatus.

At the outset of this study, it was hypothesized that all dependent variables would be influenced by increases in speed and in body weight support. Yet, only VO_2 was observed to be influenced by speed and body weight support. VCO_2 only decreased with speed decreases, and muscle activity remained unchanged across all conditions. Previous literature (Grabowski & Kram, 2008; Teunissen et al., 2007; Grabowski, 2010) reporting that oxygen consumption and muscular activity both decrease with body weight support indicated that the two variables could be related to one another. This hypothesis makes sense considering that muscle activity during endurance running would seem to be linked to VO2. However, based on the findings of the current investigation it appears as if another mechanism plays a more important role in decreasing oxygen cost during supported locomotion than muscular activity as measured by surface EMG. There is the additional potential that relationships between muscular activity and VO2/VCO2 do exist, but only exist at levels of body weight support higher than the 20% utilized in this

investigation. It may be that although the average muscular activity during each condition remained unchanged, the pattern of muscle activation within a stride could have been altered during different conditions. This idea has been refuted in previous research, with Liebenberg et al. (2011) presenting correlation values ranging from 0.921 to 0.999 when comparing patterns of muscle activation at different body weight support levels and different speeds. Future research is necessary to determine how body weight support may influence the patterns of muscle activation, rather than just the average activity over time.

The observation of changes in VO_2 intake and VCO_2 output without seeing changes in muscle activity is interesting. This phenomenon may be occurring as a result of the harness system utilized as part of the Alter-G treadmill. Chang & Kram (1999) examined the effects of assistive and resistive horizontal forces acting on the body from a harness system, and reported that just a 6% (of total body weight) assistive horizontal force decreased VO_2 values by 22.8% compared with a no assistive force condition. The harness used in the study by Chang & Kram was not the same as the harness utilized by the Alter-G, however, it is safe to assume that there is at least the potential for some assistive forces as a result of the Alter-G harness. This would help to explain why VO_2 was influenced by both speed and body weight, but why average muscle activity did not change. The reduction in VO_2 may at least partially be a result of an assistive force during running, rather than the addition of body weight support.

Body weight supported locomotion has been utilized for clinical rehabilitation purposes for roughly thirty years (Finch, Barbeau, Bertrand 1991; Eastlack et al. 2005; Hesse at al. 2003). The results presented in this investigation should be considered carefully before being applied to a clinical population. First and foremost, the participants

utilized here were all healthy, recreationally active runners. Inherently, healthy people will locomote differently than someone who is not healthy, and even the same person may potentially locomote differently based on being healthy or injured. Additionally, with the results presented here arguing that there are subtle physiological differences while not experiencing any differences in the biomechanical variables, it is important to determine exactly what area needs the rehabilitation efforts, and how this may affect the rehabilitation methodology used. If the goal of rehabilitation is to reduce impact forces acting on the leg, it may be that levels of body weight support greater than 20% are needed to achieve this. However, if the goal of rehabilitation is to allow a person to do some exercise while experiencing a decrease in metabolic activity, utilizing 10% and 20% body weight support may be a plausible way to achieve that goal.

This study is limited in its findings for a few different reasons. First off, subjects were running in an Alter-G treadmill. Many people do not have access to this type of equipment, and would not be able to access this type of treadmill. Even when comparing to other harness systems, it has been shown that harnesses affect individuals through changes in kinematics and assistive/resistive force production. Care should be taken to understand fully the differences in an Alter-G treadmill and a harness system before applying these results to that type of system. Additionally it should be noted that although this investigation utilized experienced runners, most of these runners never had experience running in an Alter-g treadmill. There may have been somewhat of an acclimation period for these runners, and as a result this may have unintentionally changed gait kinematics. One affect this may have had was on trunk lean of the participants, especially during the supported conditions. The airflow resulting from the

lower body positive pressure is designed to push directly upwards, however, based on the location and fit of the Alter-g shorts, subjects may have experienced a feeling of leaning backwards. This appeared to be an acute response to being introduced to the Alter-g, and more work is needed to determine how acclimation to body weight support running may affect some of these variables.

On a related note, those who do find themselves with access to an Alter-G should understand that this analysis was only performed using three body weight support conditions. Clinicians that are using body weight support as a tool for chronic injury/illness rehabilitation are likely interested in conditions providing support at higher levels than 20%, and should be careful when reviewing these findings. Additionally, others that are using the Alter-G for acute injury rehabilitation or training may only be going as high as 20% body weight support, and likely running at some values between 0% and 10% support, as well as between 10% and 20% support. There may be inherently different bodily responses at different percentages of body weight support, and those using it for these purposes should be aware of the possible differences. A final limitation to the generalization of this study would be the quality of runner utilized for participation. These subjects were recreationally trained runners that were running at least 10 miles/week across different days. Untrained runners, as well as runners who may be training at a significantly higher volume than these participants, may experience different effects as a result of body weight support and speed manipulations.

When referring to the tibial acceleration data it is important to understand the concept of effective mass, which refers to the amount of mass being accelerated. During running, it is not known how someone's effective mass is changing. It may be true that

the magnitude of effective mass is changing during running with body weight support, where increasing body weight support leads to a potential decrease in effective mass. This could potentially speak to the fact that the overall magnitude of force may be decreased as more and more body weight support is added, and that a difference in effective mass may be responsible for the unchanged tibial acceleration values. More work is necessary to determine the effect of body weight support on effective mass.

Taking an average of three trials at each body weight support condition to determine preferred speed may not target an exact preferred speed for participants. Yet when examining this data, it appears as if most of these runners were able to determine their preferred pace relatively easily. All participants except one were able to stay within a range of ± 0.2 mph for each of the three conditions determining preferred speed. The fact that these participants were able to consistently determine their preferred speed levels indicates that preferred speed values (or at least values extremely close to preferred speed) were successfully determined for each level of support.

Conclusion

It is concluded that oxygen consumption was affected by both body weight support and speed; yet, lower extremity muscle activity was not. More research is needed to determine the mechanism behind oxygen consumption being lower despite seeing no difference in the amount of muscle firing, and whether this may have to do with unintentional assistive forces acting on the user by the Alter-G treadmill. People who are considering using the Alter-G treadmill should understand that although it may be possible to achieve similar levels of muscle activation while running at different body

weight support levels of 10% and 20% of body weight, the metabolic benefits are continually less and less as more body weight support is added.

Appendix I

IRB Consent Form

UNLV

INFORMED CONSENT

Department of Kinesiology and Nutrition Sciences

TITLE OF STUDY: Examining the Effects of Reduced Body Weight and Speed on Physiological Measures and Lower-Leg Muscle Activity

INVESTIGATOR(S): Michael Soucy, Shelby Hughes, Dr. John Mercer

For questions or concerns about the study, you may contact Michael Soucy **(661.904.4855, soucym1@unlv.nevada.edu)** or Dr. Mercer (**702.895.4672, john.mercer@unlv.edu).**

For questions regarding the rights of research subjects, any complaints or comments regarding the manner in which the study is being conducted, contact **the UNLV Office of Research Integrity – Human Subjects at 702-895-2794, toll free at 877-895-2794 or via email at IRB@unlv.edu.**

Purpose of the Study

You are invited to participate in a research study. The purpose of this study is to investigate whether the metabolic cost of running and/or lower limb muscle activity is changed as body weight is reduced while running at a preferred pace.

To reduce body weight, you will be asked to run in a special treadmill (Alter-G) that partially lifts you up while you are running (and therefore, effectively reducing your body weight). You'll be able to see the treadmill before deciding to participate.

Additionally, we will look to determine if metabolic cost or muscle activity is affected while running at plus and minus 10% of preferred speed while at reduced body weight. Metabolic cost is a measurement that tells us how hard you are running.

The muscles of interest are the Rectus Femoris (Quadriceps), Semitendinosus (Hamstrings), Medial Gastrocnemius (Calf), and Tibialis Anterior (Front part of leg)

Participants
You are being asked to participate in the study because you fit these criteria: you are over the age of 18 years old, you are not pregnant or think you are pregnant, you do not have any injury that would interfere with your ability to run, and you currently run at least 10 miles/week.

Procedures
If you volunteer to participate in this study, you will be asked to do the following:

- Attend a single testing session that will last approximately two hours and fifteen minutes
- We will measure/record basic descriptive information (height, weight, age, running mileage/week).
- Wear or have placed on you several instruments so we can make measurements while you run.
    - Muscle activity
        - We will place EMG leads on the skin directly covering the muscles of interest – this will allow us to measure muscle activity

          'EMG' is a tool we use to measure how active muscles are. It stands for 'electromyography'.

          To do this, we will need to shave, abrade (using a paper towel and gel, we will "rub" the skin where sensors will be attached in order to remove dead skin), and clean where we put the sensors on your skin. Each sensor is about the size of a quarter. We'll put these sensors on your quadriceps (front part of your thigh), hamstrings (back part of your thigh), calf muscle, and the front part of your lower leg. These are sort of like the type of sticker you would see if someone was doing an EKG (electrical activity of the heart)

    - Acceleration
        - We will place an accelerometer on the lower part of your leg using tape.

    - Heart rate
        - We will have you wear a heart rate monitor around the chest area to measure heart rate during each trial.
    - Oxygen cost
        - We will have you breathe into a mask (that covers your mouth and nose) that measures how much air you inhale and exhale throughout the duration of all trials. This mask is attached to an instrument that you'll is placed in a small backpack that you will

wear.

- Testing will consist of 9 total trials (3 body-weight conditions and 3 speed conditions)
    - We will use a special treadmill (Alter-G) that will partially lift you up while you are running.
    - The treadmill utilizes air pressure acting in the upward direction on a person to reduce a person's effective body weight. This is accomplished by fitting each subject with special Alter-g shorts, which then able to be zipped in to the Alter-g treadmill to create an air-tight seal. A picture of the Alter-g shorts can be found below.
- Each trial will consist of 7 minutes of running, followed by 3 minutes of rest

Benefits of Participation
There *may not* be direct benefits to you as a participant in this study. However, we hope to learn more about the relationship that speed and body weight play on lower limb muscle activity and metabolic cost of locomotion.

Risks of Participation
There are risks involved in all research studies. This study may include only minimal risks. Minimal risks include muscle fatigue as well as skin irritation from the attachment of the EMG sensors to the skin as well as the accelerometer. Additionally, the short/harness set-up in the Alter-g treadmill may cause some temporary chafing and discomfort.

To minimize the risks, we'll give you time to get used to running with different amounts of body weight support. Also, let us know if you are having any irritation or

discomfort with the EMG or accelerometer.

While you are testing, there might be other people in the laboratory who are not part of our research team. They may be observing data collection or collecting data for another study. There is the risk that you may feel uncomfortable with other people in the laboratory. We try to minimize this risk by allowing access to the lab by people who have a specific need (e.g., data collection for another project, instruction, etc.).

Cost /Compensation

There *will not* be financial cost to you to participate in this study. The study will take approximately 2 hours and fifteen minutes of your time. You *will not* be compensated for your time, however, a UNLV daily parking pass can be provided to you if need be.

Confidentiality

All information gathered in this study will be kept as confidential as possible. No reference will be made in written or oral materials that could link you to this study. All records will be stored in a locked facility at UNLV for at least 3 years after completion of the study. After the storage time the information gathered will be destroyed.

Voluntary Participation

Your participation in this study is voluntary. You may refuse to participate in this study or in any part of this study. You may withdraw at any time without prejudice to your relations with UNLV. You are encouraged to ask questions about this study at the beginning or any time during the research study. If you are currently a student in a class taught by one of the research team members, please know that participation in the research study has no influence on the class, your grade, or your relationship with UNLV.

Participant Consent:

I have read the above information and agree to participate in this study. I have been able to ask questions about the research study. I am at least 18 years of age. A copy of this form has been given to me.

____ __________________

Signature of Participant Date

____

Participant Name (Please Print)

Audio/Video Taping: The investigators may wish to take pictures of the set up and protocol to include in a manuscript/presentation. If pictures are taken, faces will not be included to ensure anonymity of subjects.

I agree to be audio or video taped for the purpose of this research study.

____ ____________________

Signature of Participant Date

____

Participant Name (Please Print)

Appendix II

Data Collection Sheet

Subject #: ___________

Date: ______________

Age: _____________

Height: ____________

Mass: ____________

Running mileage/week: ______________

Condition Order: _________________

Seniam Sensor Locations

Rectus femoris: Electrodes placed at 50% on the line from the ASIS to the superior part of the patella

Semitendinosus: Electrodes placed at 50% on the line between the ischial tuberosity and the medial epicondyle of the tibia

Medial Gastrocnemius: Electrodes placed on the most prominent bulge of the muscle

Tibialis Anterior: Electrodes placed at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus

Condition	Notes

References

Alter G. 2015. *About us*. Retrieved February 16, 2016, from http://www.alterg.com/who-we-are

Barbeau, H., & Visintin, M. (2003). Optimal outcomes obtained with body-weight support combined with treadmill training in stroke subjects. *Archives of Physical Medicine & Rehabilitation, 84*(10), 1458-1465.

Capellini, G., Ivanenko, Y.P., Poppele, R.E., & Lacquaniti, F. (2006). Motor Patterns in Human Walking and Running. *Journal of Neurophysiology,* 95(6). 3426-3437.

Centers for Disease Control. (2015). *Stroke Facts.* Retrieved September 23, 2015. http://www.cdc.gov/stroke/facts.htm

Chang, Y.H. & Kram, R. (1999). Metabolic cost of generating horizontal forces during human running. *Journal of Applied Physiology*, 86(5). 1657-1662.

Chumanov, E.S., Wille, C.M., Michalski, M.P., & Heiderscheit, B.C. (2012). Changes in Muscle Activation Patterns when Running Step Rate is Increased. *Gait and Posture,* 36(2). 231-235.

Colby, S.M., Kirkendall, D.T., & Bruzga, R.F. (1999). Electromyographical analysis and energy expenditure of harness supported treadmill walking: implications for knee rehabilitation. *Gait & Posture,* 10. 200-205.

Derrick, T.R., Hamill, J., & Caldwell, G.E. (1998). Energy absorption of impacts during running at various stride lengths. *Medicine & Science in Sports and Exercise,* 30(1). 128-135.

Eastlack, R.K.; Hargens; A.R.; Groppo, E.R.; Steinbach, G.C.; White, K.K.; Pedowitz, R.A. (2005). Lower Body Positive Pressure Exercise after Knee Surgery. *Clinical Orthopaedics and Related Research,* 431. 213-219.

Farley, C.T. & McMahon, T.A. (1992). Energetics of walking and running: insights from simulated reduced-gravity experiments. *Journal of Applied Physiology*, 73. 2709-2712.

Finch, L. & Barbeau, H. (1985). Influences of partial weight bearing on normal human gait: the development of a gait retraining strategy. *Can J Neurol Sci*, *183(12)*. Abstract.

Finch, L., Barbeau, H., & Arsenault, B. (1991). Influence of body weight support on normal human gait: Development of a gait retraining strategy. *Physical Therapy, 71*, 842-855.

Fredericson, M., & Misra, A.K. (2007). Epidemiology and Aetiology of Marathon Running Injuries. *Journal of Sports Medicine,* 37. 437-439.

Gazendam, M.G.J., Hof, A.L. (2007). Averaged EMG profiles in jogging and running at different speeds. *Gait & Posture*, 25. 604-614.

Gojanovic, B., Cutti, P., Shultz, R., and Matheson, G.O. (2012). Maximal Physiological Parameters during Partial Body-Weight Support Treadmill, *Med. Sci in Sports and Exsc.* 1935-1941.

Grabowski, A. (2010). Metabolic and Biomechanical Effects of Velocity and Weight Support Using a Lower-Body Positive Pressure Device During Walking*, Arch Phys Med Rehabil*, 91. 951-957.

Grabowski, A., Farley, C.T., & Kram, R. (2005). Independent metabolic costs of supporting body weight and accelerating body mass during walking. *Journal of Applied Physiology,* 98. 579-583.

Grabowski, A., & Kram, R. (2008). Effects of Velocity and Weight Support on Ground Reaction Forces and Metabolic Power During Running. *Journal of Applied Biomechanics,* 24. 288-297.

Hamill, J., Bates, B.T., Knutzen, K.M., and Sawhill, J.A. (1983). Variations in Ground Reaction Force Parameters at Different Running Speeds. *Human Movement Science,* 2. 47-56.

Hamill, J., Derrick, T.R., & Holt, K.G. (1995). Shock attenuation and stride frequency during running. *Huuman Movement Science,* 14. 45-60.

Hargens, A.R., Cutuk, A., White, K., Rabbani, R., & Pedowitz, R.A. (1999). Cardiovascular impact of lower body positive pressure. *Medicine & Science in Sports and Exercise*, 31(5). S367.

Hesse, S., Werner, C., Seibel, H., von Frankenberg, S., Kappel, E. M., Kirker, S., et al. (2003). Treadmill training with partial body-weight support after total hip arthroplasty: A randomized controlled trial. *Archives of Physical Medicine & Rehabilitation, 84*(12), 1767-1773.

Hof, A.L., Elzinga, H., Grimmius, W., Halbertsma, J.P.K. (2002). Speed dependence of averaged EMG profiles in walking. *Gait and Posture*, 16. 78-86.

Hoffman, M.D. & Donaghe, H.E. (2011). Physiological Responses to Body Weight-Supported Treadmill Exercise in Healthy Adults. *Arch Phys Med Rehabil,* 92. 960-966.

Hreljac, A., Marshall, R.N., & Hume, P.A. (2000). Evaluation of lower extremity overuse injury potential in runners. *Med. Sci. Sports Exerc.,* 32(9). 1635-1641.

Ivanenko, Y.P., Poppele, R.E., & Lacquaniti, F. (2004). Five basic muscle activation patterns account for muscle activity during human locomotion. *J Physiol,* 556(1). 267-282.

Klarner, T. Chan, H.K., Wakeling, J.M., & Lam, T. (2010). Patterns of Muscle Coordination vary with Stride Frequency during Weight Assisted Treadmill Walking, *Gait & Posture,* 31. 360-365.

Kyrolainen, H., Komi, P.V., & Belli, A. (1999). Change in Muscle Activity Patterns and Kinetics with Increasing Running Speed. *Journal of Strength and Conditioning Research,* 13(4). 400-406.

Lafortune, M.A., Lake, M.J., & Hennig, E.M. (1996). Differential shock transmission response of the human body to impact severity and lower limb posture. *Journal of Biomechanics,* 29. 1531-1537.

Liebenberg, J., Scharf, J., Forrest, D., Dufek, J.S., Masumoto, K., & Mercer, J.A. (2011). Determination of muscle activity during running at reduced body weight. *Journal of Sports Sciences,* 29(2). 207-214.

Masumoto, K., Delion, D., & Mercer, J. A. (2009). Insight into muscle activity during deep water running. *Medicine & Science in Sports & Exercise, 41*(10), 1958-1964.

Masumoto, K., Soucy, M.T., Bailey, J.P., & Mercer, J.A. (2016). Muscle Activity During Backward and Forward Running at Reduced Body Weight, Abstract. *National Meeting of the American College of Sports Medicine*. May 31st – June 4th.

Mercer, J.A., Applequist, B.C., & Masumoto, K. (2013). Muscle Activity While Running at 20-50% Normal Body Weight. *Research in Sports Medicine: An International Journal,* 21(3). 217-228.

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

Mercer, J.A., Bezodis, N.E., Russell, M., Purdy, A., & Delion, D. (2005). Kinetic Consequences of Constraining Running, *Journal of Sports Science and Medicine*, 4. 144-152.

Mercer, J.A. & Chona, C. (2015). Stride length-velocity relationship during running with body weight support. *Journal of Sport and Health Science,* 4(4). 391-395.

Mercer, J.A., Vance, J., Hreljac, A., & Hamill, J. (2002). Relationship between shock attenuation and stride length during running at different velocities. *Eur J Appl Physiol,* 87. 403-408.

National Institute of Health. (2013). *Questions and answers about hip replacement.* Retrieved September 23, 2015, from http://www.niams.nih.gov/Health_Info/Hip_Replacement/

Patil, S., Steklov, N., Bugbee, W.D., Goldberg, T., Colwell Jr., C.W., & D'Lima, D.D. (2013). Anti-gravity treadmills are effective in reducing knee forces. *Journal of Orthopaedic Research,* 31(5). 672-679.

Patino, M.S., Goncalves, A.R., Monteiro, B.C., Santos, I.L., Barela, A.M.F., & Barela, J.A. (2007). Kinematic, Kinetic, and electromyographic characteristics of young adults walking with and without partial body weight support. *Revista Brasileira de Fisioterapia,* 11. 19-25.

Running USA (2014). *2014 State of the Sport – Part II: Running Industry Report*. Retrieved January 30, 2016, from, http://www.runningusa.org/2014-state-of-the-sport-part-ii- running-industry-report

Running USA (2015). *2015 State of the Sport- U.S. Racing Trends*. Retrieved September 23, 2015, from, http://www.runningusa.org/2015-state-of-sport-us-trends?returnTo=annual-reports

Sloniger, M.A., Cureton, K.J., Prior, B.M., & Evans, E.M. (1997). Lower extremity muscle activation during horizontal and uphill running. *J Appl Physiol,* 83(6). 2073-2079.

Stamford, B.A. (1975). Maximal oxygen uptake during treadmill walking and running at various speeds. *J Appl Phsyiol*, 39(3). 386-389.

Sullivan, K.J., Knowlton, B.J., & Dobkin, B.H. (2002). Step Training with Body Weight Support: Effect of treadmill Speed and Practice Paradigms on Poststroke Locomotor Recovery, *Arch Phys Med Rehabil,* 83. 683-691.

Teunissen, L. P. J., Grabowski, A., & Kram, R. (2007). Effects of independently altering body weight and body mass on the metabolic cost of running. *Journal of Experimental Biology, 210*, 4418-4427.

Verbitsky, O., Mizrahi, J., Voloshin, A., Treiger, J., & Isakov, E. (1998). Shock Transmission and Fatigue in Human Running. *Journal of Applied Biomechanics,* 14. 300-311.

Visintin, M., Barbeau, H., Korner-Bitensky, N., & Mayo, N. E. (1998). A new approach to retrain gait in stroke patients through body weight support and treadmill stimulation. *Stroke, 29*, 1122-1128.

Curriculum Vitae

Michael Soucy
2101 W. Warm Springs Rd. #3323
Henderson, NV 89014
(661) 904-4855
soucym1@unlv.nevada.edu

Education

University of Nevada, Las Vegas: 2014 to Present
Master of Science, Kinesiology
Emphasis in Biomechanics
Advisor: Dr. John Mercer
GPA: 3.93

California Lutheran University: 2010 to 2014
Bachelor of Science, Exercise Science
Emphasis in Health Professions
Graduated with Departmental Distinction
Advisor: Dr. Michele LeBlanc
GPA: 3.37

Professional Experience

University of Nevada, Las Vegas: August 2014 to Present
Graduate Research and Teaching Assistant
Research responsibilities: Research design, data collection, data processing, data analysis, statistical analysis, grant writing, manuscript preparation, and presentation of original research within the Sports Injury Research Center and Department of Kinesiology and Nutrition Sciences.
Teaching responsibilities: Instruct Kin-346L Biomechanics Labs. Development of presentations and lab materials for topics relating to linear and angular kinetics/kinematics, functional anatomy, projectile motion, and motion capture.

Self-Employed: September 2009 to Present
Private Tutor
Assist elementary, junior high, and high school students become more proficient in math and science courses. This includes helping students with homework, note taking, and studying for exams.

California Lutheran University: September 2013 to May 2014
Exercise Science Departmental Assistant

Provide assistance to professors in instructing lower-level Exercise Science courses. Additionally, hold office hours and provide tutoring help to other students in the Exercise Science Department.

California Lutheran University: May 2013 to August 2013

Darling Summer Research Fellowship

Conduct research investigating the effect of fatigue on patellar tendon strain and torque production. Responsibilities included data collection, data processing, and dissemination of results.

California Lutheran University: September 2011 to May 2013

Event Services/ Game Management Staff

Set up and tore down various events on campus. Responsible for in-game services during NCAA sanctioned games including keeping score, running the scoreboard, and running the shot clock.

Manuscripts

Investigating Muscle Activity in Single Leg Hopping and Running. **Soucy, M.T.,** Lopez, C., Lau, T., Mercer, J. Manuscript in Preparation.

Does wearing a triathlon wetsuit influence resting blood pressure for females?. **Soucy, M.T.,** Ciulei, M., Mercer, J. Manuscript accepted for publication. *Gazzetta Medica Italiana*.

Kinematic and Ground Reaction Force Characteristics of Keeper-Independent and Keeper-Dependent Soccer Penalty Kicks. LeBlanc, M., Kacena, J., McCardell, M., Prosser, K., **Soucy, M.T**., Truver, B. Manuscript in Preparation.

Peer-Reviewed Articles

Dufek, J., Harry, J., **Soucy, MT,** Guadagnoli, M., & Lounsberry, M. Effects of Active Workstation Use on Walking Mechanics and Work Efficiency, *Journal of Novel Physiotherapies,* 6(3).

Presentations

American College of American College of Sports Medicine, National Meeting. Presentation Title: Muscle Activity of the Lower-Limb During Single-Legged Hopping. May 2016.

American College of Sports Medicine, Southwest Chapter. Presentation Title: Muscle Activity of the Lower-Limb During Single-Legged Hopping. October 2015.

American College of Sports Medicine, National Meeting. Presentation Title: Influence of body weight support and direction of locomotion on preferred gait. May 2015.

Rebel Grad Slam: 3-Minute Thesis Competition, UNLV Graduate College. Presentation Title: Influence of body weight support and direction of locomotion on preferred gait. November 3-7, 2014.

Undergraduate Research Symposium, California Lutheran University. Presentation title: The Effect of Music on Performance of the Twenty-Yard Shuttle Run. December 2014.

American College of Sports Medicine, Southwest Chapter. Presentation title: The Effect of Fatigue on Patellar Tendon Strain In Healthy Subjects. October 2013.

Professional/Community Service

Student mentor for STEM days at UNLV, 2015-2016

Student mentor for summer research projects through STEP-UP, 2015-2016

Student mentor for summer research projects through INBRE, 2015-2016

Funding

Mercer, J.A, Bailey, J.P, Barker, L, **Soucy, M.T**. Innovative intervention exercise program using body weight support and real-time impact feedback. Arthritis Foundation. 2015. $75,594. Not funded.

Honors

Departmental Distinction, Department of Exercise Science, California Lutheran University, 2014

Scholar Athlete, California Lutheran University, 2011

Professional Memberships

American College of Sports Medicine, Southwest Chapter, 2012 to Present

Certifications

CPR/ First Aid/ AED: American Red Cross, 2011 to Present