Influence of Tactile Feedback on Trunk Coordination and Electromyography During Walking

by

Braden Heath Romer

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Approved by

Wendi H. Weimar, Chair, Associate Professor of Kinesiology
Sean Gallagher, Associate Professor of Industrial & Systems Engineering
Nels H. Madsen, Professor of Mechanical Engineering
Matthew W. Miller, Assistant Professor of Kinesiology
Gretchen D. Oliver, Assistant Professor of Kinesiology

Abstract

Footwear is a necessary component of modern life, serving wide-ranging purposes from fashion apparel to personal protective equipment. Despite what would appear as an otherwise obscure addition of clothing or equipment for individuals, past research has indicated considerable footwear effects on spatiotemporal, kinematics, and kinetics during walking and running. Likewise, footwear has been implicated in altered lumbar and pelvic positioning while standing. While early gait-related research focused on descriptive information to examine the relative timing of body segments throughout the gait cycle, the relationship between trunk, pelvic, and lower extremity movements in maintaining forward momentum and increasing economy during gait is not yet fully understood. Furthermore, despite the considerable footwear effects on gait mechanics, it is not yet known whether footwear has a significant effect on segment synchronicity during walking. Therefore, the purposes of this investigation was: (1) to investigate the effects of velocity and cadence on lumbopelvic rhythm during overground walking; (2) to investigate the effects of footwear on lumbopelvic rhythm during gait; (3) to examine the effects of enhanced tactile feedback (i.e. barefoot, textured insole) on the onset of muscle activity at foot strike; (4) to examine the effects of spatiotemporal and footwear variation on the magnitude of lumbar (i.e. erector spinae) and lower extremity muscular activity (e.g. gluteus medius, biceps femoris, soleus, peroneus longus) during the stance phase of gait, and (5) to investigate differences in the kinematics and electromyography between males and females.

Results indicated significant increases in height normalized stride length, cadence, and velocity as spatiotemporal variables were manipulated from 75% of Normal, to Normal, and 125% of Normal. Results also indicated that altering velocity or cadence had a significant effect for TP_{mean}, TP_{SD}, TL_{mean}, TL_{SD}, PL_{mean}, PL_{SD}, Pelvic Rotation, Trunk Rotation, and Leg Rotation; however the influence of velocity appeared to be smaller in which significant differences were found only for TP_{mean}, Pelvic Rotation, and Trunk Rotation. Likewise, normalized peak sEMG during the stance phase increased as velocity and cadence increased from 75% of Normal, to Normal, to 125% of Normal for lower extremity and lumbar musculature.

When examining footwear effects, results indicated that footwear had a significant effect on the preferred walking velocity, cadence, and normalize stride length of the participants. Specifically, the barefoot and barefoot-like condition (insole-only) exhibited a significantly shorter normalized stride length and lower velocity relative to both shod conditions (shod and insole-shod). Conversely, the present study found no significant footwear effect on normalized peak muscle activity during the stance phase or onset of muscle activity at foot strike. The only kinematic variable to display a significant footwear effect was the relative phase of the pelvis-leg, with barefoot and insole-only exhibiting greater asynchronicity than the shod and insole-shod conditions.

Finally, the present study revealed some gender differences are present during walking. For instance, females exhibited a significantly longer normalized stride length and higher cadence than males for the altered footwear conditions only. Conversely, there were no significant differences between males and females for during any of the altered walking velocity or cadence conditions. Relative phase differences were only found to be significantly different during the footwear conditions, with males exhibiting a significantly greater TP_{SD}, indicating that the trunk-pelvis of males was significantly more variable than females when footwear was altered. Finally, no significant gender differences were noted for normalized peak muscle activity during the stance phase of walking during altered velocity, altered cadence, or altered footwear.

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Chapter I. Introduction

For many individuals, walking is a fundamental means of moving from one location to another. Walking also serves as the foundation for other higher-level motor skills (e.g. running, skipping), providing a base for individuals to build more advanced skillsets (Haywood & Getchell, 2009). It is also a convenient and versatile means of locomotion, owing in large part to the functional versatility of the lower extremity and lumbopelvic regions. While walking is often thought of as a secondary motor task, a significant amount of resources have been spent on understanding normal and pathological gait (Perry, 1992; Whittle, 2003). Specifically, changes in gait economy and variability associated with neural deficits, altered postures, musculoskeletal limitations, and footwear are areas that continue to be examined by researchers (Perry, 1992; Smidt, 1990; Whittle, 2003). Furthermore, the aforementioned factors are capable of altering limb motion patterns associated with timing of musculature, thereby reducing economy and increasing the likelihood of cumulative trauma disorders (Bird, Bendrups, & Payne, 2003; Braunstein, Arampatzis, Eysel, & Bruggemann, 2010; Crosbie & Vachalathiti, 1997; Crosbie, Vachalathiti, & Smith, 1997a, 1997b; Divert, Mornieux, Baur, Mayer, & Belli, 2005; Divert et al., 2008; Gefen, Megido-Ravid, Itzchak, & Arcan, 2002; Y. Huang et al., 2010; Y. P. Huang et al., 2011; C. J. Lamoth, Beek, & Meijer, 2002; C. J. Lamoth, Meijer, et al., 2002; C. J. C. Lamoth, Daffertshofer, Meijer, & Beek, 2006; Maki, Perry, Norrie, & McIlroy, 1999; Murley, Landorf, Menz, & Bird, 2009; W. Wu et al., 2002; W. Wu et al., 2004; W. H. Wu et al., 2008). Footwear, in partially fulfilling its intended purpose, has been shown to significantly alter posture and gait mechanics when compared to barefoot gait; however, kinematic and kinetic changes associated with footwear are not always positive as the intended design would suggest (Lieberman et al., 2010; Robbins & Gouw, 1990; Sato, Fortenbaugh, & Hydock, 2012; Shakoor, Lidtke, Sengupta, Fogg, & Block, 2008; Shorten & Mientjes, 2011). Recently, researchers have suggested that, despite significant improvements in footwear technology and materials science within the last several decades, kinematic and kinetic changes associated with shod gait may be a contributing factor to increased lower extremity injury rates (Lieberman et al., 2010; van Gent et al., 2007).

When focusing on the lower extremity, researchers have found that 37% – 79% of all individuals that regularly take part in weight-bearing physical activity develop a lower extremity injury, with a recurrence rate of 20% - 70% (van Gent et al., 2007). Likewise, the U.S. Department of Labor has repeatedly found that sprains, strains and tears are the dominant type of occupational injury for all types of workers, with injuries to the low back and lower-extremity accounting for almost 60% all workplace related injuries. In addition, while many individuals often associate a low back or lower-extremity injury with a specific event, epidemiological research has shown that most injuries result from a cumulative response over a considerable period of time (Andersson, 1998; Kraus, Schaffer, McArthur, & Peek-Asa, 1997; Magee, 2008; Marras, 2000; Marras et al., 1995; McGill, 2007). Therefore, it is not altogether surprising that four of the top five occupations with the highest injury rates also contain high volumes of physical activity and significant amounts of standing/ walking (laborers and manual material handlers; nurses or other healthcare individuals; janitors and cleaners, and lightweight delivery services).

Another common concern within ergonomics is the role of awkward postures, or positions which significantly deviate from neutral or normal during a task, in the development of cumulative trauma disorders, a common characteristic for the previously mentioned occupations (M. Jackson, Solomonow, Zhou, Baratta, & Harris, 2001; Marras, 2000; Marras et al., 1995; Xu, Bach, & Orhede, 1997). Though the ergonomics field does not typically view awkward postures as directly affecting the role of active and passive tissues during task completion, there is support that awkward limb positioning will reduce the capability of a tissue to tolerate work (Aultman, Drake, Callaghan, & McGill, 2004; Aultman, Scannell, & McGill, 2005; M. Jackson et al., 2001). Therefore, it has been suggested that awkward postures will, by altering the behavior of the fatigue failure curve, reduce the number of cycles to failure (e.g. development of pain or injury) of a particular tissue. As a result, awkward postures have been implicated as a compounding factor in the development of cumulative trauma disorders.

Research from the human performance field suggests that awkward postures may be directly involved in altering the response and loading patterns of active and passive tissues (Burton et al., 2005; Gallagher, Marras, Litsky, & Burr, 2005; N. D. Jackson, Gutierrez, & Kaminski, 2009; Kibler, 1998). The kinetic chain model of movement is based on the supposition that the proper sequencing of muscle activity patterns occurs in a proximal-to-distal nature, which thereby determines the level of movement economy and coordination. Furthermore, the kinetic chain model emphasizes the existence of a neuromuscular linkage, in which movements are initiated by torso musculature, followed closely by the extremities (McMullen & Uhl, 2000; Oliver, Dwelly, Sarantis, Helmer, & Bonacci, 2010; Oliver, Stone, & Plummer, 2010). Therefore, the ideal firing and recruitment of the involved musculature is key to maintaining optimal biomechanical patterns during a movement and avoiding awkward postures. Within a sports and clinical setting, it has been postulated that detriments in muscular firing patterns, particularly those within the lumbopelivc hip complex (e.g. core), are contributing factors in a variety of injuries, such as shoulder complex pathology, low back pain and ACL injuries, among others (McGill, 2007; Willson, Dougherty, Ireland, & Davis, 2005).

Furthermore, other research has found that individuals with a history of lateral ankle sprains displayed altered firing patterns of the hip extensors on the ipsilateral and contralateral sides, indicating a significant relationship between lower extremity mechanoreceptors and torso musculature (Bullock-Saxton, Janda, & Bullock, 1994).

While not yet completely understood, it is believed that the early onset of muscle activation within the core is necessary to improve the strength of the system, thereby providing a stable and reliable base from which more complex movement patterns can begin (Kibler, 1998). Similarly, others have found that the transverse abdominis, followed almost simultaneously by the multifidus, fire prior to any other limb movement or individual-initiated adjustments (Hodges & Richardson, 1997). In addition to the aforementioned muscles, more recent results have indicated that erector spinae activity also occurs prior to the onset of extremity excursion (Moseley, Hodges, & Gandevia, 2002). Therefore, it has been postulated that injuries or equipment which alter the mechanical and physiological variables involved with a movement will also considerably raise the associated injury risk.

Despite the often perceived passiveness of footwear during movement, there is support within the literature to indicate that footwear contributes to awkward posture, resulting in significant physiological and mechanical alterations during movement. For instance, Sato and colleagues (2012), examining the effects of traditional running shoes and weightlifting footwear on barbell squat kinematics, reported significant differences in lower extremity and trunk kinematics. Specifically, participants displayed significantly greater trunk lean and trunk displacement when wearing traditional running shoes. Likewise, Shakoor et al. (2008) reported an 8-12% reduction in the peak external knee adduction moment, a contributing factor to the development of knee osteoarthritis, when participants wore specialized footwear as opposed to self-chosen or control footwear. Aghazadeh and Liu (1994) reported a significant reduction in the lifting capacity of participants when tasks were performed in footwear with elevated instead of flat heels. Similarly, Opila-Correia (1990) reported significant changes in spatiotemporal variables and joint kinematics during high-heeled gait relative to low or flat-heeled walking. Specifically, it was reported that participants walked significantly slower, exhibited shorter stride lengths, increased knee flexion at foot strike and during stance phase, and displayed a reduced range of pelvic motion in the sagittal plane during high-heeled gait. More recently, it has been reported that high-heeled walking significantly reduced the ankle plantarflexor moment during stance, followed by a significantly increased hip flexor moment during the subsequent swing phase (Esenyel, Walsh, Walden, & Gitter, 2003). Clinically, footwear manipulations, through rehabilitative interventions or orthotic prescription, have routinely been utilized to significantly alter posture and movement patterns. (Gross, Davlin, & Evanski, 1991).

There is also support within the motor behavior literature for understanding the role of footwear in gait mechanics. In 1986, Karl Newell suggested that human movement results from the interaction of the individual with the environment in which the movement occurs and the task to be undertaken (Haywood & Getchell, 2009). Furthermore, any variability in the three aforementioned factors will result in changes to the movement pattern. Therefore, the three constraints (individual, environmental, and task) create a dynamic model for understanding the development and execution of particular movement (Haywood & Getchell, 2009). The appeal of Newell’s Model of Constraints is that emphasis is placed on the influence of where the individual moves and what the individual does during particular movements. For example, it has been argued that while footwear choice may not considerably alter a carefully controlled walk, increasing the environmental or task demands would notably alter gait patterns (Haywood &

Getchell, 2009). As a result, a cumulative response of altered movement patterns will lead to changes in long-term motor behavior. Through its protection of the dorsal and plantar surfaces of the foot, footwear could be considered a necessary equipment choice for successful translation. Because the foot is the only extremity in direct contact with the ground during normal gait, proper functionality is vital to the execution of the chosen movement. Likewise, any equipment which alters the behavior of the lower extremity would lead to altered motor patterns. Therefore, footwear type would appear to be intricately linked to the dynamics of any movement including walking.

Understanding the kinematic differences in barefoot versus shod gait may allow for a greater understanding of the kinetic differences. Saunders and colleagues (1953) were one of the first to attempt to classify the movements of limbs during gait, resulting in the formation of a system collectively referred to as the Six Determinants of Gait. Of particular interest to the proposed research is the pelvic rotation, which primarily occurs during late swing and terminal stance. It is believed that the pelvic rotation allows for a functional leg length increase during ipsilateral swing, reducing hip amplitude range of motion and center of mass excursion, thereby decreasing the physiological costs of walking. However, unlike the linear relationship between running velocity and oxygen consumption, it has been found that there is a curvilinear relationship between these variables during walking (Margaria, 1976). The increased physiological cost associated with walking at higher velocities is likely related to an increased stride length and related changes in mechanical variables (e.g. anterior-posterior forces) (Divert et al., 2005; Divert et al., 2008; Margaria, 1976). While walking velocity is the product of stride length and stride rate (e.g. cadence), increasing stride rate appears to be the predominant factor in increasing walking velocity during barefoot translation (Divert et al., 2008; Lieberman et al., 2010; Romer et al., 2012). Conversely, recent research suggests that increasing stride length is the predominant factor to increasing walking velocity during shod gait (Divert et al., 2008; Lieberman et al., 2010).

The alteration of the aforementioned spatiotemporal variables may have considerable implications for the pelvic step. As described by Ducroquet et al. (1968), the pelvic step is a transverse pelvic girdle rotation, acting to lengthen the step as velocity increases. It is generally accepted that there is a positive relationship between walking velocity and the magnitude of the pelvic step for velocities greater than 0.8 ms^{-1} (Bruijn, Meijer, van Dieen, Kingma, & Lamoth, 2008; Crosbie & Vachalathiti, 1997; Crosbie et al., 1997a; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002). Furthermore, Ducroquet et al. (1968) asserted that the counter-rotation of the thorax is initiated to limit total angular momentum about the vertical axis during gait; however, more recent findings question the thorax and pelvic contributions to total body angular momentum during walking (Bruijn et al., 2008). Therefore, it has been argued that the current knowledge regarding the understanding of the pelvic step is poor, necessitating an updated approach to understand the mechanisms responsible for pelvic step alterations (Bruijn et al., 2008). For instance, yet to be understood is the effect of walking velocity, stride length, and cadence on the relative timing and magnitude of thorax, pelvic, and hip movements during gait.

As previously alluded to, a near universal finding of footwear research is an increased step length during shod locomotion, regardless of walking velocity or cadence (Nurse, Hulliger, Wakeling, Nigg, & Stefanyshyn, 2005; Romer et al., 2012; Wegener, Hunt, Vanwanseele, Burns, & Smith, 2011). It had been hypothesized that observed changes in shod gait kinematics are the result of mechanical factors (i.e. increased inertia of the distal segment); however, other authors have hypothesized a neural cause (e.g. reduced afferent feedback) as the root of alterations in shod gait mechanics (Divert et al., 2008; Kurz & Stergiou, 2003; Mundermann, Stefanyshyn, & Nigg, 2001; Murley et al., 2009; Nurse et al., 2005). Specifically, Kurz and Stergiou (2003) found that gait variability can be significantly affected by peripheral sensory information. Furthermore, it has been suggested that tactile feedback is an important control mechanism for maintaining balance and locomotion, with authors finding augmented plantar facilitation capable of improving balance during static trials and altering muscle activity during gait (Bird et al., 2003; Maki et al., 1999; Nashner, 1980; Nurse et al., 2005). Likewise, Kurz and Stergiou (2003) found the reduced sensory information associated with shod gait resulted in a significant reduction in kinematic variability as compared to barefoot locomotion. Therefore, it appears that footwear mutes or reduces the tactile and proprioceptive feedback received by the lower extremity, resulting in altered gait mechanics.

Support for a neural interaction in gait mechanics has also arisen from areas of research such as peripheral neuropathy, textured insoles, and balance. For instance, there is evidence that footwear reduces the tactile and proprioceptive feedback of the lower extremity, resulting in an altered response (i.e. timing and magnitude) of lower extremity and torso musculature during standing and gait (Allet et al., 2009; Bird et al., 2003; Maki et al., 1999; Murley et al., 2009; Nawoczenski & Ludewig, 1999; Nurse et al., 2005; Nurse & Nigg, 2001; Wakeling & Liphardt, 2006; Wakeling, Liphardt, & Nigg, 2003). Nurse et al. (2005) found significant differences in the magnitude of lower leg muscle activity (i.e. soleus, tibialis anterior) during walking when using a textured versus smooth insert. Furthermore, Bird et al. (2003) reported significant differences in the timing of torso and hip musculature (i.e. erector spinae, gluteus medius) during walking with varying foot wedges. Relative to barefoot gait, shod gait most often displays an increased stride length, increased anteroposterior ground reaction forces (GRF), and increased vertical GRF impact transient, changes which may assist in increasing sensory feedback of the foot during locomotion (Kurz & Stergiou, 2003; Lieberman et al., 2010).

Beyond footwear research, support for the neural interaction in gait mechanics can be found in the mechanical and physiological changes during the gait of pathological populations (e.g. peripheral neuropathy) or healthy individuals with an induced reduction in plantar sensation (Manor & Li, 2009; Savelberg et al., 2010). Consequently, these findings suggest that footwear may result in considerable differences in the loading of the passive and active tissues during gait. If it is possible to cause a meaningful change in the efferent response of the active musculature through an altered afferent response, greater loading of the associated elastic tissues may spare the passive tissues from considerable microtrauma.

Summary

In conclusion, there is considerable evidence that footwear significantly alters gait mechanics (Braunstein et al., 2010; Kristen, Kastner, Holzreiter, Wagner, & Engel, 1998; Lieberman et al., 2010; Wegener et al., 2011). Specifically, authors have found an increased stride length, decreased cadence, increased impact transient, and altered joint kinematics (e.g. increased hip flexion, decreased plantarflexion) during shod locomotion (Braunstein et al., 2010; Kristen et al., 1998; Lieberman et al., 2010; Wegener et al., 2011). Physiological changes have also been found during shod gait, including a reduced metabolic economy and altered or delayed muscular activation of lower leg and torso musculature (Bird et al., 2003; Divert et al., 2008; Maki et al., 1999; Murley et al., 2009; Nurse et al., 2005; Nurse & Nigg, 2001). Furthermore, footwear has been implicated in altered lumbar and pelvic positioning while standing and may even alter the response of lower extremity and lumbar musculature (Aghazadeh & Lu, 1994; Chiu & Wang, 2007; Cikajlo & Matjacic, 2007; Keenan, Franz, Dicharry, Della Croce, &

Kerrigan, 2011; C.-M. Lee, Jeong, & Freivalds, 2001). The combined findings of the Kinetic Chain model and Newell's Model of Constraints provide a basis for the role of footwear leading to altered movement patterns during gait. Specifically, the role of footwear on mechanical and physiological variables during gait must be further delineated to provide an understanding of injury pathways and future product innovation. The relationship between trunk, pelvic, and lower extremity movements during gait is important in maintaining forward momentum and increasing economy; however, the relevance of the timing of the segmental movements is not yet fully understood. While countless studies have examined the role of footwear on walking and running mechanics, only one study has examined the effect of spatiotemporal variables on lumbopelvic rhythm during gait (Y. Huang et al., 2010). However, to the knowledge of the author, no study has simultaneously examined the role of footwear on limb segment timing and muscle activity of the lumbopelvic and lower-extremity areas, nor the effect of independently manipulating spatiotemporal variables (i.e. velocity, cadence) on thorax-pelvic coordination. To advance the understanding of human movement, the present project examined the effect of spatiotemporal variables and footwear manipulation on trunk coordination and muscle activation during walking.

Purpose of the Study

The purposes of this investigation was multi-faceted: (1) to investigate the effects of velocity and cadence on lumbopelvic rhythm during overground walking; (2) to investigate the effects of footwear on lumbopelvic rhythm during gait; (3) to examine the effects of enhanced tactile feedback (i.e. barefoot, textured insole) on the onset of muscle activity at foot strike; (4) to examine the effects of spatiotemporal and footwear variation on the magnitude of lumbar (i.e. erector spinae) and lower extremity muscular activity (e.g. gluteus medius, biceps femoris, soleus, peroneus longus) during the stance phase of gait, and (5) to investigate differences in the kinematics and electromyography between males and females.

Hypotheses

The null hypotheses for the present study are as listed:

H_{01}: Velocity will not have a significant impact on the magnitude or timing of lumbopelvic rhythm during overground walking.

H_{02}: Cadence will not have a significant impact on the magnitude or timing of lumbopelvic rhythm during overground walking.

H_{03}: There will not be any significant differences in lumbopelvic rhythm between barefoot and shod overground walking.

H_{04}: Enhanced tactile feedback will not result in a significantly different onset of muscular activation for either lumbar or lower extremity musculature.

H_{05}: Males and females will not differ in kinematics or electromyography during overground walking.

Limitations

Limitations for the present study are:

1. Self-reported health status will be used.
2. Not every type of footwear will be evaluated.
3. Participants will not be demarcated based on the amount of barefoot gait experience.
4. The age of participants will be restricted to individuals younger than 35 years of age.

Delimitations

The delimitations for the present study are:

1. Participants will be required to wear retroreflective markers bilaterally.

2. Surface electrodes for the peroneus longs, soleus, and biceps femoris will be collected unilaterally (right side of body) while the gluteus medius will be collected bilaterally.
3. Individuals with chronic low-back pain will be excluded from participation.
4. The footwear and inserts that will be utilized for this study will be new.
5. The participants will be assessed in the Auburn University Lower Extremity Laboratory which contains space specifically designed for gait-related research.

Definition of Terms

Active Tissues

The musculoskeletal tissues that directly contribute to human movement and participate in internal force production (e.g. muscles).

Kinematics

Kinematics is a branch of mechanics that is involved in the description of movement without regard to the forces that have caused the movement. Variables include linear and angular displacement, velocity, and acceleration.

Kinetics

Kinetics is a branch of mechanics that focuses on the internal and external forces that have caused a movement.

Kinetic Chain

The kinetic chain model is based on a network of individual links that, through mutual dependency and sequencing of muscle activity patterns in a proximal-to-distal nature, allow for the execution of coordinated and efficient movements.

Lumbopelvic Rhythm

Refers to the postural relationship between the pelvis and lumbar spine during static and dynamic movements.

Passive Tissues

The tissues of the body which may passively produce force, restrict motion, modify the direction of the applied force, or provide support, yet do not take part in active force production. Traditionally, these tissues would be regarded as fascia, ligaments, and various retinacula.

Relative Phase

The timing relationship between segments during walking is referred to as the relative phase, which provides a measure of the coordination between two segments during a movement.

Surface Electromyography

Electromyography is the study of electromyograms (EMG), the electrical signal associated with muscular contraction. Surface electromyography (sEMG) is a technique for evaluating and recording the voluntary muscle activity. Finally, an electromyograph is representative of the spatiotemporal summation of motor unit action potentials (MUAP) within the capture volume of the electrode.

Chapter II. Review of Literature

Walking has been the fundamental means of locomotion for humans for thousands of years, providing a convenient means of short-term transportation. Similarly, various forms of footwear have been used throughout mankind's history, however, the burgeoning footwear and fitness market of the 1970's resulted in a significant transformation in the design and materials utilized in modern footwear (Bramble & Lieberman, 2004; Lieberman et al., 2010; McDougall, 2009). This study will build upon previous literature examining the influence of footwear on gait mechanics and the role of lower extremity, pelvic, and thorax movements to better understand movement pattern changes that may lead to low back or lower-extremity pathology. The purposes of this investigation was multi-faceted: (1) to investigate the effects of velocity and cadence on lumbopelvic rhythm during overground walking; (2) to investigate the effects of footwear on lumbopelvic rhythm during gait; (3) to examine the effects of enhanced tactile feedback (i.e. barefoot, textured insole) on the onset of muscle activity at foot strike; (4) to examine the effects of spatiotemporal and footwear variation on the magnitude of lumbar (i.e. erector spinae) and lower extremity muscular activity (e.g. gluteus medius, biceps femoris, soleus, peroneus longus) during the stance phase of gait, and (5) to investigate differences in the kinematics and electromyography between males and females.

This chapter is divided into five sections. Section one describes human gait and the various phases of walking. Section two describes the role of lumbopelvic region in human locomotion. Section three describes the influence of footwear on gait mechanics. Section four describes the role of plantar sensation and tactile feedback in human locomotion. Section five summarizes the pertinent findings of the previously discussed literature, emphasizing its relevance to the present study.

Section 1: Human Gait and the Phases of Walking

To differentiate between walking and other gait variations (i.e. running), walking can be defined as involving the use of two alternating legs, providing both support and propulsion, in which one foot is in contact with the ground at all times, and often accompanied by a period of double support (Perry, 1992; Smidt, 1990; Whittle, 2003). While gait can refer to a variety of movements (e.g. walking, running, trotting), for the purposes of this paper, gait will be in reference to the normal, bipedal pattern of forward locomotion. A complete gait cycle is one in which a limb cycles from a period of support, to non-support, to ipsilateral support. Because of the cyclic nature of gait, this cycle is completed alternately by each of the lower limbs in continuous fashion. Perry (1992) has outlined three basic approaches to gait analysis: (1) division of the gait cycle based on the reciprocal contact of the two lower extremities; (2) the utilization of spatiotemporal variables to describe the gait cycle; and (3) the examination of the various components of the gait cycle based on the functional significance of each event. This section provides an overview of each of the aforementioned gait analysis approaches. Furthermore, information concerning the phases and functions of the gait cycle are also described.

Division of Gait Cycle based on the Reciprocal Contact of the Lower Extremity

The simplest approach to gait analysis is a division based upon the reciprocal patterns of ground contact by the feet (Perry, 1992; Whittle, 2003). During walking, at any point in time, one limb serves as the base of support while the other limb is in motion to the next support site. As previously mentioned, a complete gait cycle involves a period of support and non-support for the same limb. Because of the cyclic nature of walking, this process continually repeats itself, with no true ending or starting point, until the destination is reached. In traditional gait analysis, the moment of initial foot contact has been utilized as the start of the gait cycle. The term initial contact is utilized as the generic term due to variations in foot strike patterns between pathological or types of gait (e.g. running). Within this approach, the gait cycle can be divided into two distinct periods: stance phase and swing phase. The stance phase is used to designate the entire time period that the limb is in ground contact, beginning at initial contact and ending at toe-off. The swing phase encompasses the period of time in which the limb is airborne, beginning at toe-off and ending at initial contact, for the purposes of limb advancement (Figure 1).

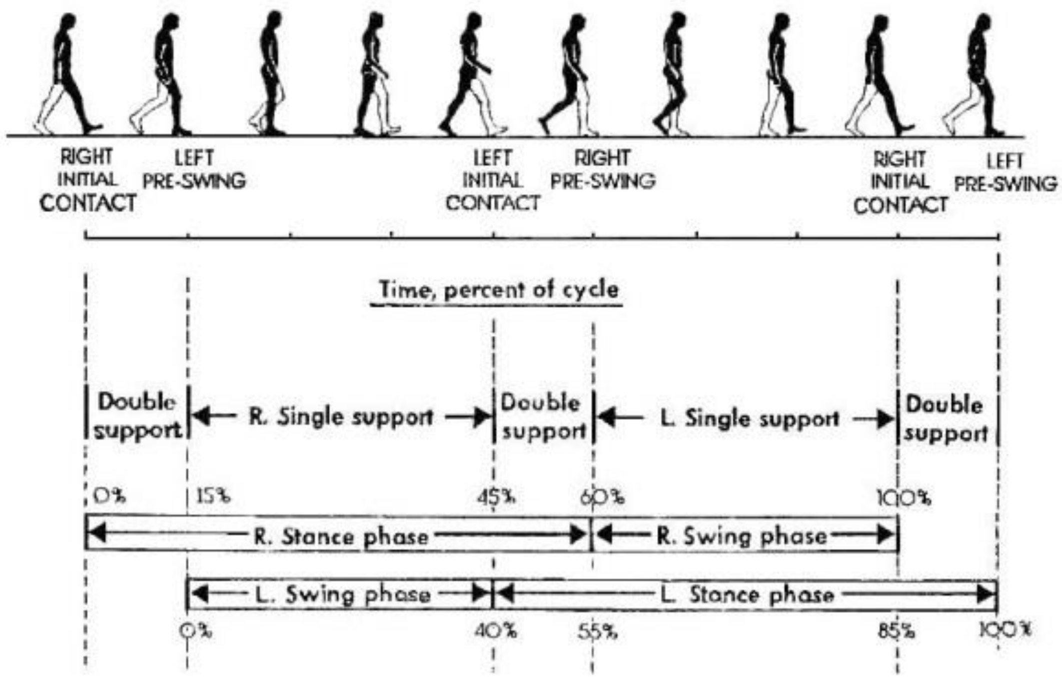

Figure 1. Phases of Gait (Kirtley, 2008).

Division of Gait Cycle based on Spatiotemporal Variables

The second approach to analyzing gait builds upon the foundation of the first approach, but further divides the sequencing of gait events based on the temporal relationship between the lower limbs (Perry, 1992). Specifically, the stance phase is divided into initial double stance phase, single limb support, and terminal double stance phase. Initial double stance occurs after initial contact (ipsilateral initial contact), signifying the beginning of the gait cycle. Though there is some load sharing between limbs during this phase, the forward limb is in the initial stages of weight acceptance while the rear leg is beginning limb advancement. Single limb support occurs when the contralateral foot ceases ground contact (contralateral toe-off) for the swing phase. As the name suggests, there is no load sharing between limbs and the entire weight of the body is supported by the stance leg. Terminal double support stance is most begins at the instant of ground contact by the swing leg (contralateral initial contact), continuing until the original stance limb ceases ground contact (ipsilateral toe-off). Because of the cyclic and offset nature of walking, the initial double stance phase for one limb corresponds to contralateral terminal double stance (Figure 2).

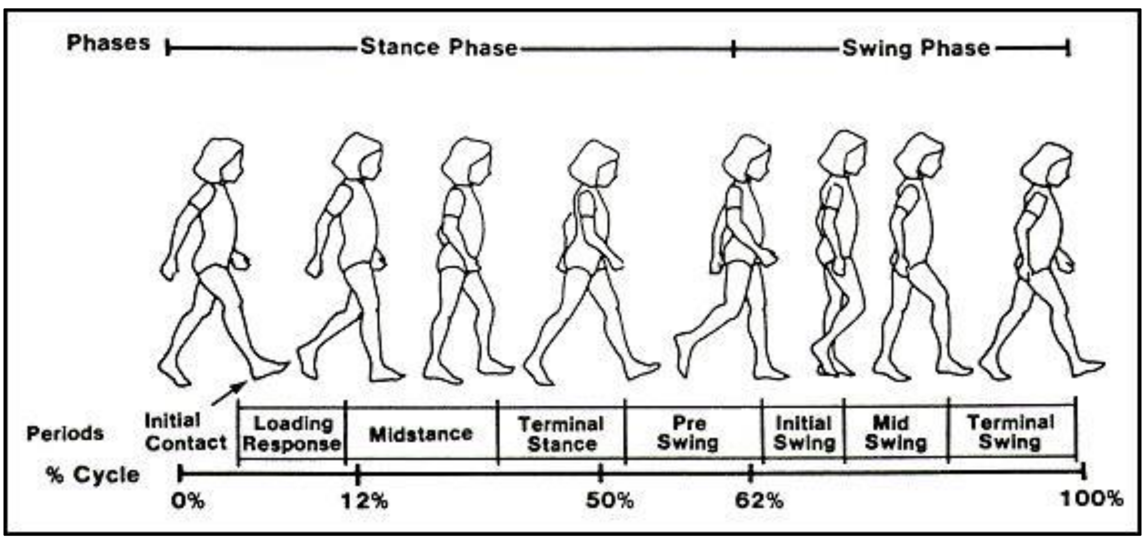

Figure 2. Functional Phases of Gait (Kirtley, 2008).

Normal walking is typically associated with a 60% stance phase and 40% swing phase for each limb, though this relationship is dependent on velocity. Specifically, as walking velocity increases, there is an increased proportionality of the swing phase, with the stance phase becoming shorter. Perry (1992) has reported that the initial double stance period accounts for up to 10% of the gait, or about one-sixth of the overall stance phase (Table 1). Approximately 40% of the gait cycle, or two-thirds of the stance phase, is spent in single-limb support. Another 10%

of the gait cycle, or one-sixth of the stance phase, is spent in terminal double stance as the stance leg prepares for limb advancement (40% of the gait cycle).

Table 1
Percentages of stance and swing phases during gait (Perry, 1992).

	Ground Contact	
Stance	60%	
- Initial Double Stance		10%
- Single Limb Support		40%
- Terminal Double Stance		10%
Swing	40%	

As previously cited, the 60% / 40% is a rough estimate and is based upon the normal walking velocity (a function of stride length and cadence). Blanc and colleagues (1999) demonstrated that healthy adults exhibited a stance phase of ~62% and a swing phase of ~38% when walking barefoot at a preferred velocity (values not reported). However, it has previously been reported that there is a curvilinear relationship between phase times and walking velocity (Murray, Drought, & Kory, 1964). Specifically, as walking velocity increases, there is a decrease in the initial and terminal double stance phase intervals, yet a concurrent increase in single-limb support. Therefore, as walking velocity increases, an individual begins to spend a greater percentage of time in single limb support, until ultimately the double stance phase is entirely eliminated. The disappearance of a period of double support signifies the transition from walking to running as there is now a period of non-support or a flight phase (Blanc, Balmer, Landis, & Vingerhoets, 1999; Perry, 1992; Whittle, 2003). Finally, as walking velocity decreases, there is an increase in the duration of the stance phase. Conversely, as walking velocity increases there is an increase in the duration of the swing phase.

Additional descriptive terms exist to describe the placement of the lower limbs within a gait cycle (Perry, 1992). The stride length is the distance between two successive placements of the ipsilateral foot and is often supplanted by step length; however, step length refers to the distance between the initial contact of one foot to the initial contact of the contralateral foot. Therefore, stride length is composed of two successive steps. In the absence of a pathological condition, step length has been found to be fairly symmetrical, with each step contributing approximately the same distance to the overall stride length (Perry, 1992; Whittle, 2003). The walking base, also known as step width or base-of-support, is the side-to-side distance between two steps, typically measured from the mid-posterior points of the heel. During normal gait there is an inverse relationship between walking velocity and step width, with increasing velocity associated with a decreasing step width.

Division of Gait Cycle using Functional Significance of Events

While the aforementioned divisional components provide a basis for understanding the gait cycle, researchers have incorporated additional divisions which more readily identify the functional demands of each phase of gait (Perry, 1992; Smidt, 1990; Whittle, 2003). This system has resulted in the identification of eight distinct sub-phases for each stride (Figures 2 & 3). Retracing the previous steps, the gait cycle can be divided into stance and swing phases. It has been further divided into periods of initial double stance, single limb stance, terminal double stance, and swing phase. However, these limited descriptions do little to explain the functional significance and contribution of the pelvic and lower extremity joints to the overall movement. Hence, the utilization of analyzing walking patterns through sub-phases aids in understanding the contribution of a particular phase to the overall movement (Perry, 1992). Furthermore, this approach aids in understanding pathology during gait. A posture or movement appropriate for one phase would create dysfunction in another phase. Therefore, based on the task demands of gait cycle, the stance phase is sub-divided into weight acceptance and single limb support phases. Likewise, the purpose of the swing phase is for limb advancement.

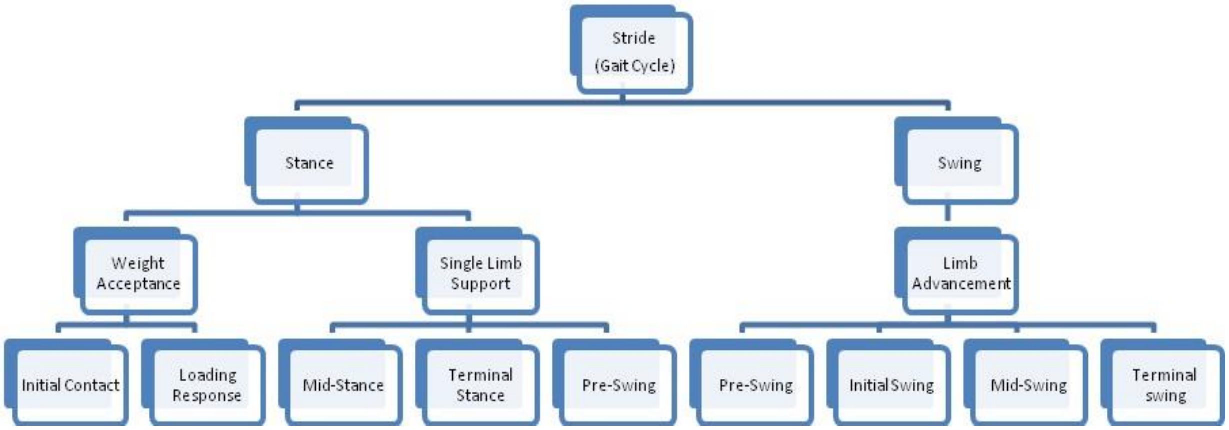

Figure 3. Gait Cycle Divisions (Perry, 1992).

The onset of the stance phase signifies weight acceptance, which includes two additional phases: initial contact and loading response. Single limb support is the next component which is made up of the mid-stance, terminal stance, and pre-swing phases. The final component of the gait cycle is limb advancement, made-up of pre-swing, initial swing, mid-swing, and terminal swing phases. Pre-swing has been determined to be a component of the single limb and limb advancement phases due to the importance of this component and because postural changes begin during the stance phase to meet the high demands of limb advancement (Figure 3).

Weight acceptance is often considered the most demanding task of the gait cycle, coordinating movements for shock absorption, initiating limb stability, and maintenance of forward progression. Furthermore, the weight acceptance task is involved with the initial contact and loading responses phases. Initial contact, accounting for 0-2% of the gait cycle, begins the instant of foot contact. Joint positioning during this phase is a determining factor of the limb's loading response (Perry, 1992). Another purpose of this phase is to position the lower limb and provide a starting point for the stance phase. This allows for the maintenance of the forward momentum of the body, often referred to as the heel rocker. The second phase is the loading response, accounting for 0-10% of the gait cycle (Perry, 1992; Whittle, 2003). This phase corresponds to the initial double stance period, beginning with initial contact and concluding at contralateral toe-off. The loading phase is multi-functional serving the purposes of shock absorption, establishing stability of the limb for weight-bearing, and preserving the forward progression of the body.

The task of single limb support involves the functional phases of mid-stance, terminal stance, and pre-swing. The purpose of the single limb support task is to support the body with the stance leg while the contralateral limb advances to the next site of foot contact. Mid-stance accounts for approximately 20% of the overall gait cycle, or the time period from 10 – 30% of the total gait cycle interval (Perry, 1992; Whittle, 2003). This phase begins at contralateral toe-off (e.g. contralateral swing), continuing until the body is situated over the forefoot. While the mid-stance phase assists in maintaining the forward momentum of the body, it also aids in limb and trunk stability. Terminal stance, accounting for another 20% of the overall gait cycle, occurs over the 30-50% interval of the total gait cycle (Perry, 1992; Whittle, 2003). While the primary objective of the terminal stance phase is similar to that of mid-stance (maintain forward progression of the body), the difference is the manner in which the forward progression is maintained. Specifically, during terminal stance, which begins at heel-off and continues until foot strike of the contralateral limb, the center of mass of the body is anterior to the forefoot. Therefore, rotation occurs about the forefoot rocker, aiding in the considerable force production of the plantarflexors (Whittle, 2003).

Limb advancement is the final task of the gait cycle, constituting approximately 50% of the overall gait cycle (Perry, 1992). While pre-swing is formally included in this section, it is in fact a carryover from postural adjustments that begin during stance and is synonymous with the terminal double stance interval. During normal gait, the pre-swing phase is approximately 10% of the overall gait cycle, covering the 50-60% interval of the gait cycle (Perry, 1992; Whittle, 2003). Like terminal double stance, this phase begins at initial contact of the contralateral limb, ending at ipsilateral toe-off. The primary purpose of this phase is to unload the limb so the postural and muscular actions are more readily able to contribute to latter phases. The sixth phase of the gait cycle is initial swing, accounting for one-third of the overall swing period (60-73% of the gait cycle interval) (Perry, 1992; Whittle, 2003). Beginning at ipsilateral toe-off, the initial swing phase ends when the swing foot is in-line with the contralateral stance foot. This phase is synonymous with mid-stance of the contralateral limb. Initial swing serves to aid in clearing the swing foot of the floor or other ground obstacles while advancing the limb from the trailing position. Mid-swing is the seventh phase of the gait cycle, constituting the interval of 73-87% of the total gait cycle (Perry, 1992; Whittle, 2003). Serving the same purposes of the initial swing phase, the mid-swing phase begins when the swinging limb is opposite the stance limb, ending when the lower leg of the swinging segment is perpendicular to the ground. The final phase of the gait cycle, terminal stance, accounts for the remaining 13% of the gait cycle (87-100% of the gait cycle interval) (Perry, 1992; Whittle, 2003). Beginning when the lower leg is vertical to the floor, it terminates upon foot contact of the swinging limb. Therefore, while limb advancement is served during this phase, it must also prepare the limb for weight acceptance at ground contact.

Locomotor and Passenger Units

While much of the previous discussion has focused on the lower body, it could be considered remiss to so simply isolate the task of walking. To address the relative importance of various segments, researchers have divided the body into two functional units that represent distinct movement patterns: the passenger unit and the locomotor unit (Perry, 1992). The head, neck, trunk, and arms comprise the passenger unit, often referred to as HAT. The locomotor unit consists of the lower limbs and pelvic area, including eleven articulations of interest during walking: lumbosacral, bilateral hip, knee, subtalar, and metatarsophalangeal joints. Comprising almost 70% of the total mass of the body, the HAT segments have been collectively referred to as the passenger unit due to the relative passiveness of the segments to the contribution of walking. Specifically, there is a low need for muscular activity beyond maintaining the segments in the chosen posture; however, there may still be considerable movement. For instance, arms swing out of phase with the legs. Therefore, the left leg, left pelvis, and right arm tend to rotate anteriorly while the right leg, right pelvis, and left arm rotate posteriorly. Despite the considerable movement, there appears to be minimal active contribution of the arms to overall gait economy during normal velocity. The primary purpose of the passenger unit is to maintain the balance and posture of the HAT, and center of mass of the body, which aid the lower limbs in maintaining the forward progression and base of support translations.

Six Determinants of Gait

The locomotor unit is responsible for generating propulsive force, maintaining the upright stability of the body, aiding in shock absorption during ground contact, and conserving energy during translation. Controlling the large number of joints within the locomotor unit, 57 muscles must act in a controlled and coordinated manner to ensure successful execution of the multi-segmental limbs. To aid in understanding the relationship between the passenger unit, locomotor unit, and walking economy, Saunders and colleagues (1953) identified the mixture of six motion patterns that are now known as the Six Determinants of Gait. Three of the determinants are related to the pelvic motions (pelvic rotation, contralateral drop, and lateral shift) while three are focused on lower extremity movements (knee flexion during stance, ankle mechanism, and terminal rocker). While it was originally proposed that the primary purpose of the six determinants was to minimize center of mass excursion, and therefore energy expenditure during gait, this hypothesis has been debated in recent years (Croce, Riley, Lelas, & Kerrigan, 2001; Gard & Childress, 1997; Whittle, 2003). Nonetheless, the movements have been found to occur and are of clinical and research interest.

Pelvic rotation is the first determinant of gait and the one of greatest relevance to the present project. Occurring as the swing limb advances through the swing phase, pelvic rotation twists about a vertical axis, thereby placing the hip joint of the swing leg anterior to that of the stance leg (Figure 4) (Saunders, Inman, & Eberhart, 1953; Whittle, 2003). Furthermore, the rotation can also be seen during terminal stance; however, the rotation during terminal stance occurs in the posterior direction. Pelvic rotation aids in accomplishing several tasks. First, pelvic rotation results in a functional increase in limb length as stride length increases. Second, the pelvic rotation assists in moving the hip joints closer to the midline of the body. Finally, pelvic rotation reduces the flexion-extension range of motion, relative to the distal extremity, necessary to achieve a given stride length. Another determinant is the contralateral drop that occurs during the swing phase of gait. Contralateral drop appears similar to a positive Trendelenburg test, but is not pathological in origin, and occurs about an anteroposterior axis when the swing limb lowers approximately 4° during the swing phase (Figure 5). The purpose of the contralateral drop is to aid in reducing the vertical translation of the HAT during stance phase.

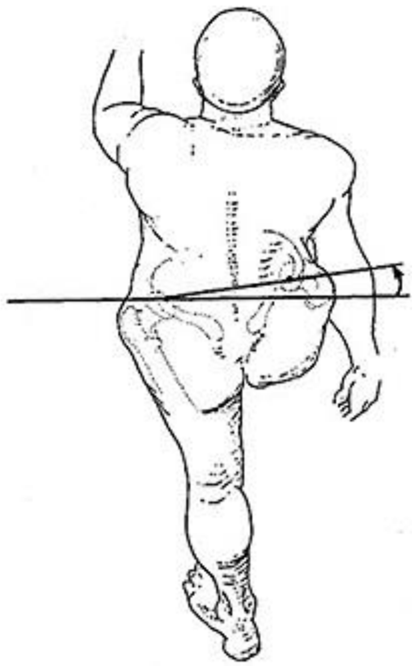

Figure 4. Six Determinants of Gait: Pelvic Rotation (Ayyappa, 1997).

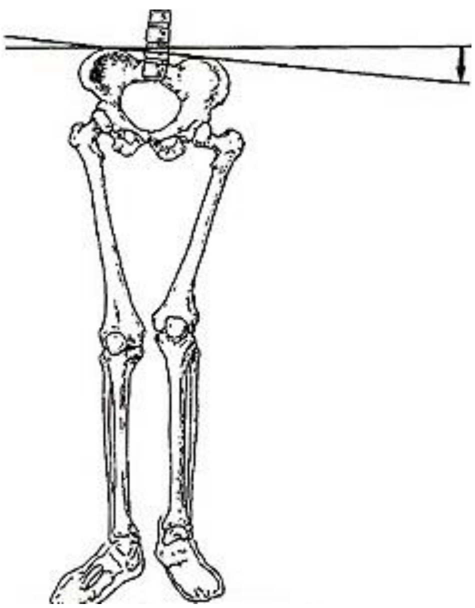

Figure 5. Six Determinants of Gait : Contralateral Drop (Ayyappa, 1997).

Lateral displacement of the pelvis is the third determinant of gait and consists of two factors (Figure 6) (Saunders et al., 1953; Whittle, 2003). First, the natural valgus arrangement between the femur and tibia aids in placing the feet closer to the middle of the body during ground contact. Second, the *genu* abduction that occurs during loading shifts the center of center of mass towards the supporting foot. Another determinant is the terminal or forefoot rocker, which corresponds to heel rise of the stance limb (Figure 7). The purpose of heel rise is to aid in increasing the functional length of the leg, allowing the foot to shift the center of pressure towards the forefoot. The fifth determinant, the ankle mechanisms, acts similar to heel rise. However, instead of increasing the effective length of the limb during terminal stance, the ankle mechanism increase acts to lengthen the limb prior to initial contact. This effect is achieved through dorsiflexion at the talocrural joint, aiding initial contact by extending the foot during walking. The final and sixth determinant of gait is knee flexion during the stance phase (Figure 8). As the hip passes from flexion to extension, concurrent knee flexion acts to reduce the vertical height between the center-of-mass and ground, thereby reducing vertical displacement.

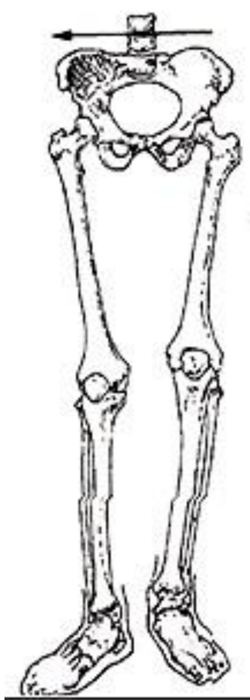

Figure 6. Six Determinants of Gait: Lateral Displacement (Ayyappa, 1997).

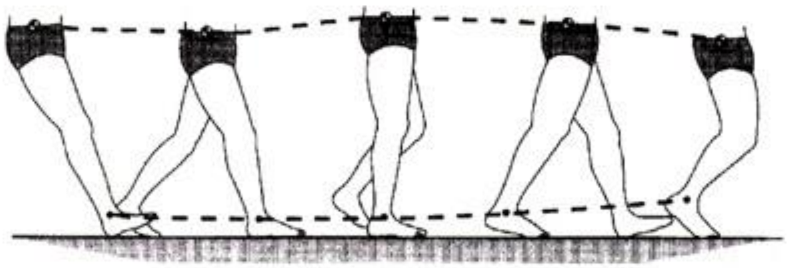

Figure 7. Six Determinants of Gait: Foot & Ankle Motion (Ayyappa, 1997).

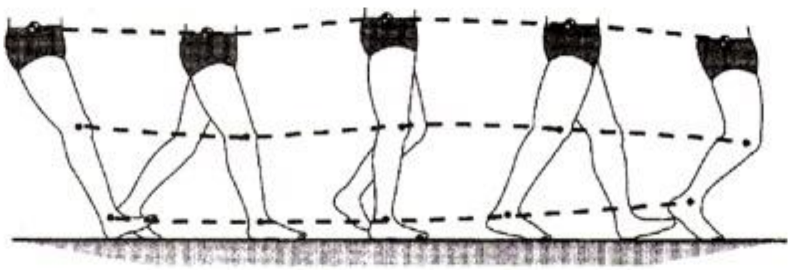

Figure 8. Six Determinants of Gait: Knee Motion (Ayyappa, 1997).

The previous discussion provides a foundation for a clinical and scientific approach to gait analysis. The present project will be concerned with understanding the effects of footwear on relationships between limb advancement, the passenger unit, and the locomotor unit. Of particular interest is the relationship between these variables and lumbopelvic movements during gait. Specifically, this project focused on: (1) stride length from initial contact of the right limb until subsequent initial contact of the ipsilateral limb; (2) the relative timing of leg, pelvic, and thorax movements; (3) the absolute amplitude of leg, pelvic, and thorax movements; and (4) sEMG activity of the erector spinae, gluteus medius, biceps femoris, soleus, and peroneus longus in conjunction with joint kinematics of the lumbopelvic region and lower extremity.

Section 2: Role of the Lumbopelvic Region on Human Locomotion

Walking is the most common form of human locomotion and one of the most researched areas of functional human activity. The predominant focus of most gait research is on the lower extremity, with a secondary focus on the lumbopelvic complex. Historically, much of the lumbopelvic research has relied on descriptive literature of spatiotemporal parameters to explain movements during gait. As discussed in Section 1, Saunders and colleagues (1953) developed a list of gait characteristics, now referred to as the Six Determinants of Gait. Three of the determinants describe specific movements by the pelvis, which aid in the maintenance of forward momentum and energy conservation. Likewise, Ducroquet et al. (1968) introduced the pelvic step concept, stating that from a certain walking velocity and faster, pelvic rotations are dominantly responsible for increases in step length and walking velocity (Ducroquet, Ducroquet, & Ducroquet, 1968). Though Ducroquet and colleagues described the relationship in very general terms, more recent research has suggested that the pelvic step becomes an important contributor to step length at velocities greater than 0.8 ms^{-1} (Bruijn et al., 2008; Wagenaar & Beek, 1992).

Recently, the timing relationship (e.g. relative phase) between thorax and pelvis rotations during walking has received some attention within the literature (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; van Emmerik & Wagenaar, 1996). At all walking velocities, the thorax counter-rotates with respect to the upper legs; however the pelvic timing appears to change from in-phase (synchronous) to out-of-phase (asynchronous) as velocity changes. Remaining unanswered though is the mechanism underlying thorax-pelvis coordination during normal and pathological gait. Similarly, there are conflicting hypotheses as to the role of the vertebral column during gait. Some authors assert spinal motion occurs in response to the action of the lower limbs while other authors have argued for a ‘spinal engine’ generation of gait patterns (C. J. Lamoth, Beek, et al., 2002; Taylor, Goldie, & Evans, 1999; van Emmerik & Wagenaar, 1996). Finally, the author is unaware of any studies examining the effect of footwear on the lumbopelvic complex, with only minimal literature available regarding the footwear effect on activity of trunk musculature during gait. Literature examining the relative phase (RP) relationship of the lumbopelvic complex is reviewed in greater detailed below.

Several authors have examined relative phase differences during gait in healthy populations, with much of the research originating from only a handful of research laboratories. In one of the earlier papers of interest, van Emmerik and Wagenaar (1996) published a paper aimed at understanding the effects of walking on trunk motion. Seven young, male participants walked on a treadmill while gait kinematics were collected utilizing a motion capture (LED) set, operating at 60 Hz, and attached at the T6 level, pelvis, and a single upper leg. During van Emmerik and Wagennar's (1996) study, participants walked at a pre-determined velocity, beginning at 0.3 ms^{-1} and ascending to 1.3 ms^{-1} in 0.1 ms^{-1} intervals before decreasing back to the original velocity. Furthermore, participants walked barefoot for the duration of the protocol. Not surprisingly, the authors found that stride length exhibited a significant increase as walking velocity increased. Interestingly though, the descending portion of the protocol exhibited a hysteresis effect, with a considerably larger stride length than during the ascending portion. Similar effects were found for stride frequency, however, a significant hysteresis effect was found with lower stride frequencies during the descending portion (van Emmerik & Wagenaar, 1996). Likewise, while total range of motion (ROM) of pelvic rotation was greater during higher walking velocities, there was also greater pelvic rotation ROM during the descending portion of the protocol as compared to the ascending. Similar significant effects were found for thorax rotation. The authors reported that thorax-pelvis RP increased from approximately 25° at 0.3 ms^{-1} to 110° at 1.3 ms^{-1}. A similar decrease was witnessed during the descending portion of the protocol, however, no significant hysteresis effect was found. Together, the results support the hypothesis that thorax-pelvis rotations transition from more in-phase to more out-of-phase as walking velocity increases. Remaining unanswered though is the mechanism responsible for these changes (e.g. stride length, cadence).

A series of papers were published in 1997 by Crosbie et al. that examined age and gender effects on spinal, pelvic, and hip kinematics during gait. Unlike the aforementioned studies, these studies utilized a rather large sample (n=108), composed equally of each gender, and from a considerable age range (20 to 82 years of age). Also, participants completed all trials barefoot. The authors relied on the use of more traditional methodology (e.g. passive motion capture system) which included thirteen retroreflective markers on the dorsal surface of the thorax and lower extremity. Peak lateral trunk flexion was found to occur at 15% and 65% of the total gait cycle, corresponding to early swing phase. During this period, thorax and lumbar regions exhibited a displacement towards the stance leg while the pelvis shifted towards the swing leg. The authors also reported that the pelvis and trunk returned to a neutral posture during late swing or early double support. The authors noted a significant posterior pelvic girdle tilt at initial contact, followed by an anterior pelvic girdle rotation during the initial period of ground contact, before displaying a progressive posterior pelvic girdle tilt before the next foot strike. Likewise, maximum lumbar flexion was found to occur at heel strike, with a return to neutral during single limb support, before exhibiting flexion prior to contralateral foot strike. Finally, the authors found that lumbar spine motion tended to follow pelvic movements, though exhibited a slight phase lag.

The second paper by Crosbie and colleagues (1997) utilized similar methodology as the initial study; however, the primary purpose was to examine the age, gender, and velocity effects of walking on spinal kinematics. Participants performed overground walking at two, randomized velocities: self-selected and faster than self-selected. Participants were asked to walk at a normal, comfortable pace for the former. During the latter condition, participants were instructed to adopt a gait pattern as if they were hurrying for an appointment. Height normalized velocity showed no difference between groups Senior Male (SM ≥ 50 years of age), Senior Female (SF ≥50 years of age), Junior Male (JM < 50 years of age), Junior Female (JF < 50 years of age) at a normal, freely chosen velocity; however, during a faster condition there were significant differences in velocity between all groups except JM and JF. Instructions during the fast condition consisted of asking participants to choose a velocity which was comfortable, but similar to that which would be adopted if hurrying to a meeting. In addition, there were gender differences for height normalized step length during the free walking conditions, as was age, with seniors displaying significantly shorter strides. Similar trends were found for the faster condition as in the free walking condition; however, SF and SM displayed a significant gender effect with SM displaying a significantly longer height normalized stride length. With regard to pelvic and lumbar motion, similar motion patterns were exhibited by all participants at both speeds. During the free-walking condition, pelvis motion appeared to exhibit a period of stabilization, yet during the faster condition, the pelvis and trunk appeared in a constant state of motion. Overall, similar pelvic girdle rotations were observed between genders as in the earlier study, however, the amplitude of pelvic girdle movements increased with a slight offset in timing of events noted. Specifically, while positive pelvic girdle tilt still coincided with single limb support, posterior pelvic tilt began occurring prior to mid-stance. The authors also observed significant age effects on the amplitude of displacement for the thorax, lumbar spine, and pelvis; however the authors attributed this to reduced walking velocity of the SM and SF. This final conclusion of the authors was unusual given the lack of significant difference between height-normalized velocities for the free-walking conditions.

The final study by Crosbie and Vachalathiti (1997) examined the synchronicity of the pelvic and hip kinematics during walking. As before, the authors utilized identical participant samples and similar methodology for data collection. Also included in this study were the two walking velocities previously described: self-selected and faster than self-selected. The authors found that males tended to demonstrate significantly greater vertical displacement at both velocities. Conversely, females demonstrated significantly greater pelvic rotation. Interestingly, the authors utilized separate multiple regressions to conclude that changes in step length contribute to larger changes in hip flexion amplitude in females than males. Furthermore, the authors assert that as walking velocity increased, a pseudo-locking of phases occurred, thereby concluding the pelvic contribution is of minimal consequence to normal gait. While noteworthy, more recent findings suggest this hypothesis is incorrect (Y. Huang et al., 2010; Y. P. Huang et al., 2011; C. J. Lamoth, Beek, et al., 2002; C. J. Lamoth, Meijer, et al., 2002; C. J. C. Lamoth et al., 2006).

In 1999, Taylor and colleagues examined the effect of walking velocity on pelvic and lumbar spine kinematics. The goal of this study was to better understand the effects of slower walking velocities on the spine and pelvis so as to better understand changes in pathological gait. Specifically, since reduced walking velocity is often witnessed in individuals with low back or pelvic girdle pain it was hoped to develop an understanding of lumbopelvic movements during slower walking speeds. During treadmill walking, PEAK 3-D motion measurement was utilized to record 27 participants that were randomly allocated to one of two walking velocities: self-selected or 60% of self-selected overground walking velocity. During the trials, the speed of the treadmill was matched to the previously recorded overground walking velocity of that participant. The authors found a significantly lower pelvic list, or lateral pelvic girdle rotation and transverse pelvic girdle rotation in the slow walking group, but no significant differences in pelvic tilt. The authors also found significantly lower lumbar lateral flexion relative to the pelvis in the slow walking group, yet significance was not reached for lumbar rotation or flexion-extension, contradicting earlier research (Stokes, Andersson, & Forssberg, 1989). Interestingly, the authors noted that when the frame of reference was changed from the pelvis to a global reference, the significant differences for lumbar lateral flexion disappeared which was in agreement with earlier findings (Crosbie et al., 1997b; Krebs, Wong, Jevsevar, Riley, & Hodge, 1992; Opila-Correia, 1990). Therefore, the authors asserted that spinal movements should be defined relative to the pelvis when studying underlying intervertebral movement during gait.

Lamoth, Beek, and Meijer (2002) published a similar paper examining thorax-pelvic coordination in the transverse plane during gait. Ten participants took part in the study, conducted with similar methodology to van Emmerick and Wagenaar (1996), however, unlike earlier studies, Lamoth and colleagues (2002) introduced the use of Discrete Fourier Transform to compare continuous relative phases. Rather than attempting to manipulate stride length or cadence, however, the authors only altered velocity. Beginning at 1.4 kmh^{-1}, walking velocity was increased in 0.8 kmh^{-1} increments until a maximum of 5.4 kmh^{-1} was reached. In more traditional gait units, these values are equal to a beginning velocity of 0.39 ms^{-1} and ending velocity of 1.5 ms^{-1}, with increases in intervals of 0.22 ms^{-1}. The authors found that at lower and intermediate walking velocities, maximal thorax rotation occurred at 85 – 95% of the gait cycle (prior to the secondary foot contact), yet almost always coincided with foot contact at higher walking velocities. Furthermore, the thorax-pelvis RP was relatively in-phase at low walking velocities, implying little counter-rotation, yet shifted to more anti-phase as walking velocity increased. Therefore, the results indicated pelvic-thorax coordination was significantly altered by walking velocity, with increased spinal rotation at higher walking velocities; however, as the effects of stride and cadence were not examined, the application of these findings is limited.

In 2008, Bruijn and colleagues published findings examining the effect of pelvic and thorax transverse plane rotations to total body angular momentum. Similar to Lamoth et al. (2002), Bruijn and colleagues (2008) implemented a Discrete Fourier Transform to examined effects of walking velocity on segmental coordination during gait. Specifically, they recorded the kinematics of nine, healthy male participants walking on a treadmill at nine different velocities, ranging from 0.55 ms^{-1} to 1.44 ms^{-1}. Like previous research, the authors found that the pelvis-thorax coordination shifted from in-phase to out-of-phase as velocity increased. Interestingly, while the pelvis began to shift to a more in-phase state with the femur as velocity increased, the thorax became more out-of-phase with the femur at greater velocities. In determining the angular momentum contributions of individual segments to total body angular momentum, the authors reported that the pelvis and thorax only account for ~10% of total body angular momentum during walking. Conversely, the legs and arms account for almost 90% of the total body angular momentum. The authors therefore argued that, despite the conclusions put forth by Ducroquet and colleagues in 1968, pelvic-thorax coordination is relatively unimportant to the organization of total body angular momentum, indicating a need to develop a better understanding of the pelvic step (Bruijn et al., 2008).

Most recently, Huang and colleagues (2010) published a paper examining the effects of stride length and stride frequency on trunk coordination during walking. Sixteen participants walked on an instrumented treadmill at predetermined velocities (0.5 ms^{-1}, 1.0 ms^{-1}, and 1.5 ms^{-1}) while wearing a custom made light emitting diode (LED) marker set attached at the T6 level, pelvis, and a single upper leg. While walking velocity was controlled by the authors, the manipulation of stride length and cadence could be considered less than desirable. Likewise, information concerning footwear, or lack thereof, was not provided. Stride length, and thereby cadence, was manipulated by asking participants to walk a (1) self-selected step length, (2) larger than self-selected, and (3) smaller than self-selected. Descriptive statistics indicated that the author's instructions were successful in achieving different stride lengths; however, there was a significant step x speed interaction, indicating the manipulation of stride length was less effective at higher velocities. Conversely, research on overground walking has indicated the use of an external metronome as an effective manner in which to manipulate cadence and velocity, thereby altering stride length (Romer et al., 2012; Thaut, Miltner, Lange, Hurt, & Hoemberg, 1999). As described by Huang and colleagues (2010), when the extremes of the movements are in-phase, or maximum left pelvic girdle rotation coincides with maximum left thorax rotation, the relative phase is 0° and is indicative of a synchronous movement of the segments. Conversely, when the extremes of the movements occur at opposite time points, the relative phase equals 180° and is indicative of an out-of-phase movement of segments. Despite the limitations of the former study, results indicated that with increasing step length, thorax-pelvis RP, thorax-leg RP, and rotational amplitudes of the thorax and pelvis became significantly larger. Similar to previous findings by Bruijn et al. (2008), pelvis-leg RP decreased with increasing step length. Huang and colleagues (2010) also found that there were significant increases in thorax-leg RP, thorax-pelvis RP, and spinal rotation as walking velocity increased. The results of the study suggest that during treadmill walking, increases in thorax-pelvis RP, decreased pelvis-leg RP, and the amplitude of pelvic, thorax, and spinal rotations increased as a result of increased stride length rather than cadence.

Because the product of stride length and cadence determine walking velocity, the aforementioned results, with an emphasis on those by Huang and colleagues (2010), provide an interesting basis for a hypothesis of changes in barefoot vs. shod gait. As barefoot locomotion is characterized by a reduced stride length and higher cadence relative to shod walking, it could be hypothesized that individuals will display more in-phase thorax-pelvic RP, decreased spinal rotation, and reduced rotational amplitudes during barefoot gait relative to shod gait. Furthermore, tactile sensitivity and feedback has been previously shown to be an important component in the management of gait kinematics; however, plantar sensation appears important in the managing of onset and magnitude of muscle activation (Bird et al., 2003; Chen, Nigg, Hulliger, & de Koning, 1995; Kelleher, Spence, Solomonidis, & Apatsidis, 2010; Kurz & Stergiou, 2003; Murley et al., 2009; Nurse et al., 2005; Nurse & Nigg, 2001; Scott, Murley, & Wickham, 2012). Though some researchers have found an increased physiological and mechanical economy during shod gait, the primary foci of these papers have been on changes in the lower leg (Divert et al., 2005; Divert et al., 2008; Lieberman et al., 2010). Considering the semi-rigid linkage that exists between the distal lower appendicular skeleton and the proximal lower appendicular and axial skeletons, it would seem imperative to understand the effects of footwear and plantar sensation on trunk and pelvic movements during gait.

In summary, research supports considerable differences in the RP of healthy individuals as velocity increases. Specifically, the thorax-pelvis and thorax-leg appear to rotate in a more asynchronous manner as velocity increases. Similarly, amplitude of segmental rotation (i.e. thorax, pelvis, and spine) appears to increase as velocity increases. Conversely, the thorax-leg RP has previously been reported to become more synchronous as velocity increases (Bruijn et al., 2008; Y. Huang et al., 2010). Minimal information is available concerning the independent variables that produce velocity (i.e. stride length, cadence), though research suggests that increased step length may result in more asynchronous movements and greater rotational amplitudes (Y. Huang et al., 2010; Y. P. Huang et al., 2011). While footwear has been shown to significantly alter lower leg gait mechanics, currently unknown is the extent, if any, of the footwear effect on trunk coordination during gait. Research examining phase differences in low back or pelvic girdle pain populations suggest that these individuals tend to self-select segmental movements that result in more synchronous thorax-pelvis RP, indicative of a tenuous link between afferent feedback in gait mechanics and relative phase changes (Y. Huang et al., 2010; C. J. Lamoth, Meijer, et al., 2002; C. J. C. Lamoth et al., 2006; W. Wu et al., 2002; W. Wu et al., 2004; W. H. Wu et al., 2008). The self-selected velocities of pathological populations tend to be lower than the healthy counterparts; however, at comparable velocities, symptomatic individuals tend to display a higher cadence, or greater rotational amplitudes for a given stride length, thereby decreasing the relative phase differences of the thorax and pelvis. Furthermore, RP changes in pathological populations are suggestive of manners in which spatiotemporal variables may be manipulated when the natural constraints of the musculoskeletal system are altered. Remaining to be answered though, is the type of adaptations that arise, if any, by healthy individuals in response to the reduced plantar sensation caused by footwear. The present project intends to examine the effect of the independent manipulation of velocity and cadence on RP changes to better understand mechanisms responsible for altering RP. Therefore, the present project will attempt to differentiate the effect of footwear and augmented plantar feedback on changes in segmental coordination and muscular activation during walking.

Section 3: Influence of Footwear on Human Gait

While the author is unaware of any studies examining the effect of footwear on the coordination of the thorax and pelvis during gait, there is a wealth of information concerning the effect of footwear on human gait. This includes minimalist footwear, barefoot locomotion, sandals, flip-flops, varying height boots, athletic footwear, and numerous types of casual footwear. In fact, this is still a rapidly expanding field of research that regularly garners clinical and industrial interest. A complete review of all the various types of footwear would be beyond the scope of the current project. Instead the focus of this section will be on the differences between barefoot locomotion and shod gait. While the present project intends to focus on differences in gait mechanics during walking, the vast majority of research has focused on differences in running mechanics. Nonetheless, a pool of research is available to examine differences in walking mechanics. Likewise, the similarity between running and walking does allow for one to draw some global conclusions from differences in barefoot and shod during both types of gait.

While the examination of differences in barefoot and shod gait mechanics seems like a relatively elementary comparison, it has recently experienced a considerable upsurge in research and pop-culture interest. This has been partially attributable to the 2009 book by Christopher McDougall, which among other items, discussed the relative dominance in ultra-distance running events from the Native American Tarahumara tribe. A key component of their lifestyle is the relatively minimalist footwear that is used by the tribe members in everyday and athletic life, resembling a sandal with a thin piece of leather material serving as the sole of the shoe. At approximately the same time, Daniel Lieberman, a researcher and professor at Harvard University, was in the process of publishing a series examining the evolutionary role of the human foot (Bramble & Lieberman, 2004; Lieberman et al., 2010). In 2010, Lieberman published a paper examining the gait mechanics of habitually shod versus barefoot runners that received considerable notoriety.

Lieberman and colleagues (2010) found that habitually shod runners exhibited markedly different gait mechanics during running. Specifically, the authors found that habituated shod runners experienced a marked heel stride while habituated barefoot runners displayed a forefoot strike. The former is characterized by an initial and pronounced heel contact, with the foot later transitioning towards footflat. The latter has been defined as an initial contact of the forefoot, followed by ground contact of the heel, resulting in a flat foot. Mechanically, forefoot striking has been found to be a more economical method of running (Divert et al., 2005; Lieberman et al., 2010). Though the mechanism is not completely understood, it is believed that it allows the intrinsic and extrinsic musculature of the foot and lower leg to more successfully utilize the stretch-shortening cycle during a rapid, powerful activity such as running. Forefoot strikers also typically exhibit a decreased stride length and increased cadence during equivalent velocities. Lieberman et al. (2010) noted similar differences between the habitual shod and habitual barefoot runners. Furthermore, the authors noted that barefoot runners displayed significantly greater hip extension and plantarflexion at initial foot contact. Despite the absence of the cushioning effects of footwear, barefoot runners displayed a significantly reduced impact transient (e.g. rate of loading) as compared to heel strikers.

Kristen et al. (1998) reported similar changes in loading patterns as Lieberman et al. (2010), however, the changes were observed in children during walking. The authors examined 30 young children (< 5 years) during barefoot gait and with three different types of shoes. Using 3D kinematic and dual ground reaction force platforms, the authors found that the participants displayed a significant increase in the ground contact time (velocity was not reported), shifting of maximal load weight from the rear foot towards mid-foot, and a cushioning effect during initial impact of ground contact. The authors concluded that footwear does significantly alter loading patterns of gait, even for young children (Kristen et al., 1998).

In an attempt to understand differences in foot motion during locomotion, Morio and colleagues (2009) used three conditions (i.e. barefoot, and sandals with two levels of midsole hardness). Participants performed a series of overground walking and running trials for all three conditions. To control for any footwear effects on either walking or running velocities, gait speed was held constant through the use of a photocell timing gate (Morio, Lake, Gueguen, Rao, & Baly, 2009). The authors reported significant footwear effects, with barefoot locomotion displaying significantly greater forefoot eversion. Furthermore, the rate of forefoot eversion also occurred at a significantly higher rate than during shod locomotion. The authors also reported that the sole of the sandals constrained the torsional and adduction range of motion of the foot, restricting the natural motion of the foot and toe-off strategy. Though not examined in this study, the authors argued that the kinetics of the push off phase would likely be considerably altered due to the imposition of the motion patterns on the foot.

Braunstein et al. (2010) presented another study that examined the effects of footwear on running mechanics. The authors used five different types of running shoes, plus an additional barefoot condition, to compare the gear ratio during running on grass. The gear ratio is defined as the ratio of the moment arm of the ground reaction force to the moment arm of the counteracting musculotendinous unit, an indication of joint loading and mechanical efficiency. Fourteen, endurance trained runners took part in the study which consisted of running along a 20m x 1m running track at a predefined velocity of 4.0 ± 0.2 ms^{-1}. A Kistler ground reaction force plate was mounted flushed with the ground. The authors reported no significant footwear differences in gear ratios or lower extremity moments for any of the five different shoes, however, results indicated that wearing running shoes significantly altered the gear ratios for the ankle and knee joints. Specifically, barefoot running was associated with an increased gear ratio for the ankle during early stance, yet a lower ratio during late stance. The knee gear ratio was lower during midstance in comparison to the shod conditions. The authors concluded that footwear results in a higher mechanical stress on the knee during the stance phase, but an increased mechanical advantage of the plantarflexors during toe-off.

Following the work by Kristen et al. (1998), a 2011 meta-analysis by Wegener and colleagues performed a systematic review of the biomechanical effects of shoes on children during walking and running. The inclusion of a barefoot and at least one shod condition were required for inclusion in the screening process. A total of 11 studies were included, with a sample size ranging from 4 – 498 and a Mean Quality Index of 20/32. The results of the meta-analyses found that relative to barefoot gait, shod walking increased a number of spatiotemporal variables including velocity, step length, step time, base of support, double-support time, stance time, and time to toe-off. Furthermore, considerable changes were determined in joint kinematics with increases in sagittal plane rearfoot range of motion (ROM), maximum ankle plantarflexion, ankle ROM, subtalar ROM, foot lift to maximum plantarflexion, subtalar ROM, knee ROM, and tibialis anterior activity. Similar findings have been reported in adults during barefoot and shod locomotion (Divert et al., 2005; Divert et al., 2008; Kelleher et al., 2010; Lieberman et al., 2010; Nurse et al., 2005). Therefore, it appears that children and adults tend to walk faster, take longer steps, and have increased lower extremity ROM and muscle activity when wearing shoes. A compounding factor that has not adequately been investigated is the interaction between velocity and muscle activity. While increased tibialis anterior activity may have been associated with changes in weight to the distal segment of the foot or alterations in tactile feedback, it could also be associated with the increased velocity associated with shod walking.

In an attempt to answer the question of whether changes in spatiotemporal gait parameters were mechanical in origin, Divert and colleagues (2008) designed a study using ultra-thin diving socks that could be externally loaded with additional mass. The authors recruited twelve, recreationally trained runners for participation in the study which included six, 4-minute running bouts at 3.61 ms^{-1}. During each trial, participants wore only diving socks (unloaded, loaded to 150g, and loaded to 350g), traditional running shoes (loaded to 150g or 350g) or were completely barefoot. Oxygen consumption and three-dimensional ground reaction forces were collected through the entirety of each trial. The authors found a significantly higher stride frequency, anterior-posterior impulse, vertical stiffness, leg stiffness, and mechanical work during the barefoot conditions relative to either shod condition. Furthermore, net efficiency (e.g. metabolic and mechanical components) decreased during the shod conditions relative to the barefoot conditions. Based on the results of this study, the authors concluded that the main role of the shoe is to attenuation foot-ground impact by adding damping material; however the effectiveness of footwear in actually performing this task has previously been called into question with Robbins and Gouw (1990) reporting no change in shock attenuation during running at 3.89 $m*s^{-1}$ on a treadmill. Furthermore, an earlier study found no significant differences in impact forces between standard running shoes and running shoes that contained a 50% increase in heel cushioning (Clarke, Frederick, & Cooper, 1983). Nevertheless, it was hypothesized that changes in the response of the elastic and passive tissues in footwear could be an explanatory reason for reductions in net efficiency (Divert et al., 2008).

In conclusion, footwear has been found to significantly alter the kinematics and kinetics of walking and running gait. While velocities tend to be fairly similar between barefoot and shod locomotion, barefoot gait is associated with a reduced stride length and higher cadence than shod gait (Divert et al., 2008; Larsen, Weidich, & Leboeuf-Yde, 2002; Lieberman et al., 2010; Morio et al., 2009; Wegener et al., 2011). As a result, differences in joint kinematics have been reported, with barefoot walkers exhibiting increased plantarflexion and decreased hip flexion at initial contact (Kristen et al., 1998; Lieberman et al., 2010; Wegener et al., 2011). Shod gait has been associated with important kinetic differences including an increased impact transient, decreased vertical and leg stiffness, and increased anteroposterior ground reaction forces. Furthermore, Braunstein et al. (2010) reported significant differences in the joint kinetics and mechanical efficiency in barefoot and shod locomotion. Therefore, though significant alterations in lower extremity gait mechanics have been previously reported in the literature, the author is unaware of any existing research that have examined the footwear effect on relative phase differences during gait. Despite the significant differences that have been found in lower extremity kinetics and kinematics, peer-reviewed information regarding the footwear effect on pelvic and low back mechanics is relatively sparse. The present project hopes to fill this void by examining the effect of footwear on relative phase changes and pelvic girdle mechanics during walking. Additionally, the present project will examine the efficacy of augmented plantar feedback in altering gait mechanics. A secondary goal of the augmented plantar feedback is to determine if enhanced plantar sensation will cause gait mechanics to shift towards a more natural form of locomotion as has previously been hypothesized by some researchers (Bird et al., 2003; Kelleher et al., 2010; Kurz & Stergiou, 2003).

Section 4: Role of Plantar Sensation and Tactile Feedback in Human Locomotion

In addition to the aforementioned mechanical influences of footwear, there also appears to be a considerable interaction between footwear and gait on the onset and magnitude of muscle activity. Especially interesting are changes in the onset of muscle activity, which may have implications for the loading of active and passive tissues. While the predominant research investigating muscular activation has focused on the lower extremity, there may be important implications for the lumbopelvic musculature as previous research has shown a relationship between lower extremity and core musculature recruitment (Allet et al., 2009; Bird et al., 2003; Kersting, Janshen, Bohm, Morey-Klapsing, & Bruggemann, 2005; Li & Hong, 2007; Mundermann et al., 2001; Murley et al., 2009; Nurse et al., 2005; Wakeling et al., 2003). Delayed activation or altered recruitment of the core musculature may alter the pelvic girdle positioning during loading, thereby affecting the mechanical demands of the passive tissues (Willson et al., 2005). While there has not been extensive research in the area regarding footwear and muscle activation, it has become a recent subfield of this expanding area of research. Furthermore, research investigating muscular activation differences in individuals with reduced tactile sensitivity (e.g. pathological, mechanically, or chemically induced peripheral neuropathy) may provide a window into the effects of decreased sensory feedback on gait mechanics.

Peripheral neuropathy is often found in individuals in advanced stages of diabetes, a metabolic condition that may compound any deduction from findings to healthy populations. However, when controlling for anthropometric factors, researchers have found considerable differences in healthy controls, diabetic individuals without neuropathy, and diabetic individuals with neuropathy (PN). For instance, Allet and colleagues (2009) found that individuals with PN displayed significant differences in a number of kinematic variables (i.e. velocity, cadence, single and double support percentage, cycle time, stride length, and hip angle) compared to healthy individuals. Furthermore, individuals with PN also displayed a significantly higher stride-to-stride variability as compared to healthy individuals, which may be a sign of decreased postural control and increased core instability. Similarly, other researchers have found that individuals with PN exhibited significantly greater center-of-pressure trajectory during a standard balance test (90 seconds, normal stance, eyes closed) as compared to healthy controls (Manor & Li, 2009). Interestingly, other researchers have found that augmented plantar facilitation may increase postural control in populations with balance deficits. For instance, Maki et al. (1999) found a significant decrease in the incidence of extra limb movements and center-of-pressure trajectory in response to postural perturbations when older (65-73 years of age) and younger (23-31 years of age) individuals wore footwear that provided augmented plantar facilitation. Other research has found that providing healthy subjects with augmented plantar feedback resulted in significant changes in plantar pressure distribution during walking (Chen et al., 1995). Specifically, Chen et al. (1995) reported an increased midfoot pressure and decreased pressure in the toe area when participants were provided increased plantar sensory input. From a mechanical standpoint, other research has shown an altered windlass mechanism in individuals with PN, with a greater vertical GRF impulse and impact transient in individuals with PN as compared to healthy controls (D'Ambrogi, Giacomozzi, Macellari, & Uccioli, 2005; Maki et al., 1999). Therefore, it appears that peripheral sensory feedback may not only be an important factor in lower extremity kinematics during gait, but tactile feedback may be an important modulator in thorax and pelvic movements and stability.

A drawback of the aforementioned research is the lack of an investigation into changes in efferent activation as a result of changes in afferent feedback. Conversely, Savelberg and colleagues (2010) measured the electromyography (EMG) of seven lower extremity muscles and found that, regardless of walking velocity, individuals with PN exhibited a significant decrease in the duration of dorsiflexor and knee extensors musculature activity during the stance phase of gait. These results are especially intriguing as there were no reported differences in the joint kinematics or spatiotemporal variables of the pathological populations relative to the healthy controls (Savelberg et al., 2010). While Savelberg et al. (2010) reported no changes in the onset of muscle activity others have investigated the effect of plantar facilitation on changes in musculature timing during the gait cycle. For example, a 2003 study investigated the effect of various forefoot wedge and heel lift designs on the activity of the erector spinae (ES) and gluteus medius (GMed) in healthy individuals (Bird et al., 2003). Interestingly, bilateral heel lifts and all three forefoot wedge designs caused a significant delay in the onset of GMed activity while the heel lifts caused a significantly earlier onset of ES muscle activity. The delay in GMed muscle activity accounted for approximately 2% of the gait cycle while the decrease in the onset of ES muscle activity accounted for approximately 4% of the gait cycle. Furthermore, despite the significant alteration in onset of muscle activity, Bird and colleagues (2003) reported no significant changes in the amplitude of muscle activity. Similarly, Murley et al. (2009) reported significant changes in lower back and lower limb muscle activity in response to an elevated heel height, with the alterations becoming more pronounced as heel height increased. Interestingly, it has also been found that habitual wearers of elevated heel heights tend to display greater fatigability of the peroneus longus and the lateral gastrocnemius (Gefen et al., 2002). Unfortunately, most athletic footwear contains an elevated heel similar in magnitude to those utilized by Bird et al (2003); however, researchers have been unable to quantify the degree of the toe-heel height differential that will alter physiological and mechanical factors in movement (Chiu & Wang, 2007; Gross et al., 1991; Kerrigan, Todd, & Riley, 1998; Kersting et al., 2005; Larsen et al., 2002; Murley et al., 2009). Finally, it has also been reported that many types of orthotics cause a significant increase in the magnitude of the peroneal and anterior tibialis activity, while also increasing the duration of anterior tibialis activity during gait (Murley et al., 2009). It was recently reported that the amplitude and time to peak amplitude for the tibialis anterior, peroneus longus, and medial gastrocnemius were significantly different between barefoot and shod gait, with few differences reported between standard shoe designs (flexible shoe, stability running shoe) (Scott et al., 2012). Specifically, peak tibialis anterior and medial gastrocnemius amplitude was significantly larger, with an earlier occurrence found for the tibialis anterior, during shod gait as compared to barefoot gait. Conversely, peak peroneal amplitude was reduced during shod walking as compared to barefoot walking. Therefore, in addition to the importance of peripheral sensory information in managing lower extremity, pelvic, and torso kinematics, it appears as though footwear and footwear components are capable of producing significant changes in the muscle activity of the aforementioned body segments.

Wakeling and colleagues (2003) investigated whether muscle activity in the lower extremity is used to dampen soft-tissue resonance during gait, which displays considerable peaks at foot contact. By comparing the force transfer from the ground to the soft tissues, as well as the muscle activity of several lower extremity muscles, the authors found that increases in biceps femoris and lateral gastrocnemius activity corresponded to an increased vibration dampening. Therefore, the authors concluded that the evidence supported the hypothesis that muscle activity dampens soft-tissue resonance at foot strike, a potentially important finding for understanding the loading of passive tissues in the lower extremity and low back. Furthermore, the authors assert that shoe design may be used to modify muscle activity and joint loading during walking and running (Wakeling & Liphardt, 2006; Wakeling et al., 2003). To test this hypothesis, Nurse and colleagues (2005) performed a study to determine the effect of textured and smooth inserts on lower extremity muscle activity, limb kinematics, and joint kinetics. Amazingly, the authors found that the textured insert resulted in a significant reduction in the total energy required by the soleus and tibialis anterior during the stance phase, with a reduction in rectus femoris activity approaching, but not reaching, statistical significance. Furthermore, there was also a significant increase in the ankle angle, indicating greater plantarflexion, at heel strike when using the textured insert, indicating the ability of footwear to alter joint kinematics (Nurse et al., 2005). Finally, when wearing the textured insert was compared to the smooth insert, the participants also displayed a significant increase in the time to impact peak, implying a reduced impact transient, and a slight, but significant increase in the active peak force during the first half of the stance phase.

In summary, there is considerable support for the ability of footwear to significantly alter the onset and magnitudes of lower extremity muscle activity, with some research finding significant alterations in pelvic and trunk musculature. Not only would altered onset of muscle activation have considerable effects on joint and segment positioning, as emphasized by the kinetic chain model of movement, but would also considerably alter the loading of the passive tissues (McMullen & Uhl, 2000). As suggested by Wakeling and colleagues (2003), the active tissues of the body are important in the maintenance of vibrational and soft-tissue resonance occurring as a result of ground contact. Though caution should be considered in its application, individuals with peripheral neuropathy have shown altered gait kinematics (i.e. velocity, cadence, single and double support percentage, cycle time, stride length, and hip angle), and have been hypothesized to have a reduced postural control, as compared to healthy controls (Allet et al., 2009). Similarly, other researchers have found that individuals provided with augmented plantar feedback displayed significantly improved postural stability during unpredictable postural perturbations (Maki et al., 1999). Together, the literature base is suggestive of the ability of augmented plantar facilitation to significantly alter lower extremity and torso kinematics and muscular activation during gait; however minimal information exists in this area, with past research focusing on special populations, significant footwear alterations (e.g. heel lifts), or isolated to the lower leg differences. The present project examined the effect of augmented plantar facilitation on gait kinematics and the muscular activation of torso and lower extremity at key gait cycle events (e.g. initial contact, single support). Furthermore, the present project investigated differences in gait kinematics and muscular activation between barefoot, augmented, and shod conditions, an area that the author believes is a relatively novel concept to the gait research community. It is believed that the study approach provided a better understanding of the multi-dimensional link (e.g. physiological, mechanical, and neural) responsible for human movement.

Section 5: Summary of the Literature

Human locomotion is one of the more researched topics in biomechanics literature. During the initial stages of gait, the vast majority of the literature relied on descriptive results to explain the general movement patterns during walking. For example, Saunders et al. (1953) provided an early basis of pelvic and lower extremity movements during gait, collectively referred to as the Six Determinants of Gait. Similarly, Ducroquet and colleagues (1968) were one of the first groups to discuss the relative relationship between trunk, pelvic, and leg movements during gait. Recently, more quantitative research has been performed to understand relative phase changes as a result of changing spatiotemporal variables (Crosbie & Vachalathiti, 1997; Crosbie et al., 1997a, 1997b; Y. Huang et al., 2010; Y. P. Huang et al., 2011; C. J. Lamoth, Beek, et al., 2002; C. J. Lamoth, Meijer, et al., 2002; C. J. C. Lamoth et al., 2006; Taylor, Evans, & Goldie, 2003; Taylor et al., 1999; van Emmerik & Wagenaar, 1996; W. Wu et al., 2002; W. Wu et al., 2004; W. H. Wu et al., 2008). Results by Huang and colleagues (2010) suggest that, during treadmill walking, increases in thorax-pelvis RP, decreased pelvis-leg RP, and the amplitude of pelvic, thorax, and spinal rotations increased as a result of increased stride length rather than cadence. As barefoot locomotion is characterized by a reduced stride length and higher cadence relative to shod walking, it could be hypothesized that individuals will display more in-phase thorax-pelvic RP, decreased spinal rotation, and reduced rotational amplitudes during barefoot gait relative to shod gait.

Considerable research effort has been directed at understanding the effects of footwear on spatiotemporal gait parameters. While the vast majority of the research has been spent investigating various types of footwear, an emerging area of focus is on understanding the underlying factor for differences in barefoot locomotion. Bird and colleagues (2003) reported that bilateral heel lifts and three forefoot wedge designs caused a significant delay in the onset of GMed activity while the heel lifts caused a significantly earlier onset of ES muscle activity. Similarly, Murley et al. (2009) reported significant changes in lower back and lower limb muscle activity in response to an elevated heel height, with the alterations becoming more pronounced as heel height increased.

While the aforementioned authors significantly altered the shoe design, Nurse et al. (2005) took a relatively more passive approach. Specifically, when comparing a textured insole and smooth insole, the authors found that the textured insert resulted in a reduction in the total energy of three lower extremity muscles (tibialis anterior, soleus, rectus femoris). Furthermore, the authors also found a significant increase in the ankle angle, with greater plantarflexion at heel strike when using the textured insert. The importance of tactile sensitivity appears to be an important factor in the modulation of gait mechanics during walking and running. As suggested by Kurz and Stergiou (2003), the reduced plantar sensation and kinematic variability associated with shod locomotion may be a basis for the etiology of many lower extremity injuries. Furthermore, afferent feedback during gait appears to be an important factor in the efferent response of lower extremity muscle activity during gait, altering the loading patterns of active and passive tissues (Bird et al., 2003; Chen et al., 1995; Kelleher et al., 2010; Maki et al., 1999; Murley et al., 2009; Wakeling & Liphardt, 2006; Wakeling et al., 2003).

The current project examined the effects of changes in segmental coordination, gait kinematics, and muscle activation in response to altered spatiotemporal variables. Another purpose of the current study was to examine the effect of footwear on segmental coordination, gait kinematics, and muscle activation. Through this research, an expanded understanding of the effects of footwear on tissue loading during a high volume, cyclic activity such as walking has been developed. The methodology implemented in the present study provided a unique and comprehensive approach aimed at understanding changes in active tissue response and loading of the musculoskeletal system during gait.

Chapter III. Methodology

Walking is the most common form of locomotion for humans, and despite its relative simplicity, relies on the delicate coordination of numerous joints, muscles, and body segments. The pelvis contains a large number of muscles which are responsible for providing the stability from which movements can be initiated and a base from which propulsive forces can be generated. Likewise, the lumbopelvic complex is responsible for numerous important movements in gait, as identified in descriptive ways by Saunders and colleagues (1953) in the Six Determinants of Gait and Ducroquet et al. (1968) with the pelvic step. In addition to these early descriptive studies, researchers have been attempting to quantify and accurately describe the relationship between thorax, lumbar, pelvic, and lower extremity movements during gait. Despite these efforts, the patterns and mechanisms responsible remain partially elusive, demanding a unique and innovative approach to understand the movements of the lumbopelvic complex during gait. Furthermore, footwear is a necessary component of modern life yet has been shown to significantly alter lower extremity kinematics, kinetics, and muscular activity. Despite these findings, there is minimal information concerning the effect of footwear on the mechanics and muscle activity of the lumbopelvic complex. Therefore, the purpose of this study was to better understand the effects of altered spatiotemporal variables and footwear on relative phase differences and muscle activation during walking. Specifically, this study investigated : (1) the effects of velocity and cadence on lumbopelvic rhythm during overground walking; (2) the effects of footwear on lumbopelvic rhythm during gait; (3) the effects of enhanced tactile feedback (i.e. barefoot, textured insole) on the onset of muscle activity at foot strike; (4) the effects of spatiotemporal and footwear variation on the magnitude of lumbar (i.e. erector spinae) and lower extremity muscular activity (e.g. gluteus medius, biceps femoris, soleus, peroneus longus) during the stance phase of gait, and (5) differences in the kinematics and electromyography between males and females. The following chapter presents the methodology that will be used to address the purposes of the present study and includes the following sections: (a) participants, (b) setting, (c) instrumentation, and (d) experimental design and analysis. The research protocol was submitted, reviewed, and approved by the Auburn University Institutional Review Board for Research Involving Human Subjects.

Participants

Fifty-five college-aged males (n=27) and females (n=28) between the ages of 19-35 years were recruited as participants for the study. A power analysis was conducted (effect size = 0.25, alpha = 0.05, and power = 0.80) with G*Power v3.0.10 for Windows, determining that 40 participants would be required to demonstrate significance (Faul, Erdfelder, Lang, & Buchner, 2007). Therefore, 55 participants were chosen for inclusion to ensure adequate power. All participants wore identical clothing consisting of a compression tank-top and compression shorts. Prior to the beginning of data collection, participants were sized for footwear and were provided footwear for use throughout the duration of the protocol. Each participant completed a health screening survey and was excluded if an affirmative response was provided to any question that was deemed by the primary investigator to influence the participant's performance in the study (Appendix A). Exclusion criteria included: (1) any current or recent injury (<1 year) to the lower extremity, pelvis, low back, or trunk, (2) any previous injury/illness that could jeopardize the successful performance of the requisite tasks, (3) any allergies to adhesives or adhesive type products, (4) any known balance deficits or inner ear disturbances. Participants indicated their willingness to participate by signing an Institutional Review Board approved Informed Consent document before data collection commenced (Appendix B).

Setting

All testing and data collection occurred within the Lower Extremity Laboratory at Auburn University (003 Kinesiology). The Lower Extremity Laboratory is a large enclosed laboratory with the necessary equipment to carry out this research.

Instrumentation

Kinematics

A 10 camera Vicon® MX motion analysis system (Vicon®, Los Angeles, CA, USA) was utilized to collect three dimensional kinematics of the lower extremity, lumbopelvic complex, thorax, and upper extremity during each condition. Ten cameras were positioned to record all three cardinal planes within a capture volume large enough to capture one full gait cycle (Figure 9). Furthermore, the sampling frequency of the system was 200 Hz. Spatial positions of body segments were tracked through the use of 42 spherical, retroreflective markers at key anatomical landmarks. The markers used were 12mm in diameter (MKR-6.4, B&L Engineering, Tustin, CA, USA) and attached with double-sided tape (Double Sided Duck ®, ShurTech Brands, Avon, OH, USA). Markers were placed on the participant’s clothing and footwear or, when appropriate, directly on the skin. A modified Plug-In-Gait marker placement protocol was utilized. The modified marker protocol included the addition of medial markers on the elbow, knee, and ankle as well as markers placed bilaterally on the greater trochanter. Table 2 details marker descriptions and applied segments. Euler angles were calculated for the thorax, pelvis and right leg by The MotionMonitor (Innovative Sports Training, Chicago, IL, USA). Rotation sequences for the pelvis were Z Y’ X’’ while the thorax and right leg used an X Z’ Y’’ rotation sequence.

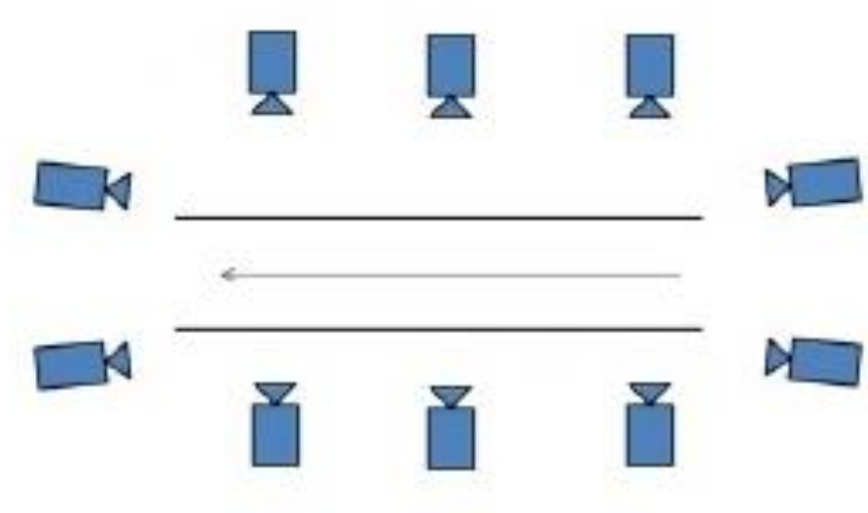

Figure 9. Vicon motion capture camera orientation.

Table 2

Names and positions of markers used in Modified Plug-In-Gait Model.

Marker(s) Name	**Position**	**Segment**
R/LFHD	Right/ Left Front of Head	Head
R/LBHD	Right/ Left Back of Head	Head
CLAV	Sternoclavicular Joint	Torso
STRN	Inferior Sternum	Torso
C7	Seventh Cervical Vertebrae	Spine
T10	Tenth Thoracic Vertebrae	Spine
R/SJO	Right/ Left Acromioclavicular Joint	Shoulder
R/L LATELB	Right/ Left Lateral Aspect of the Humeroulnar Joint	Elbow
R/L MEDELB	Right/ Left Medial Aspect of the Humeroulnar Joint	Elbow
R/LWRA	Right/ Left Radius	Wrist
R/LWRB	Right/ Left Distal End of Ulna	Wrist
R/LFIN	Right/ Left Third Metacarpophalangeal Joint	Knuckle/ Finger
R/LASI	Right/ Left Anterior Superior Iliac Spine	Pelvis
R/LPSI	Right/ Left Posterior Superior Iliac Spine	Pelvis
R/LTRO	Right/ Left Greater Trochanter	Leg
R/LTHI	Right/ Left Lateral Aspect of the Femur	Upper Leg
R/L LATKNE	Right/ Left Lateral Tibiofemoral Joint	Knee
R/L MEDKNE	Right/ Left Medial Tibiofemoral Joint	Knee
R/LTIB	Right/ Left Lateral Aspect of the Fibula	Lower Leg
R/L LATANK	Right/ Left Lateral Malleolus	Ankle
R/L MEDANK	Right/ Left Medial Malleolus	Ankle
R/LTOE	Right/ Left Distal End of Second Metatarsal	Toe/ Foot
R/LHEE	Right/ Left Calcaneus	Heel

The gait cycle for the current study was defined as beginning at initial contact of the right foot and ending at second initial contact of the ipsilateral limb. Other events included in the gait cycle were stance phase and swing phase (Figure 1). Furthermore, the cycle ensured that the entire body was within the capture volume for a full stride length, allowing the evaluation of specific points of interest.

Spatiotemporal Variables

In addition to the kinematic variables collected through the use of motion capture, spatiotemporal variables were collected through the use of an instrumented walkway (GAITRite, CIR Systems Inc., Havertown, PA, USA) (Figure 10). The 4.25m GAITRite mat was placed in the center of the capture volume, with a 3.5m lead-in and 3.5m walkout, for a minimum distance of 11.25m per trial. The pressurized mat collected footfall information at 100 Hz, providing real-time feedback of the velocity and cadence of the participant.

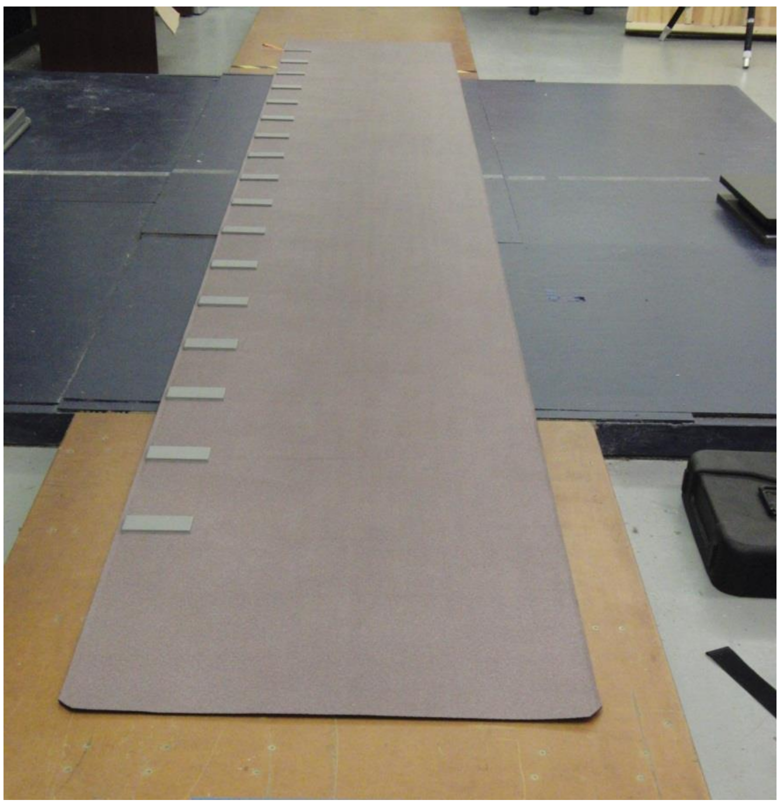

Figure 10. GAITRite Mat.

Surface Electromyography (sEMG)

Thorax (right erector spinae) and lower extremity (bilateral gluteus medius, right biceps femoris, right soleus, and right peroneus longus) muscle activity were collected using six pairs of bipolar Ag-AgCL surface electrodes (Red Dot, 3M. St. Paul, MN, USA). Pre-amplified sEMG leads were connected to a Noraxon Telemyo 2400T V2 wireless transmitter (Noraxon USA Inc., Scottsdale, AZ, USA) (Figure 11) which was connected to the Noraxon Telemyo 2400R – World Wide Telemetry receiver (Noraxon USA Inc., Scottsdale, AZ, USA) (Figure 12) which recorded at a sampling rate of 1500Hz.

Figure 11. Noraxon Telemyo 2400T V2 wireless transmitter.

Figure 12. Noraxon Telemyo 2400R World Wide Telemetry receiver.

Raw EMG signal was amplified with an input impedance of 10 MΩ and a set at 1000X (Knight & Weimar, 2011a). The data reduction procedures for the linear envelope detection of EMG occurred post-collection, including a band-pass filter to remove low-frequency and ambient noise (De Luca, 1997; Kamen & Gabriel, 2010; Knight & Weimar, 2011a, 2011b). The signal was full-wave rectified and root mean squared at 50ms, which involved taking the absolute value of the raw EMG signal, with the high-frequency components being removed due to the filtering (De Luca, 1997; Kamen & Gabriel, 2010; Winter, 2009). Finally, the double threshold method was applied to linear enveloped data to determine onset of muscle activity.

The double threshold method was employed to ensure maximum confidence in onset of muscle activity (De Luca, 1997; Kamen & Gabriel, 2010; Murley et al., 2009). This approach acts as a moving window on the linear envelope detected EMG. The double threshold method begins with calculating the mean and standard deviation of minimal EMG activity (e.g. absence of meaningful muscle contraction). It is recommended that the confidence intervals then be extended to 95% ($\mu \pm 1.96\sigma$) or greater if baseline activity is significant (De Luca, 1997; Kamen & Gabriel, 2010). The corresponding level then signified the initial threshold value needed to signal onset of muscle activity. The second threshold required that once the amplitude of EMG activity reached the initial threshold value, it must stay above the threshold for some critical time period (De Luca, 1997; Kamen & Gabriel, 2010). While, the length of the window for baseline activity has been reported as not overly important to the validity of the final results, the length of the second window is highly dependent on the movement or task being studied. Reported timeframes for the second criterion range from 10ms – 50ms, with higher baseline activities associated with longer criterion windows (Kamen & Gabriel, 2010). In the present study, a window of 25ms was chosen, indicating muscle activity needed to remain above the initial threshold valued for at least 25ms. It is believed the double threshold method reduces the likelihood of identifying a false onset. The inclusion of the both amplitude and temporal aspects would seem to satisfy worries regarding Type I (false onset) or Type II (delay in onset) concerns (De Luca, 1997; Kamen & Gabriel, 2010).

To determine the normalized peak sEMG activity of the six muscles during the gait cycle, muscle activity was trimmed to only include right stance phase (i.e. right foot strike to right toe-off). Once identified, peak activity during the stance phase was normalized relative to the maximal voluntary isometric contraction (MVIC) for the respective muscle.

Footwear

Four footwear conditions were utilized: (a) barefoot, (b) textured insole, (c) minimalist, zero-drop running shoe, and (d) textured insole inserted into the minimalist, zero-drop running shoe. During the barefoot condition, participants were completely barefoot with nothing to interfere with the foot-floor interface. The minimalist, zero-drop running shoe consisted of low-top Adidas AdiPure Adapt (Adidas, Portland OR, USA) (Figure 13). This option was chosen due to the flat style design, which minimizes any differences in the vertical height of the heel and toes. Elevated heels have been previously shown to alter posture, gait kinematics, and muscle activity; however, at what point heel height differentials become significant is not yet understood (Aghazadeh & Lu, 1994; Bird et al., 2003; Gefen et al., 2002; C.-M. Lee et al., 2001; Murley et al., 2009). Therefore, as an elevated heel is common in most mass-production footwear, participants were not allowed to utilize their own footwear. The textured insole consisted of a model utilized by Hatton and colleagues (2011), which has previously been shown to result in significant reductions in postural sway. The textured insert was a 3 mm thick ethylene-vinyl acetate (EVA) insert cut to the foot size from a commercially available outsole material (Evalite Pyramids EVA, 3mm thickness, Algeos Ltd., Dubai, UAE). The material consisted of small pyramidal peaks with center-to-center distances of approximately 2.5mm (Figure 14). The insert was worn against the plantar surface of the foot. During insole only trials, a modified sandal was designed to minimize complications with fitting the insole to each participant while allowing for natural foot motion (Figure 15). The combined effect resulted in minimal movement between the insole and the foot and reduced foot slippage during ground contact. During the combined insole and footwear condition, the insert was inserted directly into the shoe, negating the need for any additional material. All insoles were previously molded by the author to a matching shoe size. Finally, the footwear and textured inserts were only worn during data collection trials.

Figure 13. Footwear Utilized in the Present Study.

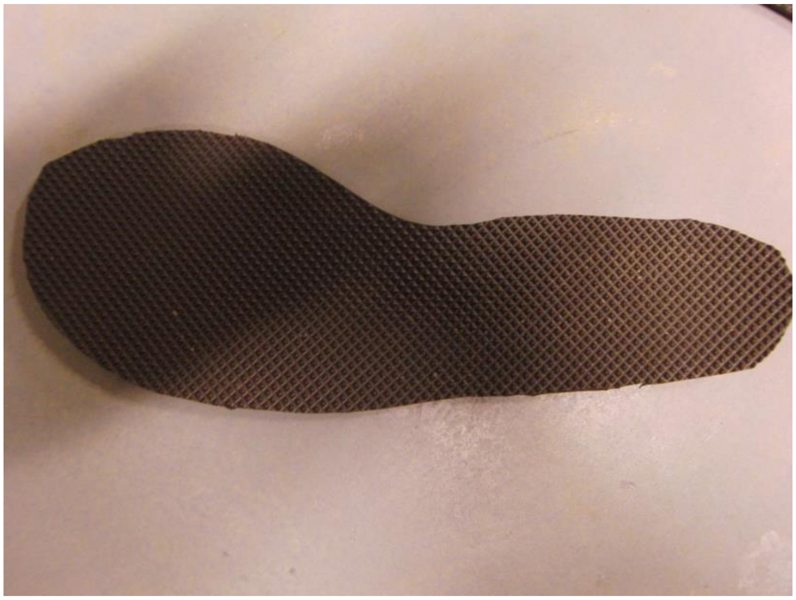

Figure 14. Textured insole utilized in the present study.

Figure 15. Huarache sandal design utilized in the present study for insole only conditions.

Design & Procedures

Participants were asked to report to the Lower Extremity Laboratory for two days of testing, with the second session conducted within 72 hours of the first. Upon arrival on the first day, the participant was provided with an Auburn University Institutional Review Board approved informed consent form and health history questionnaire (Appendices A & B). By signing the informed consent, the participants indicated voluntary participation in the study. The health history questionnaire was used as a screening component to eliminate individuals who have (1) any current or recent injury (<1 year) to the lower extremity, pelvic, low back, or trunk injury, (2) any previous injury that could jeopardize the successful performance of the requisite tasks, (3) any allergies to adhesives or adhesive type products, and/ or (4) any known balance deficits or inner ear disturbances. Upon meeting criteria for participation and indicated consent, participants were asked to try on the footwear to ensure proper shoe size selection. Next, the mass and height of the participant was measured for software inputs and normalization purposes. Additional anthropometric information (i.e. leg length, arch height index) was collected for future normalization purposes. While the aforementioned preparation steps were completed on the first visit, the following procedures were completed during both visits.

All participants were asked to change into a compression outfit and discard any jewelry so as to minimize any potential noise for the motion capture system. The skin of the participant was then prepared for sEMG electrodes. Preparation was conducted in accordance with the recommendations of the Surface Electromyography for the Non-invasive Assessment of Muscles (SENIAM). The skin area around the electrode placement site was shaved, abraded, and cleaned with a 70% alcohol solution to reduce the likelihood of electrical impedance. Electrodes were placed 2-4 cm from the innervation zone, between the myotendinous junction, and parallel to the direction of the muscle fibers for the muscles of interest (De Luca, 1997; Kamen & Gabriel, 2010; Nigg & Herzog, 2007). The lumbar portion of the right erector spinae, bilateral gluteus medius, right lateral hamstrings (e.g. biceps femoris), right soleus, and right peroneus longus were identified by palpation and manual muscle testing. Furthermore, anatomical reference charts were utilized to ensure proper placement. A ground electrode was placed on the body aspect of the tibial tuberosity. A thin, nylon type wrap was placed over the electrodes and limb segment to assist in preventing electrode movement. An MVIC was then performed and used for normalization purposes (Kendall, 2005). An MVIC for the erector spinae was recorded during a prone lumbar extension task. An MVIC for the right and left gluteus medius was recorded during separate, but identical movements (side-laying, hip abduction). An MVIC for the biceps femoris was recorded during prone, knee flexion. The MVIC for the soleus and peroneus longus were recorded simultaneously during a seated, plantarflexion task with knees bent to 90°.

The final step of preparation was to affix 42 markers to the skin and clothing of the participant with double-sided tape. Marker placement was in accordance with the modified Vicon Plug-In-Gait model (Table 2). Next, a static capture was performed with the participants standing in the center of the capture volume in an anatomical neutral position, which is used as a reference point for the software. Marker locations and limb segments were then labeled in accordance with the modified Vicon Plug-In-Gait Model.

Procedures

During the initial visit, ordering of conditions were randomized based on spatiotemporal variables (i.e. altered cadence, altered velocity). During the first day of data collection, only one variable was manipulated, while during the second day the other variable was manipulated. Next, footwear order was randomized (barefoot, shod, textured insole, shoe with textured insole). The initial trials for each condition were Normal walking, followed by a randomization of either 125% or 75% of normal cadence or velocity. Randomization of conditions occurred through the use of a random number generator. After ordering of the conditions was performed, each participant was provided a minimum of three minutes in which to acclimate to the specific footwear condition. During this period, participants were asked to free-walk through the capture volume. After the acclimation period, but before each separate condition, participants were asked to practice the given condition to ensure compliance with velocity or cadence requirements.

Verbal feedback was provided by the author so the participant knew how to adjust to the given parameters. Each condition was repeated until three trials were accurately completed. Rest was provided *pro re nata* between trials or conditions.

Experimental Design & Data Analysis

Relative Phase Calculations

Relative phase calculations were determined based on methodology utilized in previous relative phase literature (Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; Stergiou, 2004). Kinematic data was initially filtered using a dual-pass Butterworth filter. Next, phase portraits for pelvic, thoracic, and leg movements were projected onto a time series of displacement and velocity into a phase plane. The origin for each axis was reset to a midpoint between the extremes of each axis, with values normalized to the relative maximum and minimum values for displacement and velocity. Finally, the phase angle was calculated through a horizontal reference line, originating at the origin, and a projected ray intersecting the time point of interest. The angle between the reference line and ray defined the phase angle for a single segment in time. For the purposes of the present project, segmental movements were defined in relation to the right foot initial contact.

The subtraction of segmental phase angles provided the continuous estimate of relative phase, as illustrated in Equation 1. Relative phase values that are zero indicate the two segments are in-phase or synchronous, whereas relative phase values of 180° indicate asynchronous or out-of-phase movements (Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; Stergiou, 2004). Positive values indicate that the proximal segment is ahead of the distal segment, with negative values indicating the opposite (Stergiou, 2004). Furthermore, the slope of the relative phase provides information concerning segmental velocity, with a positive slope indicating the distal segment is advancing faster in space and a negative slope indicating the proximal segment is advancing faster in space (Stergiou, 2004). Likewise, the local minimum and maximum provide information regarding changes in segmental coordination.

$$\theta_{Relative\ Phase} = \theta_{Distal\ Segment} - \theta_{Proximal\ Segment}$$

Equation 1: Relative Phase Calculation (Y. Huang et al., 2010; C. J. Lamoth, Meijer, et al., 2002).

Finally, as relative phase calculations are circular measures, it was necessary to take the absolute value of the relative phase measure and then calculate linear means and standard deviations (S.D.) using circular statistics, thereby avoiding phase wrapping and providing a single number for analysis (Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; Stergiou, 2004). The mean absolute relative phase (RP_{mean}) was calculated as the summation of the mean curve by averaging the absolute values of the ensemble curve points (Equation 2) (Stergiou, 2004). In Equation 2, *N* is the number of points in the relative phase mean ensemble and $\theta_{relative\ phase}$ is the relative relationship between two segments (Equation 1). Finally, the deviation phase (RP_{SD}) was another method of determining segmental coordination and the stability of the neuromuscular system (Equation 3) (Stergiou, 2004). In the calculation of RP_{SD}, *N* is the number of points in the relative phase mean ensemble and SD is the standard deviation of the mean ensemble at the *i*th point (Stergiou, 2004).

$$RP_{Mean} = \sum_{i=1}^{N} \frac{|\theta_{Relative\ phase}|}{N}$$

Equation 2: Mean Absolute Relative Phase (Stergiou, 2004).

$$RP_{SD} = \frac{\sum_{i=1}^{N} |SD_i|}{N}$$

Equation 3: Relative Phase Standard Deviation (Stergiou, 2004).

Statistical Analyses

All statistical analyses were conducted using SPSS software (version 18.0, SPSS Inc. Chicago, IL, USA) and an alpha level of statistical significance set *apriori* at $\alpha \leq 0.05$. All data was stored in Excel databases and imported into SPSS for analysis. To investigate the effects of footwear on gait kinematics during Normal walking, the kinematic data was analyzed using a two-way repeated measure MANOVA. The independent variables were footwear types, with four levels (barefoot, shod, textured insole, shod with textured insole), and gender with two levels. The dependent variables were stride length (SL), RP_{mean} for the segments of interests (thorax-pelvis, thorax-leg, pelvis-leg), RP_{SD} for the segments of interests (thorax-pelvis, thorax-leg, pelvis-leg), and rotational amplitudes (thorax, pelvis, leg).

To investigate the effects of walking velocity on gait kinematics during shod walking, the kinematic data was analyzed with a two-way repeated measure MANOVA. The independent variables were walking velocity, with three levels (Normal, 125% of Normal velocity, and 75% of Normal velocity), and gender with two levels. The dependent variables were stride length (SL), cadence (SR), RP_{mean} for the segments of interests (thorax-pelvis, thorax-leg, pelvis-leg), RP_{SD} for the segments of interests (thorax-pelvis, thorax-leg, pelvis-leg), and rotational amplitudes (thorax, pelvis, leg).

To investigate the effects of cadence on gait kinematics during shod walking, the kinematic data was analyzed with a two-way repeated measure MANOVA. The independent variables were walking cadence, with three levels (Normal, 125% of Normal cadence, and 75% of Normal cadence), and gender with two levels. The dependent variables were stride length (SL), velocity, RP_{mean} for the segments of interests (thorax-pelvis, thorax-leg, pelvis-leg), RP_{SD}

for the segments of interests (thorax-pelvis, thorax-leg, pelvis-leg), and rotational amplitudes (thorax, pelvis, leg).

To investigate the effects of footwear on sEMG during Normal walking, the sEMG data was analyzed using a two-way repeated measure MANOVA. The independent variables were footwear types, with four levels (barefoot, shod, textured insole, shod with textured insole), and gender with two levels. The dependent variables were onset of muscular activity for the right erector spinae, right gluteus medius, left gluteus medius, right biceps femoris, right soleus, and right peroneus longus prior to foot strike, as well as the normalized peak sEMG of the aforementioned muscles during the stance phase of gait.

To investigate the effects of walking velocity sEMG during shod walking, the sEMG data was analyzed with a two-way repeated measure MANOVA. The independent variables were walking velocity, with three levels (Normal, 125% of Normal velocity, and 75% of Normal velocity), and gender with two levels. The dependent variables were onset of muscular activity for the right erector spinae, right gluteus medius, left gluteus medius, right biceps femoris, right soleus, and right peroneus longus prior to foot strike, as well as the normalized peak sEMG of the aforementioned muscles during the stance phase of gait.

To investigate the effects of cadence on sEMG during shod walking, the sEMG data was analyzed with a two-way repeated measure MANOVA. The independent variables were walking cadence, with three levels (Normal, 125% of Normal cadence, and 75% of Normal cadence), and gender with two levels. The dependent variables were onset of muscular activity for the right erector spinae, right gluteus medius, left gluteus medius, right biceps femoris, right soleus, and right peroneus longus prior to foot strike, as well as the normalized peak sEMG of the aforementioned muscles during the stance phase of gait.

For all statistical analyses, post hoc repeated measure ANOVA's were performed for each dependent variable of the repeated measure MANOVA that demonstrated statistical significance. Follow-up pairwise comparisons were then performed for significant findings from the ANOVA analyses to determine statistical significance between the levels of the independent variables.

Chapter IV. Results

The purpose of this study was to better understand the effects of footwear on relative phase differences and muscle activation during walking. Specifically, this study investigated : (1) the effects of velocity and cadence on lumbopelvic rhythm during overground walking; (2) the effects of footwear on lumbopelvic rhythm during gait; (3) the effect of enhanced tactile feedback (i.e. barefoot, textured insole) on the onset of muscle activity at foot strike; (4) the effect of spatiotemporal and footwear variation on the magnitude of lumbar (i.e. erector spinae) and lower extremity muscular activity (e.g. gluteus medius, biceps femoris, soleus, peroneus longus) during the stance phase of gait, and (5) if kinematics or muscle activity patterns display significant gender differences. While human walking has been extensively studied in previous research, the relationship between trunk, pelvic, and leg movements are not yet fully understood (Crosbie & Vachalathiti, 1997; Crosbie et al., 1997a, 1997b; Y. Huang et al., 2010; Y. P. Huang et al., 2011; C. J. Lamoth, Beek, et al., 2002; C. J. Lamoth, Meijer, et al., 2002; C. J. C. Lamoth et al., 2006; Taylor et al., 2003; Taylor et al., 1999; van Emmerik & Wagenaar, 1996; W. Wu et al., 2002; W. Wu et al., 2004; W. H. Wu et al., 2008). Likewise, while footwear has previously been shown to alter lower extremity joint kinematics and muscle activity, the extent to which tactile feedback is important in modulating walking is an area of emerging interest to researchers and clinicians. (Bird et al., 2003; Chen et al., 1995; Kelleher et al., 2010; Maki et al., 1999; Murley et al., 2009; Wakeling & Liphardt, 2006; Wakeling et al., 2003). The following chapter presents results of the current study in the following sections: (1) participant demographics; (2) the effect of walking velocity and gender on relative phase differences and muscle activation patterns; (3) the effect of cadence and gender on relative phase differences and muscle activation; and (4) the effect of footwear and gender on relative phase differences and muscle activation patterns. To eliminate the potential of confounded results, data from the shod condition only was utilized to examine velocity and cadence effects. To analyze the footwear effects on kinematics and muscle activity, only the normal (e.g. self-selected) condition was included in the analyses.

Section 1: Participant Demographics

One-hundred-twelve participants volunteered for participation in the current study, with fifty-nine participants meeting initial qualifications (**Appendix A)**. Upon completion of the initial screening, each participant signed an Informed Consent document previously approved by the Auburn University Institutional Review Board for Human Subjects (**Appendix B**). Of the fifty-nine participants, four participants did not return for either testing session, one did not complete both testing sessions, and four were excluded due to erroneous data; therefore, fifty participants (25 men, 25 women) completed both testing sessions and were included in data analysis. Demographic information for the participants' is included in Table 3.

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Table 3
Participant Demographics

All Participants	**Mean**	**Standard Deviation**
Age (years)	23.1	2.8
Height (m)	1.72	0.11
Mass (kg)	74.3	17.8
Leg Length (cm)	86.2	5.7
Females	**Mean**	**Standard Deviation**
Age (years)	23.6	3.3
Height (m)	1.64	0.06
Mass (kg)	64.7	17.6
Leg Length (cm)	82.8	4.4
Males	**Mean**	**Standard Deviation**
Age (years)	23.0	2.5
Height (m)	1.81	0.07
Mass (kg)	83.7	11.7
Leg Length (cm)	89.4	4.8

Section 2: Walking Velocity Effects

To understand the effects of walking velocity on relative phase differences during overground walking, velocity was manipulated on three levels (Normal, 125% of Normal, and 75% of Normal) during one test session. Normal was self-selected by the participant. To determine differences between conditions, a 3 (condition) x 2 (gender) repeated measures MANOVA was performed for eleven dependent kinematic variables (Table 4). A significant ($p<0.001$) Box's M test was found, therefore, the application of the more robust Pillai's Trace multivariate test was utilized for interpretation of output statistics.

Table 4 – Kinematic Variables: Velocity Effects

Independent Variables	**Dependent Variables**
Gender	Relative Phase $_{mean}$
- Male	- Trunk-Pelvis (TP $_{mean}$)
- Female	- Trunk-Right Leg (TL $_{mean}$)
Condition	- Trunk-Left Leg (PL $_{mean}$)
- Normal	Relative Phase $_{SD}$
- 125% of Normal	- Trunk-Pelvis (TP $_{SD}$)
- 75% of Normal	- Trunk-Right Leg (TL $_{SD}$)
	- Trunk-Left Leg (PL $_{SD}$)
	Euler Angles (at foot strike)
	- Pelvis
	- Trunk
	- Right Leg
	Spatiotemporal Variables
	- Cadence
	- Normalized stride length

The multivariate repeated measures ANOVA with a *p* value set *apriori* at < 0.05 yielded a non-significant Gender*Condition interaction and a non-significant Gender effect; however, significant Condition (Λ_{Pillai} = 0.998, $F(22,27) = 656.053$, $p < 0.001$, $\eta^2 = 0.998$, *Power* = 1.000) differences were found.

Spatiotemporal Variables: Condition Differences

Average walking velocity for all participants was 1.28 ms^{-1}, 1.59 ms^{-1}, and 0.96 ms^{-1} for the Normal, 125%, and 75% conditions, respectively. Follow-up ANOVAs found a significant difference for cadence ($F(2,96) = 965.977$, $p < 0.001$, $\eta^2 = 0.953$, *Power* = 1.000) and normalized stride length ($F(2,96) = 730.041$, $p < 0.001$, $\eta^2 = 0.935$, *Power* = 1.000). Post hoc analyses using the Least Significant Differences (LSD) found a significant difference (p<0.001) between cadence for all three conditions (Figure 16). Cadence was highest during the 125% velocity condition and lowest during the 75% velocity trials. Height normalized stride length was also found to be significantly different (p<0.001) between all conditions (Figure 17). Normalized stride length was longest during the 125% velocity condition and lowest during the 75% velocity trials.

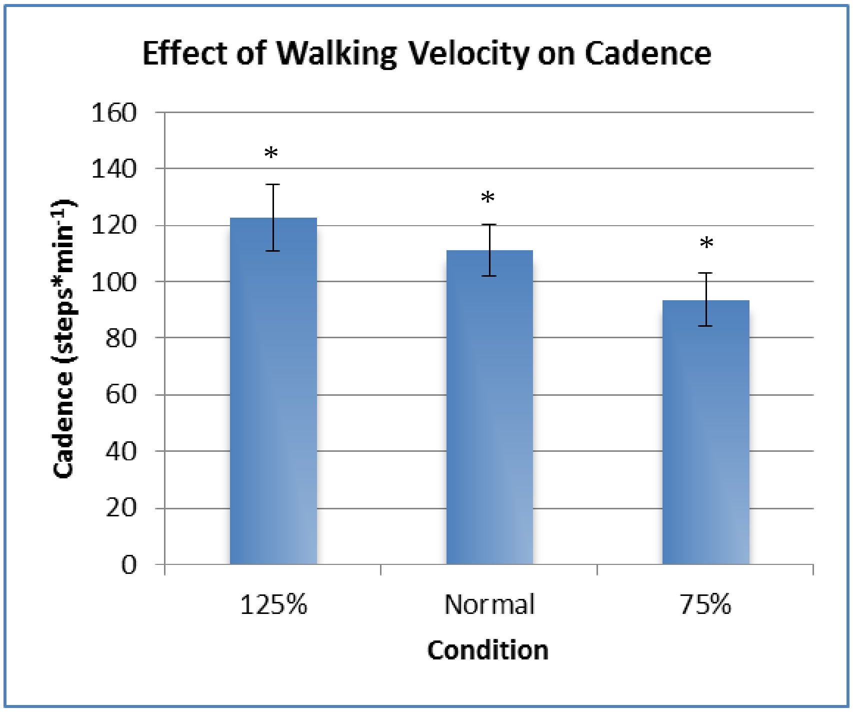

Figure 16 – Effect of walking velocity on cadence. Notations (*) indicate significance at (p<0.05).

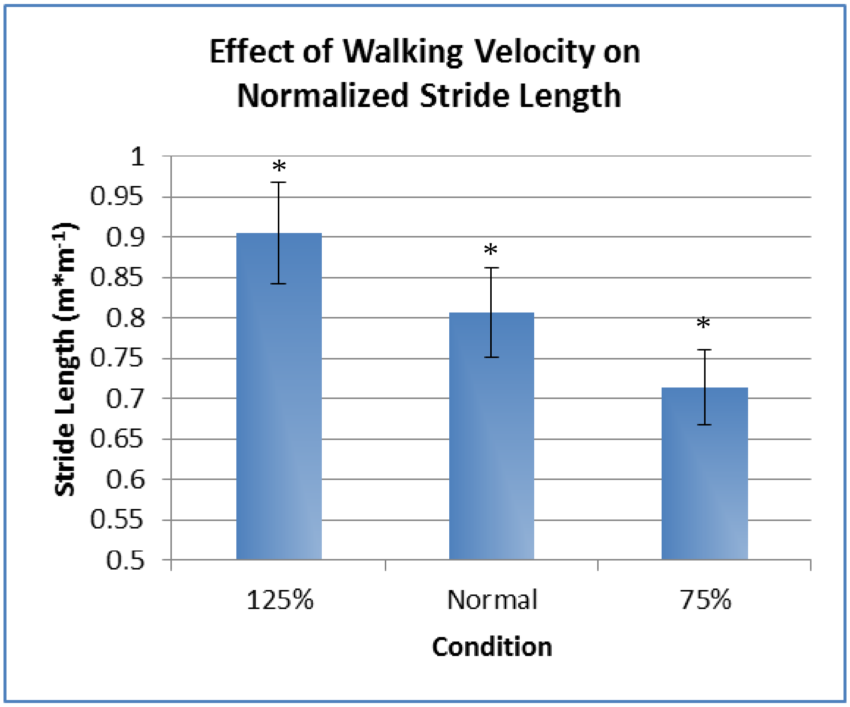

Figure 17 – Effect of walking velocity on normalized stride length. Notations (*) indicate significance at (p<0.05).

Spatiotemporal Variables: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the spatiotemporal variables (Cadence, Stride length) when velocity was altered.

Kinematics

Phase portraits were created as described in Chapter III and calculated using a custom MATLAB program (MATLAB R2013a Student Edition, Mathworks, Natick, MA, USA). Figure 18 provides an example of a phase portrait for the pelvis during a full-stride (right toe-off to right toe-off) as calculated in the present study. The present study determined stride events based upon right foot events only (right foot strike, right toe-off) for all participants. From the phase portrait, phase angles were calculated for the trunk, pelvis, and leg at foot strike for each trial, after which the subtraction of the phase angles was performed according to Equation 1, with the difference equaling the relative phase value for the two segments. Relative phase measures range in value from 0° - 180°, with larger values implying greater asynchronicity, or out-of-phase relationships, between segments. Conversely, smaller values imply a more synchronous, or in-phase relationship between segments. Because relative phase is a circular measure, circular statistics were then applied (Equation 2 & 3) prior to the application of the linear statistics, resulting in three Relative Phase $_{mean}$ measurements (thorax-pelvis (TP $_{mean}$), thorax-leg (TL $_{mean}$), pelvis-leg (PL $_{mean}$)) and three Relative Phase $_{SD}$ measurements for the segments of interests (thorax-pelvis (TP $_{SD}$), thorax-leg (TL $_{SD}$), pelvis-leg (PL $_{SD}$)) in addition to the Euler Angles for the Pelvis, Trunk, and Right Leg at foot strike. The Relative Phase $_{SD}$ measure provides a measure of the amount of variability in the Relative Phase $_{mean}$ measure, or stated in another manner, the amount of variability in the synchronicity between segments.

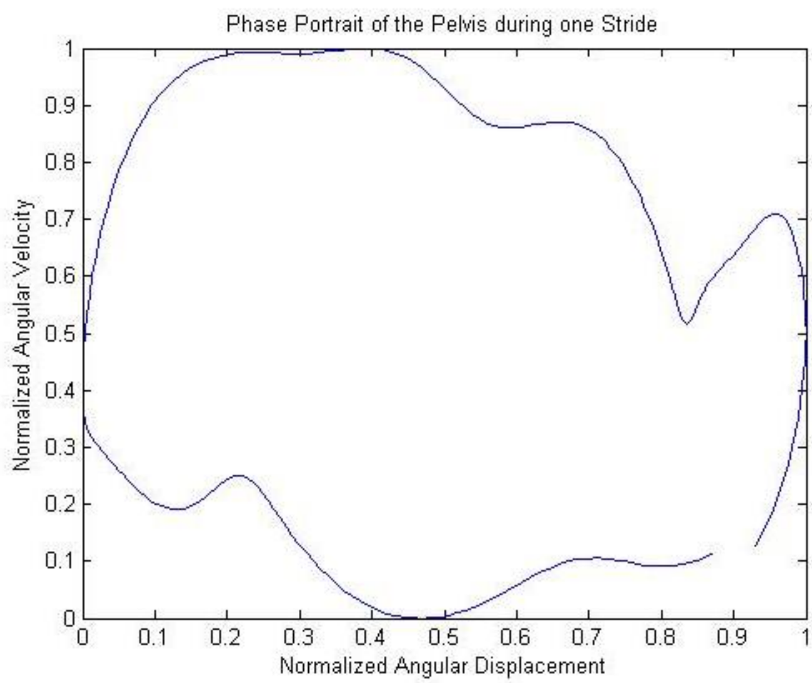

Figure 18 – Example Phase Portrait of the Pelvis during one-stride.

Kinematics: Condition Differences

Trunk-Pelvis $_{mean}$

Follow-up ANOVAs found a significant difference for TP $_{mean}$ ($F(2,96) = 41.079$, $p < 0.001$, $\eta^2 = 0.461$, *Power* = 1.000) with post hoc analyses using the LSD criterion indicating that the 75% velocity condition displayed a significantly greater ($p<0.001$) TP $_{mean}$ than the other conditions, while Normal was also significantly ($p=0.008$) greater than 125% velocity (Figure 19). Results indicate that the pelvis and trunk were most out-of-phase, indicating greater asynchronicity, during the 75% velocity condition, relative to Normal and 125% velocity, and more out-of-phase during the Normal velocity than at 125% velocity. Follow-up ANOVA's revealed no significant velocity effect on TP $_{SD}$, indicating that the relationship between the trunk and pelvis do not become more variable as velocity increases from 75% of Normal to 125% of Normal.

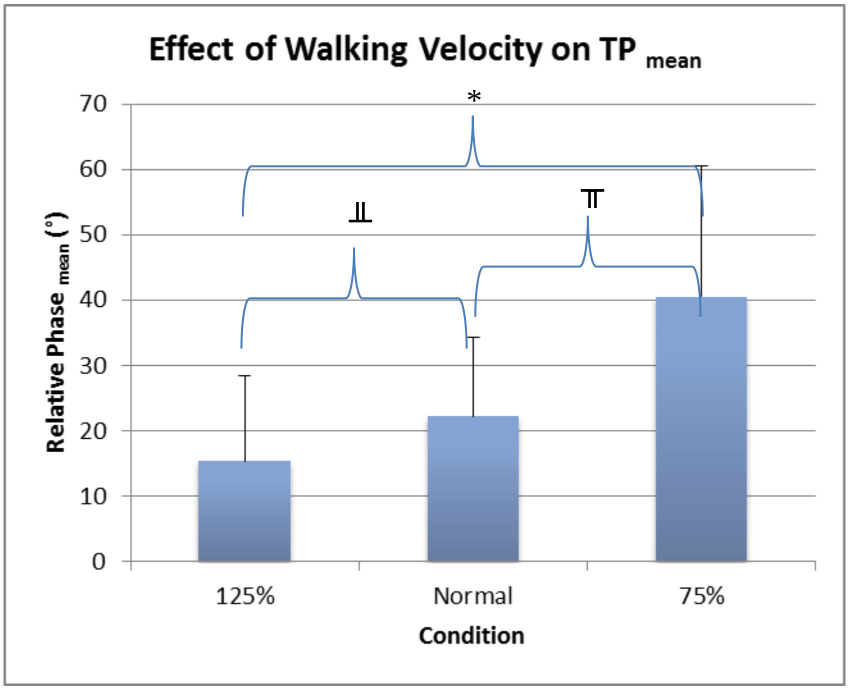

Figure 19 – Effect of walking velocity on TP $_{mean}$. Notations (*╥ ╨) indicate significance at ($p<0.05$).

Pelvis Rotation at Foot Strike

Follow-up ANOVAs found a significant effect of velocity on the magnitude of pelvic rotation ($F(2,96) = 354.773$, $p = 0.009$, $\eta^2 = 0.094$, *Power* = 0.800) with post-hoc analyses using a LSD criterion indicating that the 75% velocity condition was significantly lower than the Normal condition (p=0.004), yet no difference between Normal and 125% velocity or 125% velocity and 75% velocity was noted. Results indicate that participants exhibited the greatest left transverse pelvic rotation at right foot strike during Normal walking while the pelvis was in a relatively neutral position at foot strike during 75% of Normal walking (Figure 20).

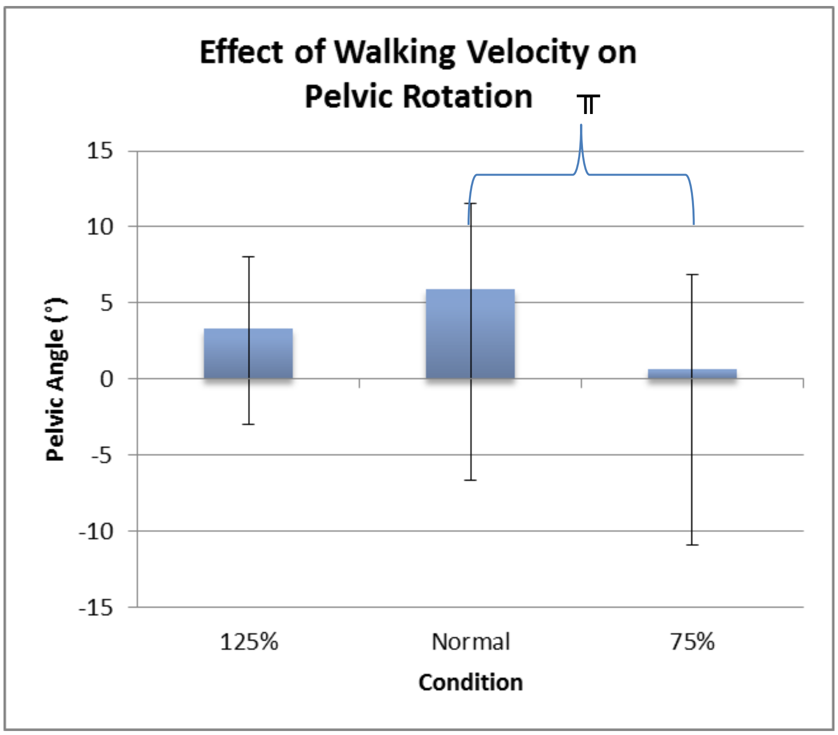

Figure 20 – Effect of walking velocity on pelvic rotation. Notations (╨) indicate significance at (p<0.05).

Trunk Rotation at Foot Strike

Follow-up ANOVAs found a significant effect of cadence on the magnitude of trunk rotation ($F(2,96) = 4.07$, $p = 0.020$, $\eta^2 = 0.078$, *Power* = 0.711) with post-hoc analyses using a

LSD criterion indicating that the 75% velocity condition was significantly lower than 125% (*p*=0.004) and Normal condition (*p*=0.043), yet no difference between Normal and 125%, indicating the trunk was more rotated to the right at foot strike during Normal or 125% velocity walking than during 75% velocity (Figure 21).

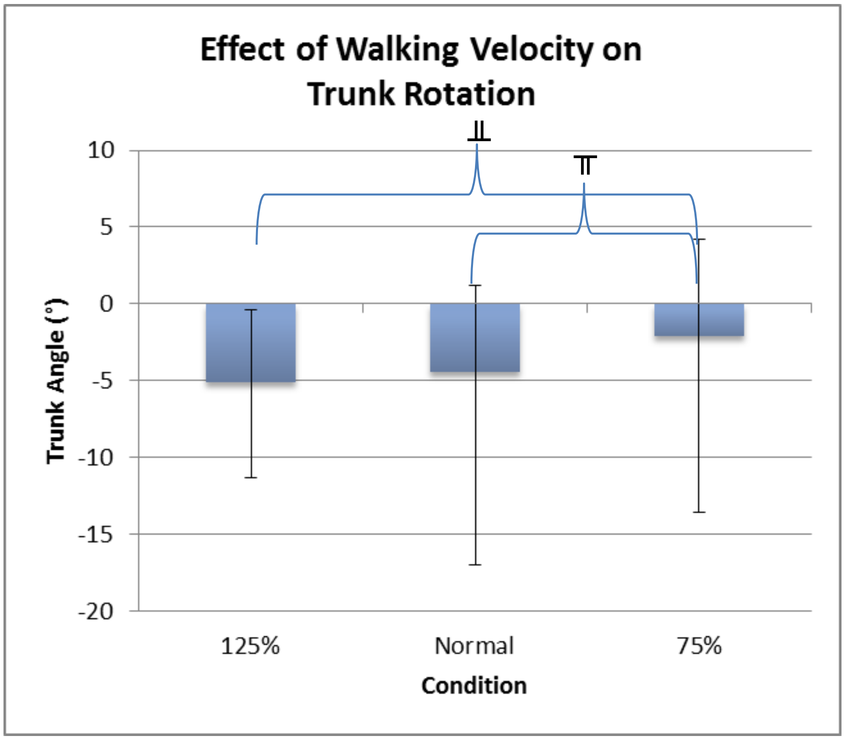

Figure 21 – Effect of walking velocity on trunk rotation. Notations (╥ ╨) indicate significance at (p<0.05).

Non-significant Dependent Kinematic Variables During Varying Velocities

Follow-up ANOVAs found no significant condition effect for the following kinematic variables (Table 5); indicating that changes in velocity did not significantly affect the participants for these dependent variables when velocity was altered at three levels (Normal, 125%, 75%).

Table 5 - Non-significant condition effects for kinematics when velocity was varied.

	F	df	*N*	*p*	η^2	*Power*
TP_{sd}	2.247	2	96	0.111	0.045	0.448
TL_{mean}	2.391	2	96	0.097	0.047	0.472
TL_{sd}	1.289	2	96	0.280	0.026	0.273
PL_{mean}	2.695	2	96	0.073	0.053	0.522
PL_{sd}	2.825	2	96	0.064	0.056	0.543
Leg Rotation	0.496	2	96	0.610	0.010	0.129

Kinematics: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the kinematic dependent variables (TP_{mean}, TP_{sd}, TL_{mean}, TL_{sd}, PL_{mean}, PL_{sd}, θ_{pelvis}, θ_{trunk}, θ_{leg}) when velocity was altered.

Surface Electromyography

To understand the effects of walking velocity on muscle activity during overground walking, electromyography (EMG) was recorded at 1500 HZ (Figure 12) using six pairs of bipolar Ag-AgCL surface electrodes (Trace-Rite, Bio-Detek Inc. Pawtucket, RI, USA), placed longitudinally over the bellies of six muscle (Table 7). To determine onset of muscle activity, a custom MATLAB program (MATLAB R2013a Student Edition, Mathworks, Natick, MA, USA) was written using the double-threshold technique as described in Chapter III, thereby determining the difference in time from onset of muscle activity and right foot strike. Visual inspections of the full-wave rectified signal, in conjunction with kinematic data, were also performed. Normalized peak EMG during the stance phase was also calculated using a custom MATLAB program, with all signals normalized to the maximal voluntary isometric contraction (MVIC) of the respective muscles. Though electrode placement sites were outlined during the initial session, MVIC's were performed at the beginning of each session for normalization purposes. To determine differences between conditions, a 3 (condition) x 2 (gender) repeated measures MANOVA was performed for twelve dependent variables (Table 6). A significant (*p<0.001)* Box's M test was found, therefore, the application of the more robust Pillai's Trace multivariate test was utilized for interpretation of output statistics. The multivariate repeated measures ANOVA with a *p* value set *apriori* at < 0.05 yielded a non-significant Gender*Condition interaction and a non-significant Gender effect; however, a significant Condition ($\Lambda_{Pillai} = 0.932$, $F(24,24) = 13.645$, $p < 0.001$, $\eta^2 = 0.932$, $Power = 1.000$) effect was found.

Table 6 – sEMG Variables

Independent Variables	**Dependent Variables**
Gender	Peak sEMG during Stance Phase
- Male	- Peroneus longus (PerL peak)
- Female	- Soleus (SOL $_{peak}$)
Condition	- Biceps femoris (BF $_{peak}$)
- Normal	- Right Gluteus medius (rGM $_{peak}$)
- 125% of Normal	- Left Gluteus medius (lGM $_{peak}$)
- 75% of Normal	- Erector spinae (ES $_{peak}$)
	Onset of Muscle Activity at Foot Strike
	- Peroneus longus (PerL onset)
	- Soleus (SOL $_{onset}$)
	- Biceps femoris (BF $_{onset}$)
	- Right Gluteus medius (rGM $_{onset}$)
	- Left Gluteus medius (lGM $_{onset}$)
	- Erector spinae (ES $_{onset}$)

Surface Electromyography: Condition Differences

Peak Peroneus longus

Follow-up ANOVAs found a significant difference for PerL $_{peak}$ ($F(2,94) = 69.815$, $p < 0.001$, $\eta^2 = 0.598$, $Power = 1.000$) during the stance phase. Post hoc analyses using the LSD criterion determined that the PerL $_{peak}$ during the 125% velocity condition was significantly greater than Normal (p<0.001) and 75% (p<0.001). Furthermore, Normal was significantly greater than 75% velocity (p=0.013) indicating that alterations in velocity have a significant effect on peroneus longus activity during the stance phase of walking (Figure 22).

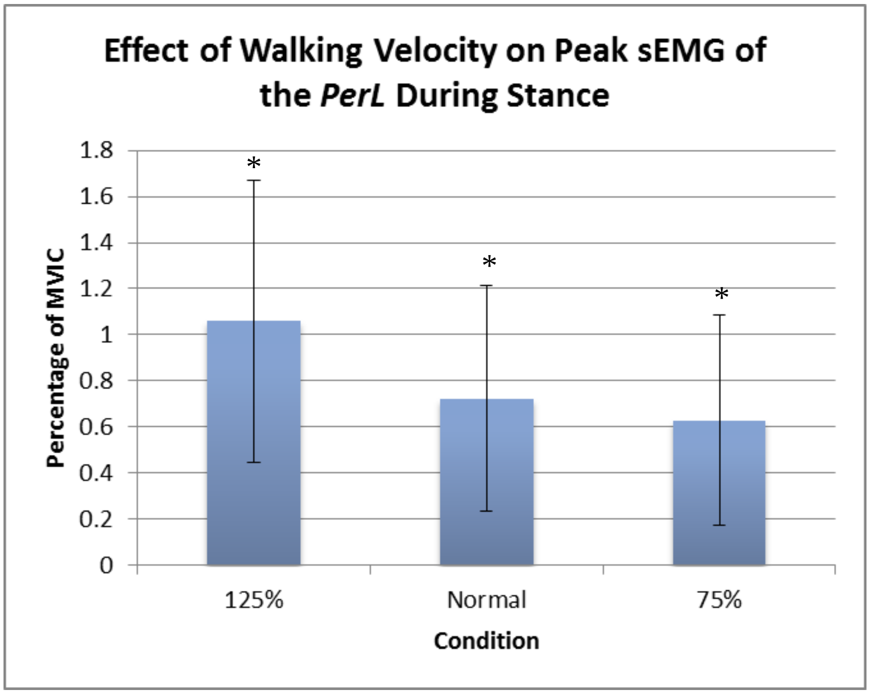

Figure 22 – Effect of walking velocity on PerL $_{peak}$ during stance. Notations (*) indicate significance at (p<0.05).

Peak Soleus

Follow-up ANOVAs found a significant difference for SOL $_{peak}$ ($F(2,94) = 20.79$, $p < 0.001$, $\eta^2 = 0.307$, *Power* = 1.000) during the stance phase. Post hoc analyses using the LSD criterion determined that the SOL $_{peak}$ was significantly greater during the 125% velocity condition than Normal (p<0.001) and 75% (p<0.001), though Normal and 75% velocity were not significantly different, indicating that velocities above normal have a significant effect on SOL $_{peak}$ (Figure 23).

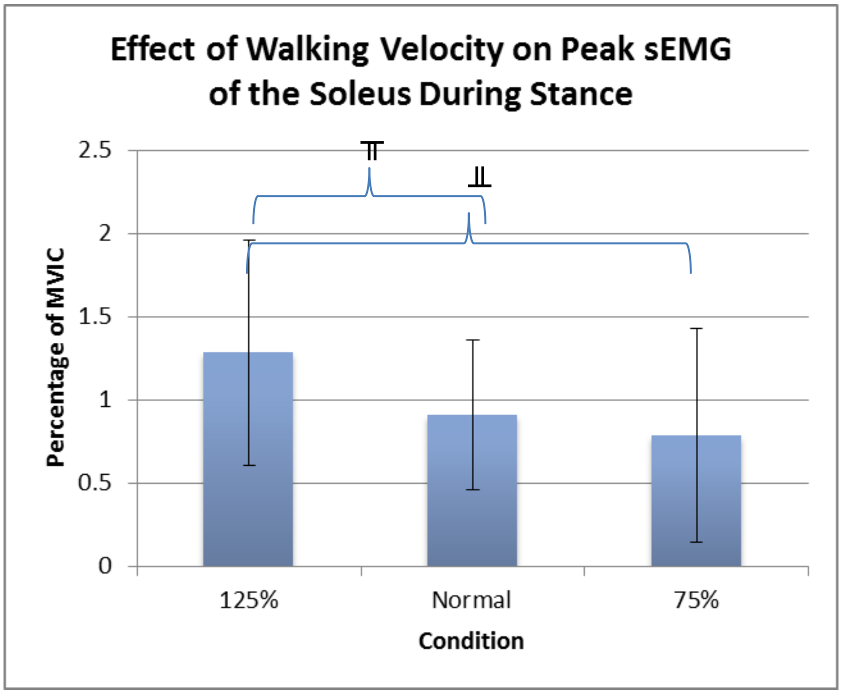

Figure 23 – Effect of walking velocity on SOL $_{peak}$ during stance. Notations (╥ ╨) indicate significance at (p<0.05).

Peak Biceps femoris

Follow-up ANOVAs found a significant difference for BF $_{peak}$ (F(2,94) = 8.737, p = 0.003, η^2= 0.157, *Power* = 0.966) during the stance phase. Post hoc analyses using the LSD criterion determined that the BF $_{peak}$ during the stance phase was significantly different between all three velocity conditions. Specifically, the 125% velocity condition was significantly greater than Normal (p=0.021) and 75% (p=0.001) while Normal was significantly greater than 75% (p=0.002) indicating that alterations in velocity have a significant effect on biceps femoris activity during the stance phase of walking (Figure 24).

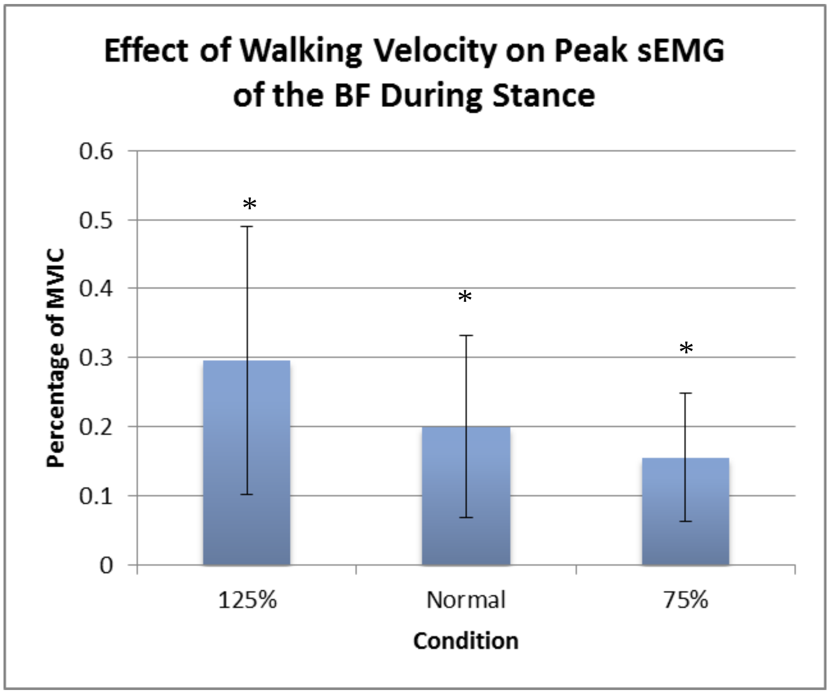

Figure 24 – Effect of walking velocity on BF $_{Peak}$ during stance. Notations (*) indicate significance at (p<0.05).

Peak Right Gluteus medius

Follow-up ANOVAs yielded a significant difference for rGM $_{peak}$ (F(2,94) = 16.222, p < 0.001, η^2= 0.257, *Power* = 0.999) during the stance phase. Post hoc analyses using the LSD criterion determined that the rGM $_{peak}$ during the stance phase was significantly different between all three velocity conditions. Specifically, the 125% velocity condition was significantly greater than Normal (p<0.001) and 75% (p=0.001) while Normal was significantly greater than 75% (p=0.027) indicating that alterations in velocity have a significant effect on right gluteus medius activity during the stance phase of walking (Figure 25).

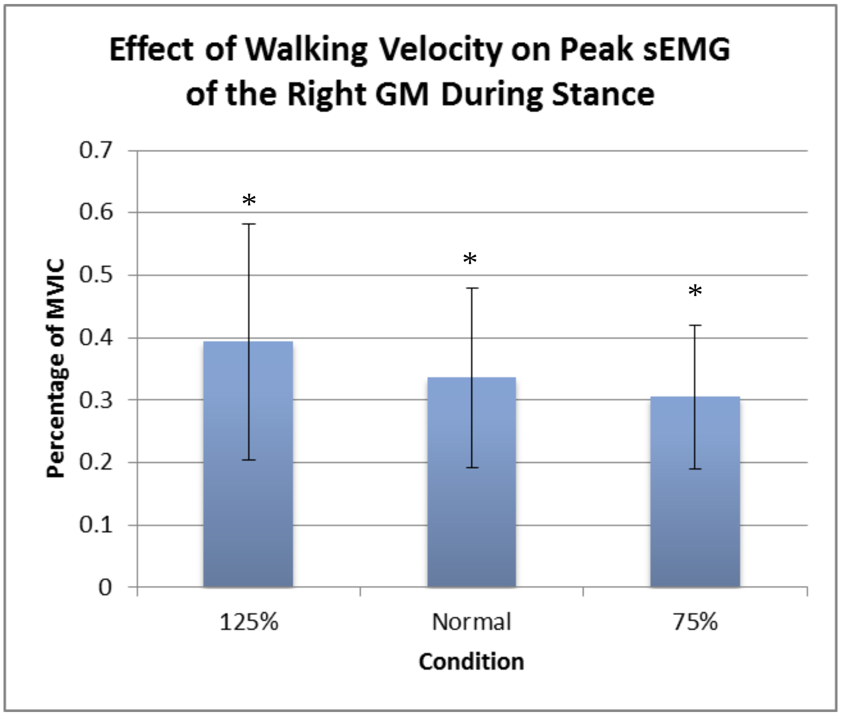

Figure 25 – Effect of walking velocity on rGM $_{Peak}$ during stance. Notations (*) indicate significance at (p<0.05).

Peak Left Gluteus medius

Follow-up ANOVAs yielded a significant difference for lGM $_{peak}$ ($F(2,94) = 39.685$, $p < 0.001$, $\eta^2 = 0.458$, *Power* = 1.000) during the stance phase. Post hoc analyses using the LSD criterion determined that the lGM $_{peak}$ during the stance phase was significantly different between all three velocity conditions. Specifically, the 125% velocity condition was significantly greater than Normal (p<0.001) and 75% (p=0.001) while Normal was significantly greater than 75% (p=0.004) indicating that alterations in velocity have a significant effect on left gluteus medius activity during the stance phase of walking (Figure 26).

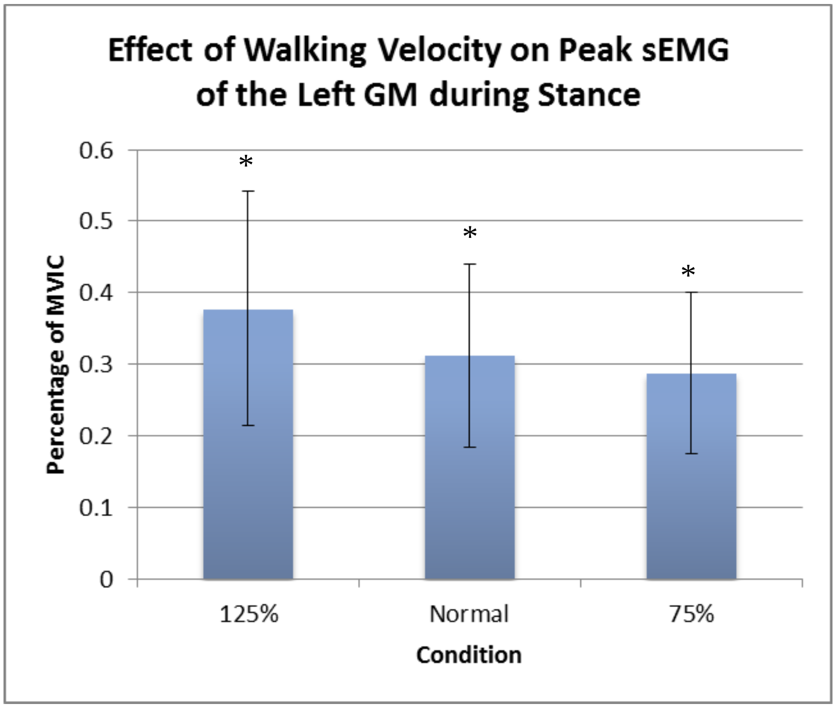

Figure 26 – Effect of walking velocity on lGM $_{Peak}$ during stance. Notations (*) indicate significance at (p<0.05).

Peak Erector spinae

Follow-up ANOVAs yielded a significant difference for ES $_{peak}$ ($F(2,94) = 11.478$, $p < 0.001$, $\eta^2 = 0.200$, *Power* = 0.935) during the stance phase. Post hoc analyses using the LSD criterion determined that the ES $_{peak}$ during the stance phase was significantly different between all three velocity conditions. Specifically, the 125% velocity condition was significantly greater than Normal (p=0.023) and 75% (p=0.001) while Normal was significantly greater than 75% (p<0.001) indicating that alterations in velocity have a significant effect on right erector spinae activity during the stance phase of walking (Figure 27).

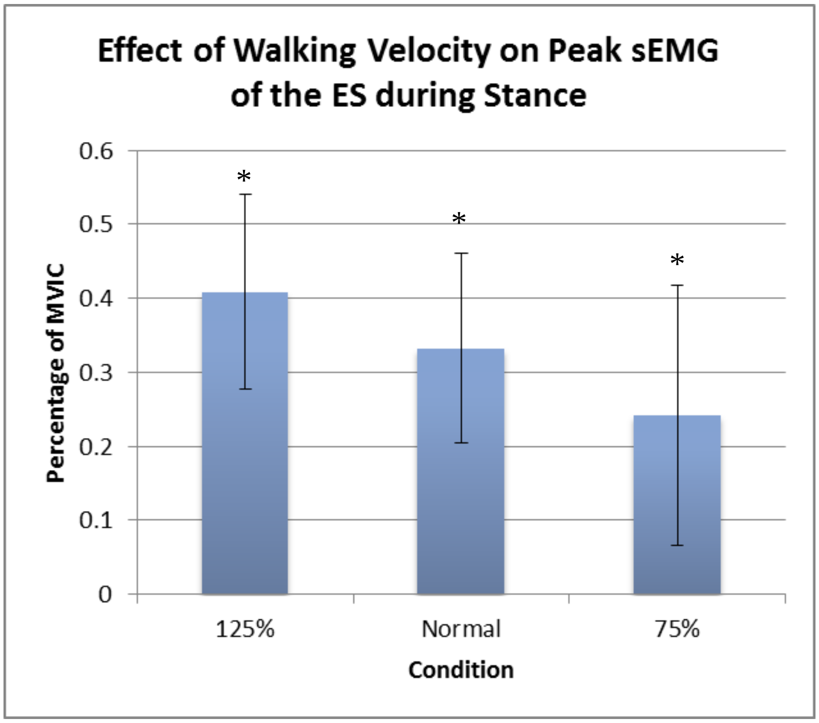

Figure 27 – Effect of walking velocity on ES $_{Peak}$ during stance. Notations (*) indicate significance at (p<0.05).

Non-significant Dependent sEMG Variables During Varying Velocities

Follow-up ANOVAs found no significant condition effect for the following sEMG variables (Table 7); indicating that changes in velocity did not significantly affect participants for the dependent variables when velocity was altered at the three levels (Normal, 125%, 75%).

Table 7 - Non-significant condition effects for sEMG when velocity was varied.

	F	df	*N*	*p*	η^2	*Power*
PerL $_{onset}$	1.420	2	94	0.47	0.029	0.298
Sol $_{onset}$	1.533	2	94	0.221	0.032	0.319
BF $_{onset}$	0.860	2	94	0.405	0.018	0.176
rGM $_{onset}$	1.632	2	94	0.210	0.034	0.337
lGM $_{onset}$	2.892	2	94	0.060	0.058	0.553
ES $_{onset}$	2.989	2	94	0.055	0.060	0.568

Surface Electromyography: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the sEMG dependent variables (PerL $_{peak}$, Sol $_{peak}$, BF $_{peak}$, rGM $_{peak}$, lGM $_{peak}$, ES $_{peak}$, PerL $_{onset}$, Sol $_{onset}$, BF $_{onset}$, rGM $_{onset}$, lGM $_{onset}$, ES $_{onset}$) when velocity was altered.

Section 3: Cadence Effects

To understand the effects of walking cadence on relative phase differences during overground walking, cadence was manipulated on three levels (Normal, 125% of Normal, and 75% of Normal) during a test session separate from the velocity session. To determine differences between conditions, a 3 (condition) x 2 (gender) repeated measures MANOVA was performed for eleven dependent kinematic variables (Table 8). A significant *($p<0.001$)* Box's M test was found, therefore, the application of the more robust Pillai's Trace multivariate test was utilized for interpretation of output statistics.

Table 8 – Kinematic Variables: Cadence Effects

Independent Variables	**Dependent Variables**
Gender	Relative Phase $_{mean}$
- Male	- Trunk-Pelvis (TP $_{mean}$)
- Female	- Trunk-Right Leg (TL $_{mean}$)
Condition	- Trunk-Left Leg (PL $_{mean}$)
- Normal Cadence	Relative Phase $_{SD}$
- 125% of Normal	- Trunk-Pelvis (TP $_{SD}$)
- 75% of Normal	- Trunk-Right Leg (TL $_{SD}$)
	- Trunk-Left Leg (PL $_{SD}$)
	Euler Angles (at foot strike)
	- Pelvis
	- Trunk
	- Right Leg
	Spatiotemporal Variables
	- Velocity
	- Normalized stride length

The multivariate repeated measures ANOVA with a p value set *apriori* at < 0.05 yielded a non-significant Gender*Condition interaction and a non-significant Gender effect; however, significant Condition ($\Lambda_{Pillai} = 0.998$, $F(22,27) = 25.670$, $p < 0.001$, $\eta^2 = 0.954$, *Power* = 1.000) differences were found.

Spatiotemporal Variables: Condition Differences

Average cadence for all participants was 139.6 steps*min^{-1}(sm^{-1}), 112.5 sm^{-1}, and 84.7 sm^{-1} for the 125%, Normal, and 75% conditions, respectively. Follow-up ANOVAs found a significant difference for velocity ($F(2,96) = 199.71$, $p < 0.001$, $\eta^2 = 0.745$, *Power* = 1.000) and normalized stride length ($F(2,96) = 66.854$, $p < 0.001$, $\eta^2 = 0.494$, *Power* = 1.000). Post hoc analyses using the LSD criterion found a significant difference ($p<0.001$) between velocity for all three conditions (Figure 28). Velocity was highest during the 125% cadence condition and lowest during the 75% cadence trials. Normalized stride length was also found to be significantly different ($p<0.001$) between all conditions (Figure 29), with the 125% cadence condition exhibiting the longest stride length and 75% cadence trials exhibiting the shortest stride lengths.

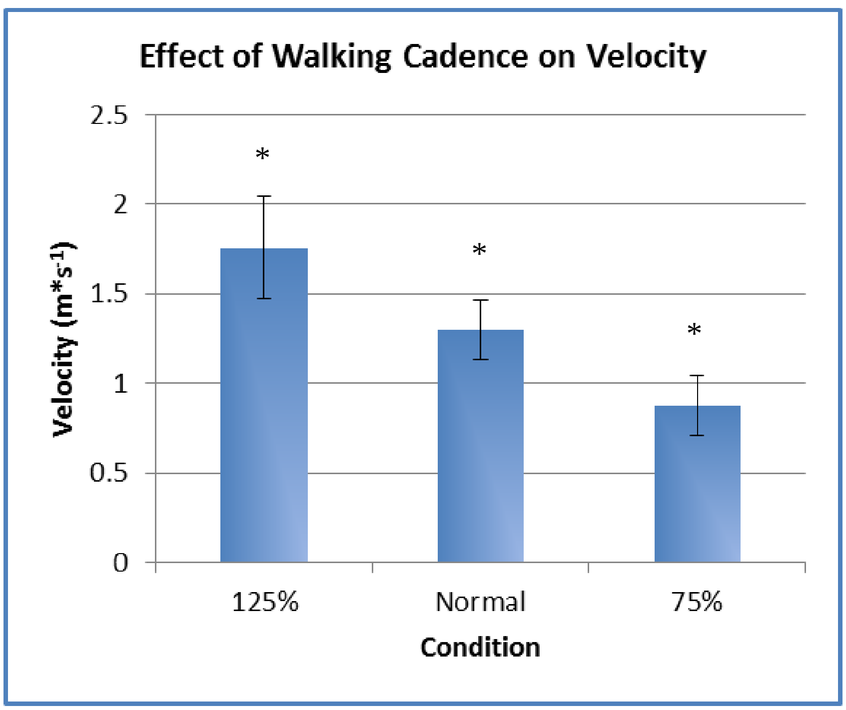

Figure 28 – Effect of cadence on velocities. Notations (*) indicate significance at ($p<0.05$).

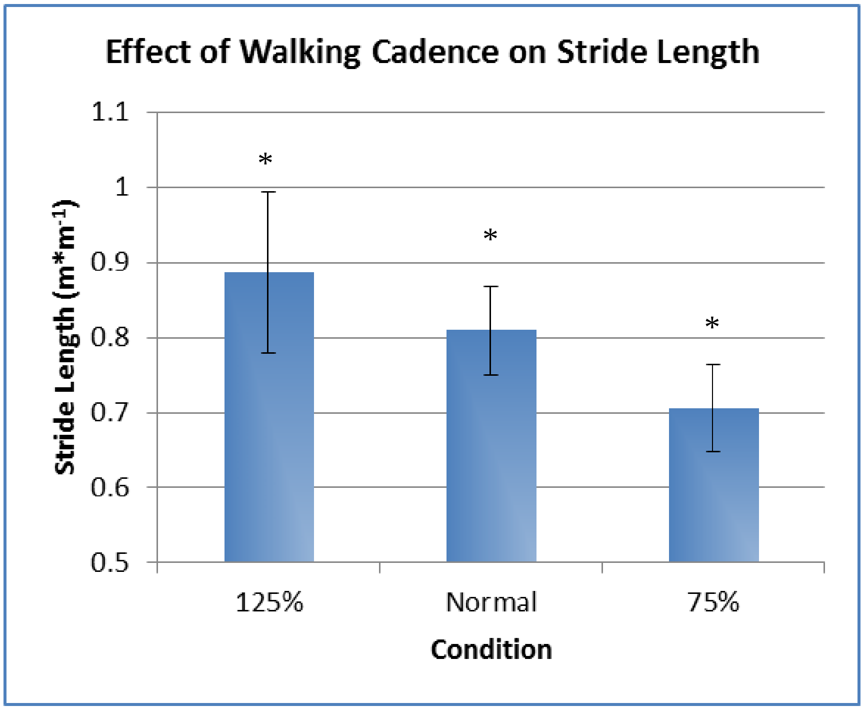

Figure 29 – Effect of cadence on normalized stride length. Notations (*) indicate significance at (p<0.05).

Spatiotemporal Variables: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the spatiotemporal variables (Velocity, Stride length) when Cadence was altered.

Kinematics: Condition Differences

Trunk-Pelvis $_{mean}$ & Trunk-Pelvis $_{SD}$

Follow-up ANOVAs found a significant difference for TP $_{mean}$ (F(2,96) = 23.771, p < 0.001, η^2= 0.331, *Power* = 1.000) and TP $_{SD}$ (F(2,96) = 3.818, p = 0.025, η^2= 0.074, *Power* = 0.681). Post hoc analyses using the LSD criterion found that the 75% cadence condition displayed a significantly greater (p<0.001) TP $_{mean}$ than 125% and Normal (Figure 30). Post hoc analyses also revealed that TP $_{SD}$ was significantly lower for 125% cadence than 75% cadence

(p=0.006) only (Figures 31). Results indicate that the pelvis and trunk were most out-of-phase during the 75% cadence condition, relative to Normal and 125% cadence.

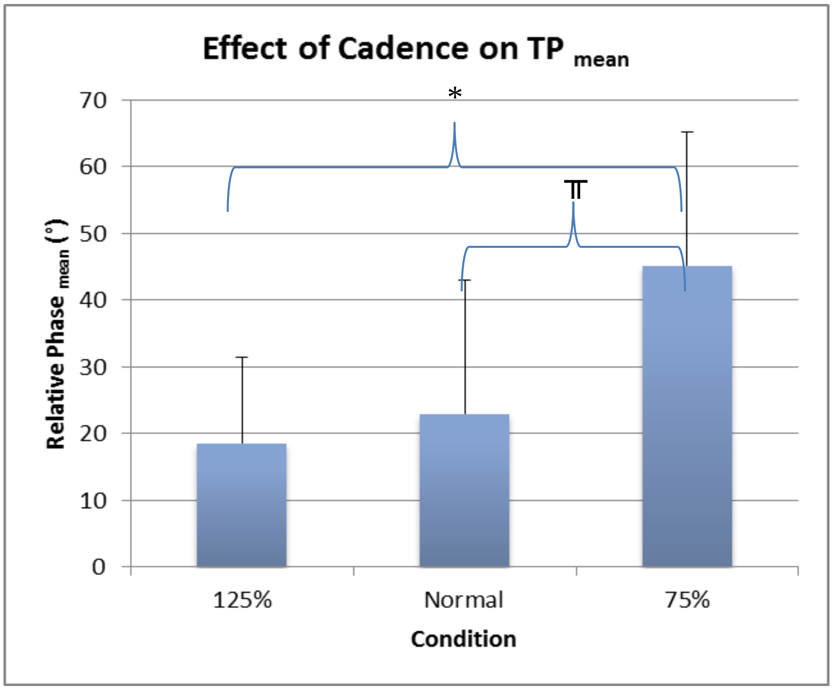

Figure 30 – Effect of walking cadence on TP $_{mean}$. Notations (*╥) indicate significance at (p<0.05).

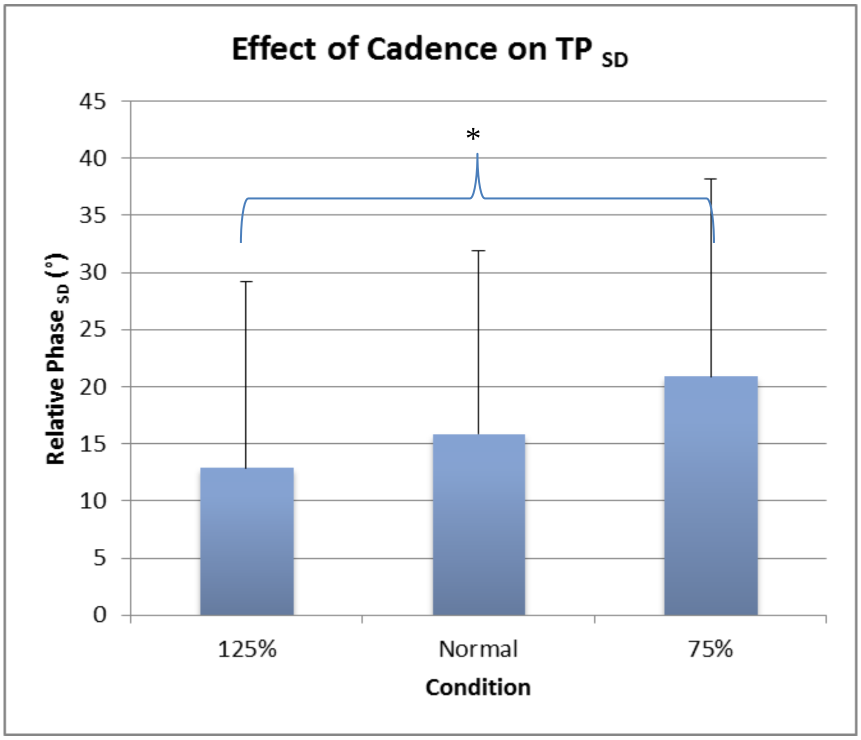

Figure 31 – Effect of walking cadence on TP $_{SD}$. Notations (*) indicate significance at (p<0.05).

Trunk-Leg $_{mean}$ & Trunk-Leg $_{SD}$

A follow-up ANOVA found a significant effect of cadence on TL $_{mean}$ (F(2,96) = 4.678, p = 0.012, η^2= 0.089, *Power* = 0.774) with post-hoc analyses using the LSD criterion indicating that the normal cadence condition was significantly lower than 125% cadence (p=0.002) and 75% cadence (p=0.037), indicating the trunk and leg were significantly more synchronized during the Normal condition relative to the 125% cadence and 75% cadence (Figure 32). A follow-up ANOVA revealed no significant difference for TL $_{SD}$ between any of the conditions (F(2,96) = 0.186, p = 0.931, η^2= 0.004, *Power* = 0.078), indicating that the variability of the TL were not significantly different across the cadence conditions.

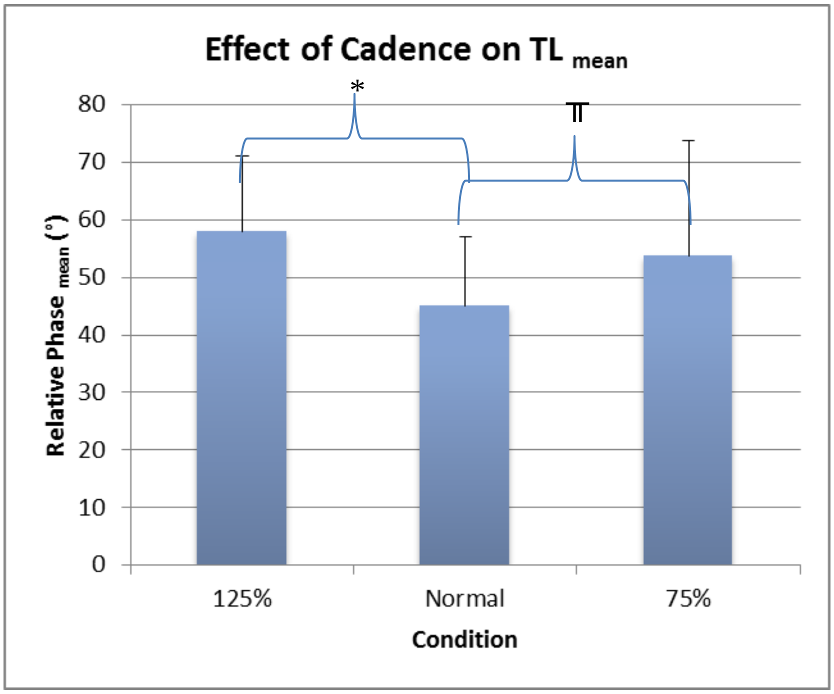

Figure 32 – Effect of walking cadence on TL $_{mean}$. Notations (*╥) indicate significance at (p<0.05).

Pelvis-Leg $_{mean}$ & Pelvis-Leg $_{SD}$

A follow-up ANOVA found a significant effect of cadence on PL $_{mean}$ (F(2,96) = 11.963, p < 0.001, η^2= 0.200, *Power* = 0.994) with post-hoc analyses using a LSD criterion indicating that the Normal cadence condition was significant lower than 125% cadence (p=0.013) and 75% cadence (p<0.001) , while 125% cadence was significantly lower (p=0.027) than 75% cadence (Figure 33). A follow-up ANOVA revealed no significant difference for TL $_{SD}$ (F(2,96) = 2.747, p = 0.069, η^2= 0.054, *Power* = 0.531). Results indicate that the pelvis and leg were significantly more synchronized during the Normal condition relative to the 125% cadence and 75% cadence while variability in TP measures were not significantly altered by the cadence.

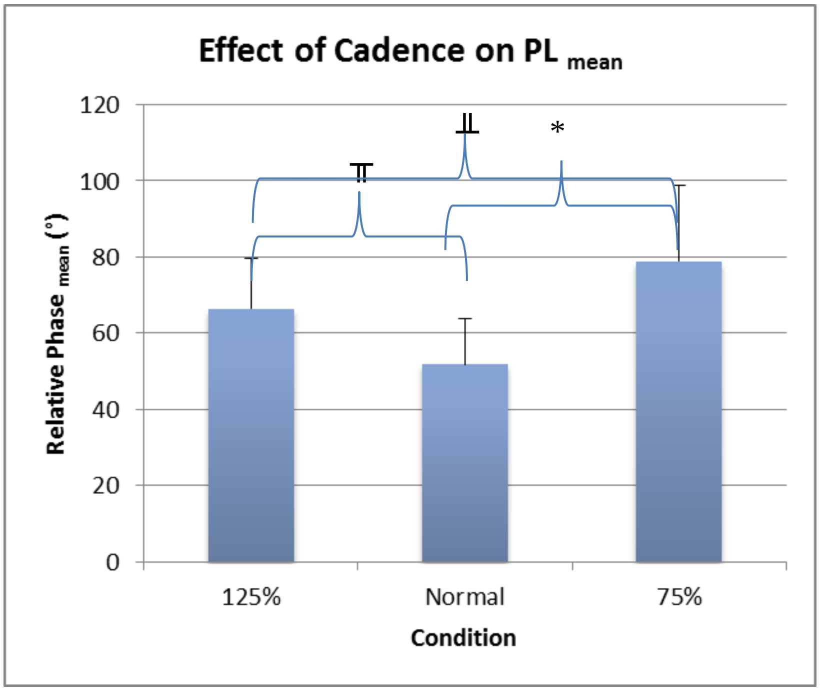

Figure 33 – Effect of walking cadence on PL $_{mean}$. Notations (*╥ ╨) indicate significance at (p<0.05).

Pelvis Rotation at Foot Strike

Follow-up ANOVAs found a significant effect of cadence on the magnitude of pelvic rotation (F(2,96) = 24.150, p < 0.001, η^2= 0.335, *Power* = 1.000) with post-hoc analyses using a LSD criterion indicating that the 125% cadence condition was significant greater than 75% cadence and Normal condition (p<0.001), yet no difference between Normal or 75% cadence was noted. Results indicate that while the pelvis was exhibiting left transverse pelvic girdle rotation at foot strike when walking at Normal or 75% cadence, the pelvis was right transverse rotated at foot strike during the 125% cadence (Figure 34).

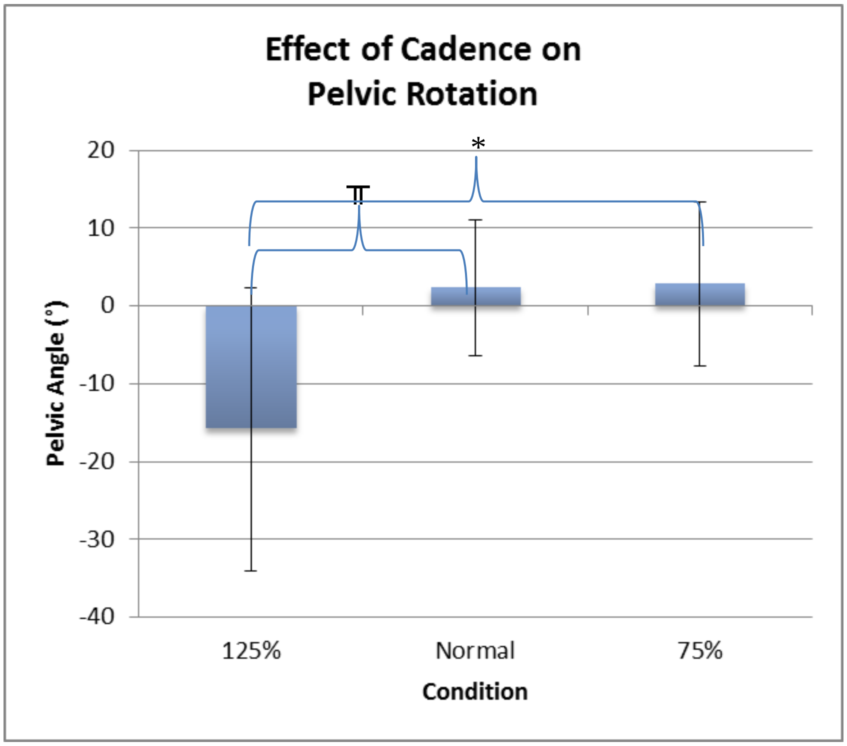

Figure 34 – Effect of walking cadence on pelvic rotation. Notations (*╥) indicate significance at (p<0.05).

Trunk Rotation at Foot Strike

Follow-up ANOVAs found a significant effect of cadence on the magnitude of trunk rotation ($F(2,96) = 67.904$, $p < 0.001$, $\eta^2 = 0.583$, *Power* = 1.000) with post-hoc analyses using a LSD criterion indicating that the 125% cadence condition was significant greater than 75% cadence and Normal condition (p<0.001), yet no difference between Normal and 75% cadence, indicating greater trunk rotation at foot strike when walking at the 125% cadence (Figure 35). Furthermore, while the trunk was rotated to the right at foot strike during the Normal and 75% cadence conditions, results indicate that the trunk was rotated to the left at the 125% cadence condition.

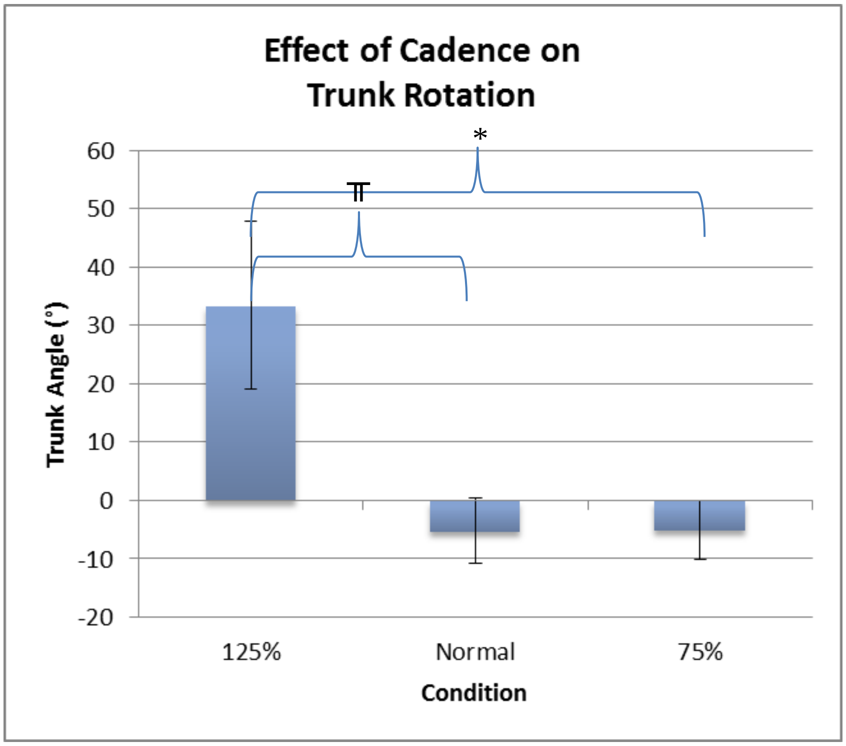

Figure 35 – Effect of walking cadence on trunk rotation. Notations (*₸) indicate significance at (*p*<0.05).

Leg Rotation at Foot Strike

Follow-up ANOVAs found a significant effect of cadence on the magnitude of pelvic rotation ($F(2,96) = 112.777$, $p < 0.001$, $\eta^2 = 0.701$, *Power* = 1.000) with post-hoc analyses using a LSD criterion indicating that the 125% cadence condition was significantly greater than than 75% cadence and Normal conditions ($p<0.001$), yet no difference between Normal or 75% cadence was noted, signifying greater external rotation at foot strike when walking at 125% of Normal cadence (Figure 36).

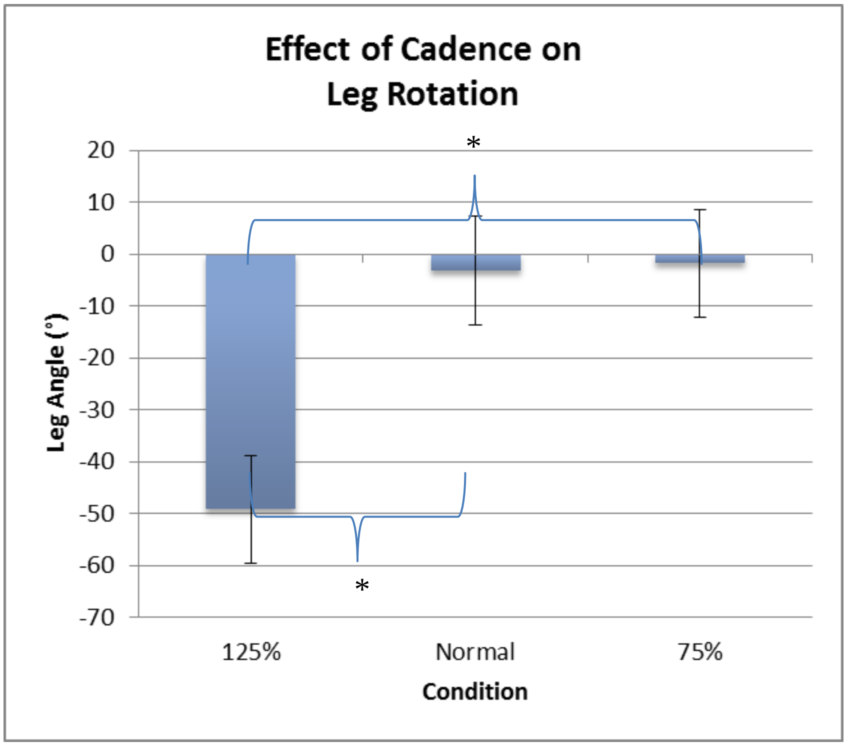

Figure 36 – Effect of walking cadence on leg rotation. Notations (*) indicate significance at (p<0.05).

Kinematics: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the kinematic dependent variables (TP_{mean}, TP_{sd}, TL_{mean}, TL_{sd}, PL_{mean}, PL_{sd}, θ_{pelvis}, θ_{trunk}, θ_{leg}) when cadence was altered.

Surface Electromyography

To determine muscle activity during overground walking, electromyography (EMG) was recorded at 1500 HZ (Figure 5) using six pairs of bipolar Ag-AgCL surface electrodes (Trace-Rite, Bio-Detek Inc. Pawtucket, RI, USA), placed longitudinally over the bellies of six muscles (Table 7). A 3 (condition) x 2 (gender) repeated measures MANOVA was performed for twelve dependent variables (Table 7).To understand the effects of cadence on muscle activity during overground walking, cadence was manipulated on three levels (Normal, 125% of Normal, and 75% of Normal) during a single test session. A significant (*$p<0.001$*) Box's M test was found, therefore, the application of the more robust Pillai's Trace multivariate test was utilized for interpretation of output statistics. The multivariate repeated measures ANOVA with a p value set *apriori* at < 0.05 yielded a non-significant Gender*Condition interaction and a non-significant Gender effect; however, a significant Condition ($\Lambda_{Pillai} = 0.847$, $F(24,24) = 5.528$, $p < 0.001$, $\eta^2 = 0.947$, *Power* = 1.000) effect was found.

Surface Electromyography: Condition Differences

Peak Peroneus longus

Follow-up ANOVAs found a significant difference for $PerL_{peak}$ ($F(2,94) = 17.960$, $p < 0.001$, $\eta^2 = 0.276$, *Power* = 1.000). Post hoc analyses using the LSD criterion determined that the $PerL_{peak}$ during the 125% cadence condition was significantly greater ($p<0.001$) than the 75% cadence and Normal Cadence ($p<0.001$) conditions (Figure 37).

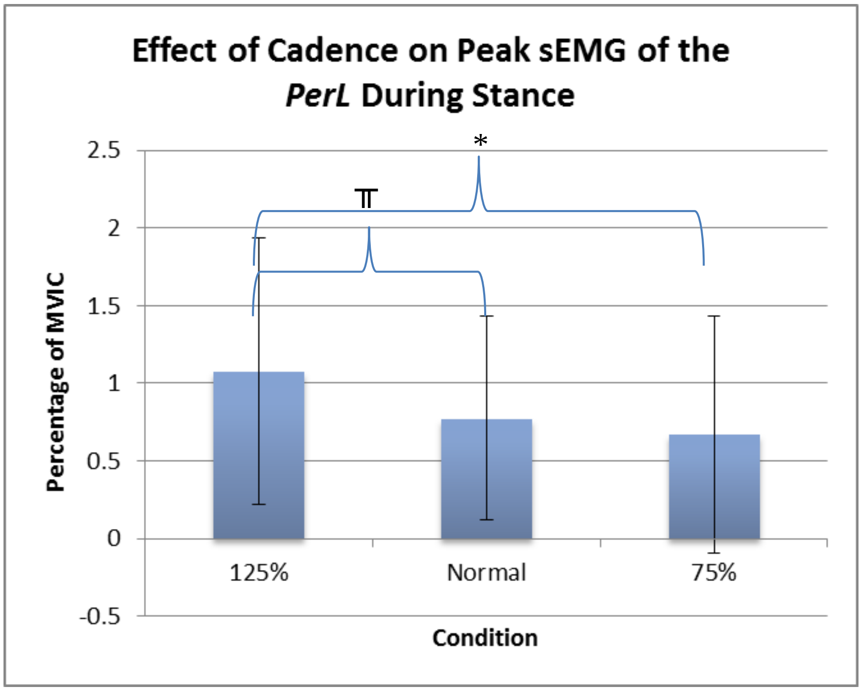

Figure 37 – Effect of cadence on PerL $_{peak}$. Notations (*╥) indicate significance at (p<0.05).

Peak Soleus

Follow-up ANOVAs found a significant difference for SOL $_{peak}$ (F(2,94) = 50.475, p < 0.001, η^2= 0.518, *Power* = 1.000). Post hoc analyses using the LSD criterion determined that the Sol $_{peak}$ during the 125% cadence condition was significantly greater than Normal (p<0.001) and 75% cadence (p<0.001). Furthermore, Normal was significantly greater than 75% cadence (p<0.001) indicating that alterations in cadence have a significant effect on soleus activity during the stance phase of walking (Figure 38).

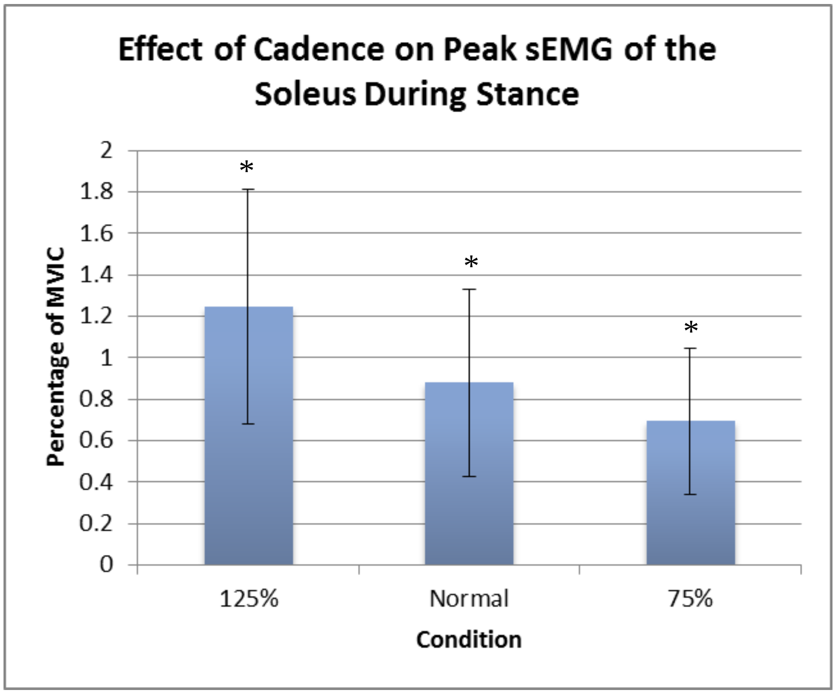

Figure 38 – Effect of walking cadence on SOL $_{Peak}$. Notations (* ╥) indicate significance at (p<0.05).

Peak Biceps femoris

Follow-up ANOVAs found a significant difference for BF $_{peak}$ ($F(2,94) = 22.052$, $p < 0.001$, $\eta^2 = 0.319$, *Power* = 1.000) during the stance phase. Post hoc analyses using the LSD criterion determined that the BF $_{peak}$ during the stance phase was significantly greater during 125% cadence conditions than either Normal (p<0.001) or 75% cadence (p<0.001); however, Normal was not significantly different than 75% cadence indicating that alterations in cadence above Normal have a significant effect on biceps femoris activity during the stance phase of walking (Figure 39).

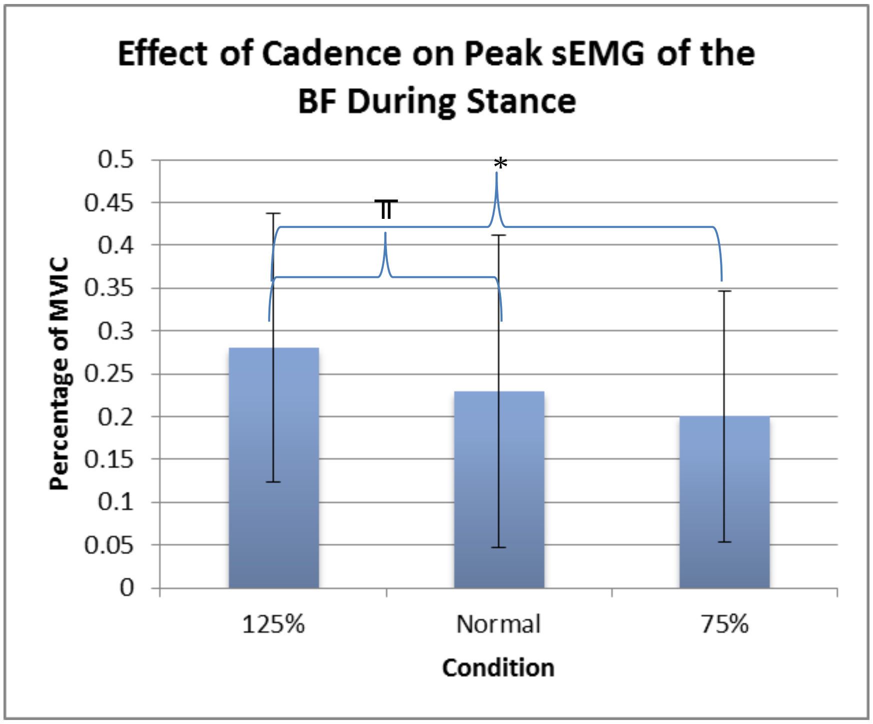

Figure 39 – Effect of walking cadence on BF $_{Peak}$. Notations (* ╥) indicate significance at (p<0.05).

Peak Right Gluteus medius

Follow-up ANOVAs yielded a significant difference for rGM $_{peak}$ (F(2,94) = 24.678, p < 0.001, η^2= 0.344, *Power* = 1.00). Post hoc analyses using the LSD criterion determined that the rGM $_{peak}$ during the 125% cadence condition was significantly greater than during the Normal (p<0.001) and 75% cadence conditions (p<0.001) only, indicating that changes in cadence at lower levels did not have a significant effect on rGM $_{peak}$, but changes in cadence above Normal did have a significant impact on peak muscle activity (Figure 40).

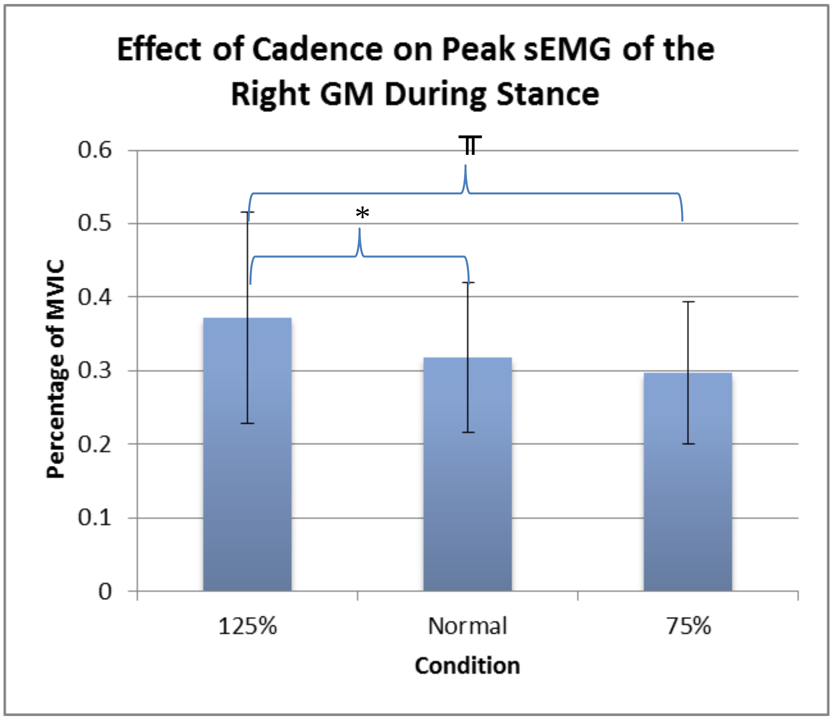

Figure 40 – Effect of cadence on rGM $_{Peak}$. Notations (*╥) indicate significance at (p<0.05).

Peak Left Gluteus medius

Follow-up ANOVAs yielded a significant difference for lGM $_{peak}$ ($F(2,94) = 38.375$, $p < 0.001$, $\eta^2 = 0.449$, *Power* = 1.000). Post hoc analyses using the LSD criterion determined that the lGM $_{peak}$ during the 125% cadence condition was significantly greater than the Normal (p<.001) and 75% cadence conditions (p<0.001) only, indicating that changes in cadence at lower levels did not have a significant effect on lGM $_{peak}$, but changes in cadence above Normal did have a significant impact on peak muscle activity (Figure 41).

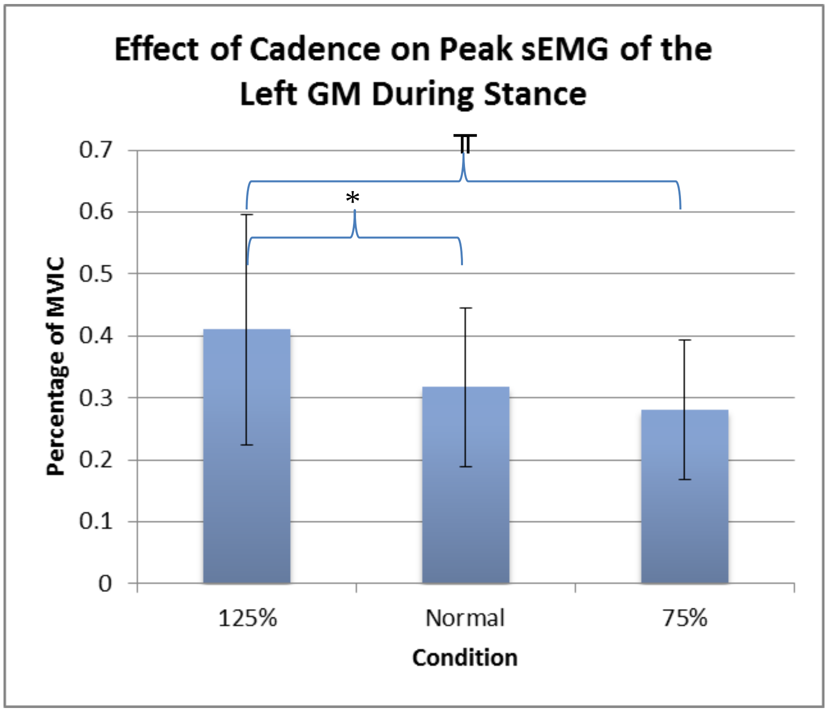

Figure 41 – Effect of cadence on lGM $_{Peak}$. Notations (*╥) indicate significance at (p<0.05).

Peak Erector spinae

Follow-up ANOVAs yielded a significant difference for ES $_{peak}$ ($F(2,94) = 6.960$, p = 0.002, η^2= 0.129, *Power* = 0.918) during the stance phase. Post hoc analyses using the LSD criterion determined that the ES $_{peak}$ during the stance phase was significantly greater during the 125% cadence condition than Normal (p=0.030) and 75% cadence (p=0.001) only, indicating that alterations in cadence above Normal have a significant effect on right erector spinae activity during the stance phase of walking (Figure 42).

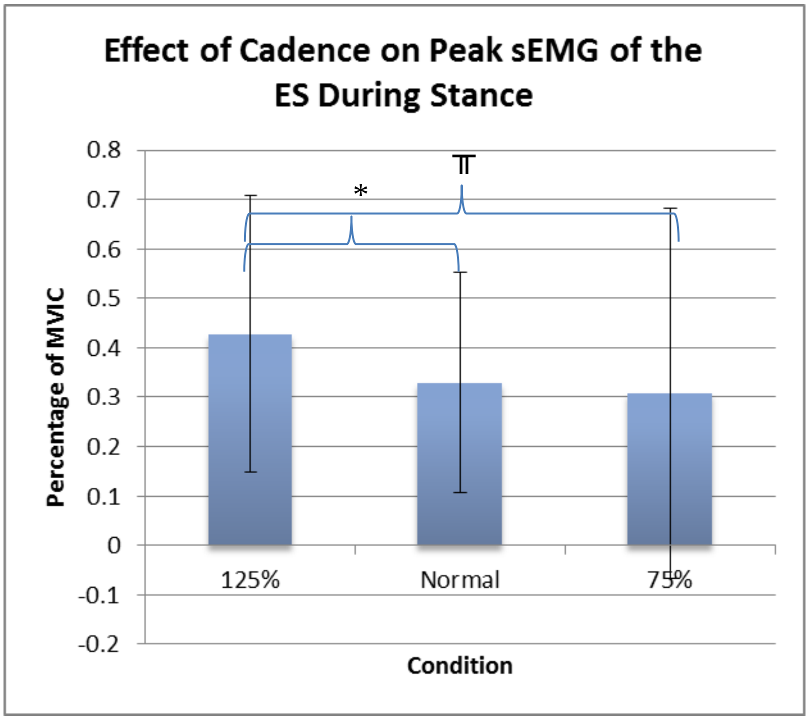

Figure 42 – Effect of cadence on ES $_{Peak}$. Notations (*╥) indicate significance at (p<0.05).

Non-significant Dependent sEMG Variables During Varying Cadences

Follow-up ANOVAs found no significant condition effect for the following sEMG variables (Table 9); indicating that changes in cadence did not significantly affect participants for the dependent variables when cadence was altered at the three levels (Normal, 125%, 75%).

Table 9 - Non-significant condition effects for sEMG when cadence was varied.

	F	df	*N*	*p*	η^2	*Power*
PerL $_{onset}$	0.027	2	94	0.974	0.001	0.054
Sol $_{onset}$	1.144	2	94	0.323	0.024	0.246
BF $_{onset}$	0.597	2	94	0.561	0.013	0.146
rGM $_{onset}$	1.082	2	94	0.343	0.022	0.234
lGM $_{onset}$	2.010	2	94	0.140	0.041	0.406
ES $_{onset}$	2.183	2	94	0.118	0.044	0.436

Surface Electromyography: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the sEMG dependent variables (PerL $_{peak}$, Sol $_{peak}$, BF $_{peak}$, rGM $_{peak}$, lGM $_{peak}$, ES $_{peak}$, PerL $_{onset}$, Sol $_{onset}$, BF $_{onset}$, rGM $_{onset}$, lGM $_{onset}$, ES $_{onset}$) when cadence was altered.

Section 4: Footwear Effects

To understand the effects of footwear on relative phase differences during overground walking, footwear was manipulated on four levels (Table 13) during two separate testing sessions. Only kinematics and sEMG data from the Normal conditions (velocity and cadence) were used to examine potential footwear effects during Normal walking (i.e. 125% and 75% conditions were excluded from analyses). To determine differences between conditions, a 4 (condition) x 2 (gender) repeated measures MANOVA was performed for twelve dependent kinematic variables (Table 10). A significant (*$p<0.001$*) Box's M test was found, therefore, the application of the more robust Pillai's Trace multivariate test was utilized for interpretation of output statistics.

Table 10 – Kinematic Variables: Footwear Effects

Independent Variables	**Dependent Variables**
Gender	Relative Phase $_{mean}$
- Male	- Trunk-Pelvis (TP $_{mean}$)
- Female	- Trunk-Right Leg (TL $_{mean}$)
Condition	- Trunk-Left Leg (PL $_{mean}$)
- Shod (SH)	Relative Phase $_{SD}$
- Barefoot (BaFt)	- Trunk-Pelvis (TP $_{SD}$)
- Insole only (IN)	- Trunk-Right Leg (TL $_{SD}$)
- Insole with shoe (INSH)	- Trunk-Left Leg (PL $_{SD}$)
	Euler Angles (at foot strike)
	- Pelvis
	- Trunk
	- Right Leg
	Spatiotemporal Variables
	- Velocity
	- Cadence
	- Normalized stride length

The multivariate repeated measures ANOVA with a *p* value set *apriori* at < 0.05 yielded a non-significant Gender*Condition interaction; however, significant Gender ($\Lambda_{Pillai} = 0.767$,

$F(12,37) = 10.152$, $p < 0.001$, $\eta^2 = 0.767$, *Power* = 1.000) and Condition ($\Lambda_{Pillai} = 0.956$, $F(36,13) = 7.839$, $p < 0.001$, $\eta^2 = 0.077$, *Power* = 0.956) differences were found.

Spatiotemporal Variables: Condition Differences

Follow-up ANOVAs found a significant footwear effect on normalized stride length ($F(3,144) = 58.310$, $p < 0.001$, $\eta^2 = 0.548$, *Power* = 1.000), velocity ($F(3,144) = 9.479$, $p < 0.001$, $\eta^2 = 0.164$, *Power* = 0.997), and cadence ($F(3,144) = 7.427$, $p < 0.001$, $\eta^2 = 0.134$, *Power* = 0.984). Post hoc analyses using the LSD criterion found a significant difference ($p<0.001$) in normalized stride length between shod (SH) vs. barefoot (BaFt), SH vs. insole (IN), insole-shod (INSH) vs. BaFt, and INSH vs. IN. Differences in normalized stride length between SH vs. INSH and BaFt vs. IN were non-significant (Figure 43).

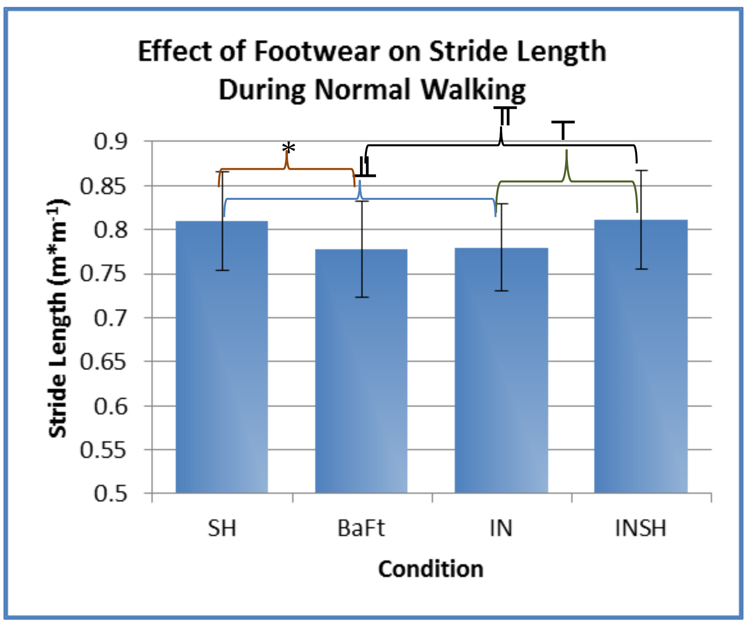

Figure 43 – Effect of footwear on normalized stride length during normal walking. Notations (*╥ ╨ ┬) indicate significance at ($p<0.05$).

Post hoc analyses using the LSD criterion found a significant difference ($p \leq 0.001$) in normalized stride length between SH vs. BaF, SH vs. IN, INSH vs. BaFt, and INSH vs. IN. Differences in velocity between SH vs. INSH and BaFt vs. IN were non-significant (Figure 44).

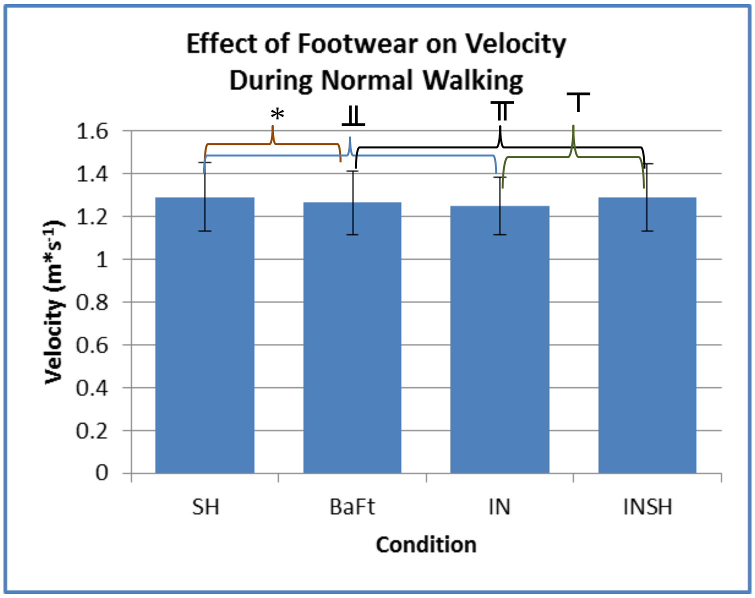

Figure 44 – Effect of footwear on velocity during normal walking. Notations (*╥ ╨ ┬) indicate significance at (p<0.05).

Post hoc analyses using the LSD criterion found a significant difference (p<0.001) in normalized stride length between SH vs. BaFt, BaFt vs. IN, and INSH vs. BaFt. Differences in cadence between SH vs. INSH, , INSH vs. IN, and SH vs. IN were non-significant (Figure 45).

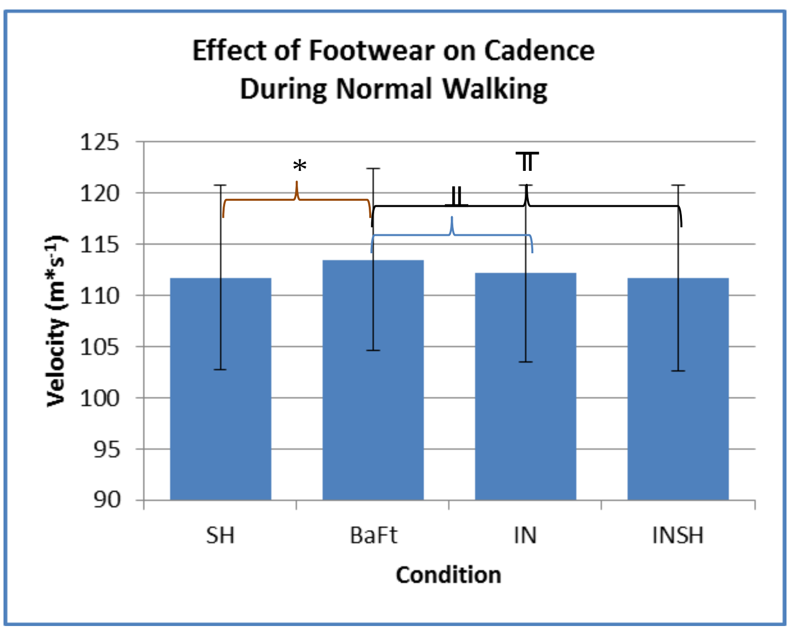

Figure 45 – Effect of footwear on cadence during normal walking. Notations (*╥ ╨) indicate significance at (p<0.05).

Spatiotemporal Variables: Gender Differences

Follow-up ANOVAs found a significant gender effect on normalized stride length ($F(1,48) = 8.224$, $p= 0.006$, $\eta^2= 0.115$, *Power* = 0.099) and cadence ($F(1,48) = 17.683$, $p < 0.001$, $\eta^2= 0.269$, *Power* = 0.206); however, a non-significant difference in velocity ($F(1,48) = 0.737$, $p =0.395$, $\eta^2= 0.0015$, *Power* = 0.086) was noted. Descriptive statistics revealed that females took significantly longer height normalized strides (Figure 46) and had a significantly higher cadence during Normal walking than males (Figure 47).

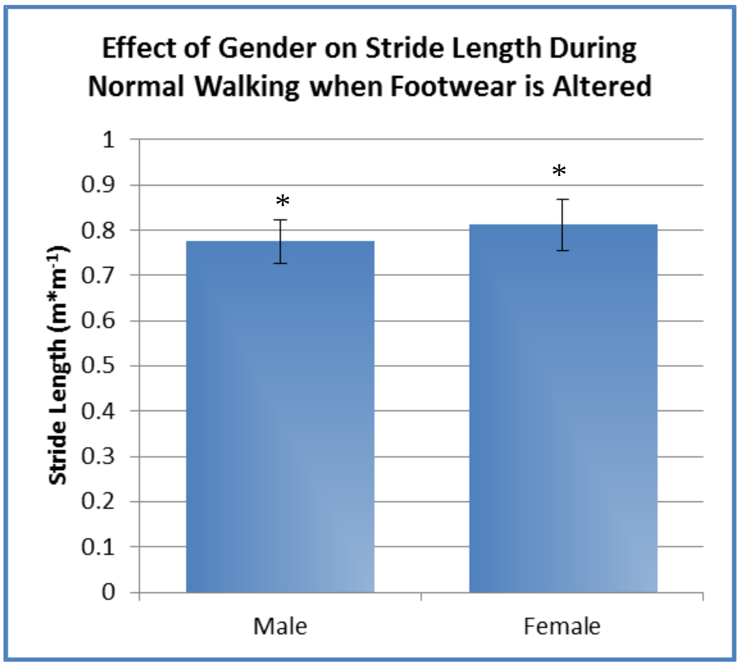

Figure 46– Effect of gender on normalized stride length during normal walking when footwear is altered. Notations (*) indicate significance at (p<0.05).

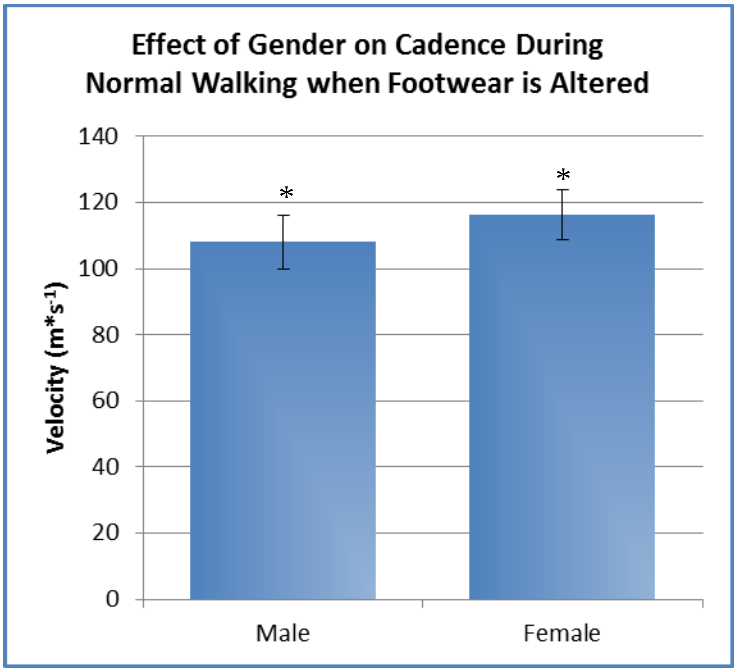

Figure 47 – Effect of gender on cadence during normal walking when footwear is altered. Notations (*) indicate significance at (p<0.05).

Kinematics: Condition Differences

Pelvis-Leg $_{mean}$

A follow-up ANOVA found a significant footwear effect on PL $_{mean}$ (F(3,144) = 4.629, p = 0.004, η^2= 0.088, *Power* = 0.884) with post-hoc analyses using a LSD criterion yielding that the BaFt condition was significantly greater than SH (p=0.048) and INSH (p = 0.008), but not IN, while IN was significantly greater than SH (p=0.022) and INSH (p=0.004) (Figure 48), indicating that the pelvis and leg were more asynchronous during the BF and IN conditions relative to the other conditions.

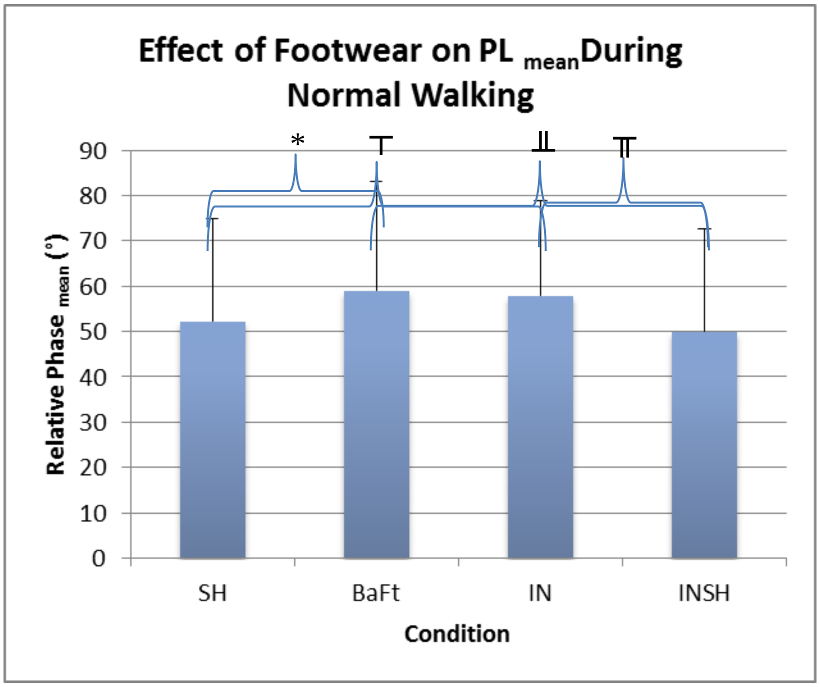

Figure 48 – Effect of footwear on PL $_{mean}$. Notations (*╥ ╨ ┬) indicate significance at (p<0.05).

Non-significant Dependent Kinematic Variables During Different Footwear

Follow-up ANOVAs found no significant condition effect for the following kinematic variables (Table 11); indicating footwear did not significantly affect these dependent variables when altered at the four levels (SH, BaFt, IN, INSH).

Table 11 - Non-significant condition effects for kinematics when footwear was varied.

	F	df	N	p	η^2	*Power*
TP $_{mean}$	0.289	3	144	0.833	0.006	0.105
TP $_{sd}$	0.961	3	144	0.413	0.020	0.258
TL $_{mean}$	2.583	3	144	0.056	0.051	0.625
TL $_{sd}$	2.274	3	144	0.082	0.045	0.565
PL $_{sd}$	0.536	3	144	0.645	0.011	0.153
Pelvic Rotation	0.937	3	144	0.424	0.019	0.253
Trunk Rotation	0.975	3	144	0.406	0.020	0.262
Leg Rotation	0.148	3	144	0.931	00.003	0.077

Kinematics: Gender Differences

Trunk-Pelvis $_{mean}$ & Trunk-Pelvis $_{SD}$

Follow-up ANOVAs revealed a non-significant gender effect for TP $_{mean}$ ($F(1,48)$ = 3.439, p = 0.070, η^2= 0.067, *Power* = 0.094); however, a significant gender difference was noted for TP $_{SD}$ ($F(1,48)$ = 6.973, p = 0.011, η^2= 0.127, *Power* = 0.481) , indicating males and females were significantly different in the level of variability between trunk and pelvis during overground walking when footwear was altered, with males exhibiting greater variability than females (Figures 49).

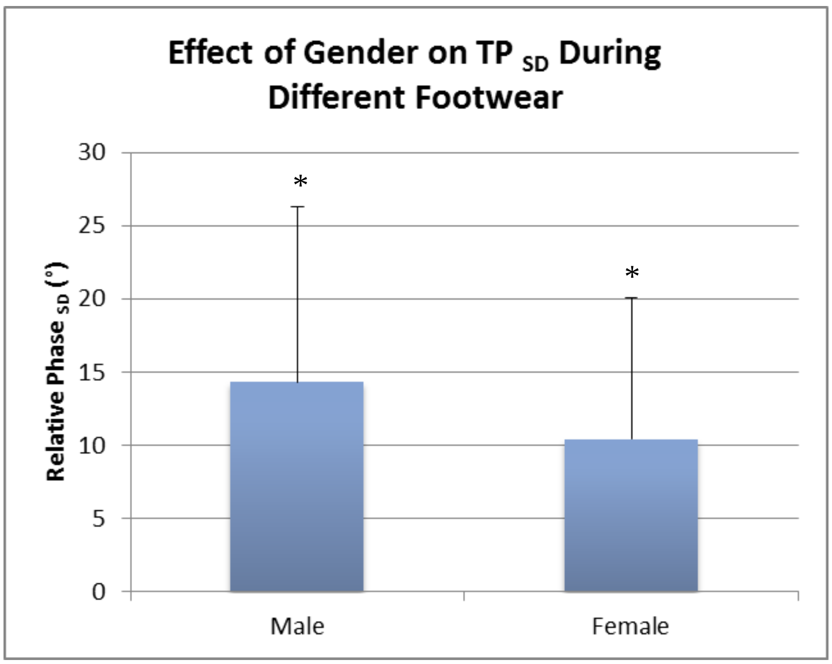

Figure 49 – Effect of gender on TP $_{SD}$ during different footwear. Notations (*) indicate significance at (p<0.05).

Non-significant Dependent Kinematic Variables between Males and Females During Altered Footwear

Follow-up ANOVAs found no significant gender effect for the following kinematic variables (Table 12); indicating changes in footwear (SH, BaFt, IN, INSH) did not affect males or females to a greater extent for the following dependent variables.

Table 12 – Non-significant gender effects when footwear was varied.

	F	df	*N*	*p*	η^2	*Power*
TL_{mean}	0.872	1	48	0.355	0.018	0.181
TL_{sd}	0.548	1	48	0.463	0.011	0.069
PL_{mean}	2.579	1	48	0.115	0.051	0.053
PL_{sd}	2.731	1	48	0.105	0.054	0.062
Pelvic Rotation	0.195	1	48	0.661	0.004	0.086
Trunk Rotation	1.838	1	48	0.181	0.037	0.170
Leg Rotation	3.144	1	48	0.083	0.061	0.107

Surface Electromyography

To understand the effects of footwear on muscle activity during overground walking, footwear was manipulated on four levels (Table 13) during two separate testing sessions. To determine differences between conditions, a 4 (condition) x 2 (gender) repeated measures MANOVA was performed for twelve dependent variables (Table 13). A significant (*$p<0.001$)* Box's M test was found, therefore, the application of the more robust Pillai's Trace multivariate test was utilized for interpretation of output statistics. The multivariate repeated measures ANOVA with a *p* value set *apriori* at < 0.05 yielded a non-significant Gender*Condition interaction. Furthermore, while there was a non-significant Gender effect ($\Lambda_{Pillai} = 0.369$, $F(12,36) = 1.757$, $p = 0.095$, $\eta^2 = 0.369$, $Power = 0.764$) and Condition effect ($\Lambda_{Pillai} = 0.835$, $F(36,12) = 1.689$, $p = 0.166$, $\eta^2 = 0.835$, $Power = 0.625$)**.**

Table 13 – sEMG Variables: Footwear Effects

Independent Variables	**Dependent Variables**
Gender	Peak sEMG during Stance Phase
- Male	- Peroneus longus (PerL peak)
- Female	- Soleus (SOL $_{peak}$)
Condition	- Biceps femoris (BF $_{peak}$)
- Shod (SH)	- Right Gluteus medius (rGM $_{peak}$)
- Barefoot (BaFt)	- Left Gluteus medius (lGM $_{peak}$)
- Insole only (IN)	- Erector spinae (ES $_{peak}$)
- Insole with shoe (INSH)	Onset of Muscle Activity at Foot Strike
	- Peroneus longus (PerL onset)
	- Soleus (SOL $_{onset}$)
	- Biceps femoris (BF $_{onset}$)
	- Right Gluteus medius (rGM $_{onset}$)
	- Left Gluteus medius (lGM $_{onset}$)
	- Erector spinae (ES $_{onset}$)

Surface Electromyography: Condition Differences

Because of the non-significant repeated measures MANOVA condition effect, follow-up ANOVAs were contraindicated, indicating the different footwear conditions had no effect on any of the sEMG dependent variables (PerL $_{peak}$, Sol $_{peak}$, BF $_{peak}$, rGM $_{peak}$, lGM $_{peak}$, ES $_{peak}$, PerL $_{onset}$, Sol $_{onset}$, BF $_{onset}$, rGM $_{onset}$, lGM $_{onset}$, ES $_{onset}$).

Surface Electromyography: Gender Differences

Because of the non-significant repeated measures MANOVA gender effect, follow-up ANOVAs were contraindicated, indicating gender had no effect on any of the sEMG dependent variables (PerL $_{peak}$, Sol $_{peak}$, BF $_{peak}$, rGM $_{peak}$, lGM $_{peak}$, ES $_{peak}$, PerL $_{onset}$, Sol $_{onset}$, BF $_{onset}$, rGM $_{onset}$, lGM $_{onset}$, ES $_{onset}$) when cadence was altered.

Chapter V. Discussion

The purpose of this study was to better understand the effects of altered spatiotemporal variables and footwear on relative phase differences and muscle activation during walking. Specifically, this study investigated : (1) the effects of velocity and cadence on lumbopelvic rhythm during overground walking; (2) the effects of footwear on lumbopelvic rhythm during gait; (3) the effect of enhanced tactile feedback (i.e. barefoot, textured insole) on the onset of muscle activity at foot strike; (4) the effect of spatiotemporal and footwear variation on the magnitude of lumbar (i.e. erector spinae) and lower extremity muscular activity (e.g. gluteus medius, biceps femoris, soleus, peroneus longus) during the stance phase of gait, and (5) differences in the kinematics and electromyography between males and females. The following chapter has been divided into five sections, with the initial three sections addressing the overall research questions. Section 1 discusses the effect of walking velocity on relative phase differences and muscle activation patterns. Section 2 discusses the effect of cadence on relative phase differences and muscle activation. Section 3 discusses the effect of footwear and gender on relative phase differences and muscle activation patterns. Section 4 provides a summary of the combined findings in light of the five research questions. Section 5 discusses final conclusions from the present research and recommendations for future research.

Section 1: Walking Velocity Effects

The purpose of this section of the study was to investigate the effect of altered velocity on spatiotemporal variables, relative phase differences, and electromyography at three velocities (Normal, 125% of Normal, 75% of Normal velocity). The kinematic variables of interest were height normalized stride length (SL), cadence, relative phase mean differences for the trunk-pelvis, trunk-leg, and pelvis leg (TP_{mean}, TL_{mean}, PL_{mean}, respectively), relative phase standard deviation differences for the trunk-pelvis, trunk-leg, and pelvis-leg (TP_{sd}, TL_{sd}, PL_{sd}, respectively), and angular positions of the pelvis, trunk, and leg at foot strike. Table 14 illustrates the spatiotemporal variables and relative phase differences for the altered velocity conditions as compared to the normal velocity condition.

Table 14 - Effect of Velocity on all Kinematic Variables as Compared to Normal

Variable	125%	75%
SL	↑	↓
Cadence	↑	↓
TP_{mean}	↓	↑
TP_{sd}	↔	↔
TL_{mean}	↔	↔
TL_{sd}	↔	↔
PL_{mean}	↔	↔
PL_{sd}	↔	↔
θ_{pelvis}	↔	↓
θ_{thorax}	↔	↓
θ_{leg}	↔	↔

Note. ↔ = no significant difference, ↑ = increase, ↓ = decrease

All indicated differences were significantly different (p <0.05)

The sEMG variables of interest were the normalized peak activity during the stance phase of the peroneus longus ($PerL_{peak}$), soleus (Sol_{peak}), biceps femoris (BF_{peak}), right gluteus medius (rGM_{peak}), left gluteus medius (lGM_{peak}), and erector spinae (ES_{peak}). Furthermore, the time to onset muscle activity for the same six muscles at foot strike ($PerL_{onset}$, Sol_{onset}, BF_{onset}, rGM_{onset}, lGM_{onset}, and ES_{onset}) were also investigated. Table 15 illustrates the sEMG (normalized peak and time to onset for the respective six muscles) differences for the altered velocity conditions as compared to the normal velocity condition.

Table 15 - Effect of Velocity on all sEMG Variables as Compared to Normal

Variable	125%	75%
PerL $_{peak}$	↑	↓
Sol $_{peak}$	↑	↔
BF $_{peak}$	↑	↓
rGM $_{peak}$	↑	↓
lGM $_{peak}$	↑	↓
ES $_{peak}$	↑	↓
PerL $_{onset}$	↔	↔
Sol $_{onset}$	↔	↔
BF $_{onset}$	↔	↔
rGM $_{onset}$	↔	↔
lGM $_{onset}$	↔	↔
ES $_{onset}$	↔	↔

Note. ↔ = no significant difference, ↑ = increase, ↓ = decrease

All indicated differences were significantly different (p <0.05)

Condition Differences in Spatiotemporal & Kinematic Variables During Altered Velocity

The present study found a significant difference in cadence between all velocity conditions (Figure 16), indicating that as velocity was increased from 75% of Normal, to Normal, to 125% of Normal, participants adjusted cadence by increasing the number of steps per minute. The present study also found a significant difference in height normalized stride length between all velocity conditions (Figure 17), indicating that as velocity was increased from 75% of Normal, to Normal, to 125% of Normal, participants adjusted stride length by taking longer height normalized strides. It was anticipated that as the requested velocity increased, the participants would assume either a cadence or stride length dominant strategy to achieve the imposed speed; however, it appears that participants adjusted both variables, to varying degrees, to achieve the chosen requested velocity. To better understand the results of the present study, an investigation of whether participants achieved the increased normalized stride length by increased hip flexion/knee extension or pelvic girdle rotation was considered. As indicated in Figure's 21 & 22, there was a significant difference the amount of pelvic rotation and trunk rotation at foot contact, suggesting that the increase in normalized stride length may have resulted from a greater pelvic rotation during terminal swing. Conversely, as pelvic rotation was greatest during the Normal condition, normalized stride length may have been achieved by a longer excursion of the center of mass prior to foot contact. Furthermore, the increased excursion must have occurred through an exaggerated toe-off which assisted in moving the foot to move forward more rapidly (i.e. increased cadence), yet the body displayed similar alignments at foot contact. This supposition is further supported by the sEMG findings as the soleus produced, relative to the Normal condition, a diminished excitation during the 75% of Normal condition and an increased excitation during the 125% of Normal. Furthermore, the significant increase in peak peroneus longus activity would also support this supposition, as the participant would presumably attempt to rapidly evert the foot to better load the head of the metatarsals (e.g. ball of the foot) and the hallux (e.g. big toe) during toe-off. As velocity (and cadence) increased, the rate of eversion would occur faster due to shorter absolute ground contact times. Support of this theory also comes from the sEMG of the BF_{peak}, which indicated a significant increase in peak activity as velocity was increased. Conversely, the increase in biceps femoris activity may have resulted from the longer normalized stride length as velocity increased. Of course, in order to draw a definitive conclusion, joint angles at toe off and kinetic data would need to be investigated.

In the present study, TP_{mean} decreased as the velocity increased from 75% to 125%, indicating a more in-phase relationship between the trunk and pelvis during higher velocities

(Figure 20). These findings are opposite of previous research examining the effect of velocity on trunk-pelvis coordination during walking (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; Wagenaar & van Emmerik, 2000). Several factors may account for the difference in findings between previous research and the present study. First, the velocities utilized in the present study were much larger than those employed in previous studies (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002). Second, to the knowledge of the author, the present study is the first to examine the relative phase differences during overground walking with all previous research completed on treadmill walking. Though debated in the literature, several studies have noted significant biomechanical differences, particularly at the knee and hip, in treadmill walking relative to overground walking (Alton, Baldey, Caplan, & Morrissey, 1998; Dingwell, Cusumano, Cavanagh, & Sternad, 2001; S. J. Lee & Hidler, 2008; Parvataneni, Ploeg, Olney, & Brouwer, 2009). Furthermore, Dingwell et al. (2001) demonstrated significant reductions in both variability and local dynamic stability in treadmill walking relative to overground walking. Interestingly, in a study examining pelvic movements during overground walking, Crosbie and Vachalathiti (1997) reported that walking velocities above normal tended to produce a tight phase locking of movement patterns. So, though the findings of the present study do not coincide with the findings of previous outcomes during treadmill walking, the present study did support the findings of other work performed on overground walking.

The present study also found a significant increase in bilateral gluteus medius (Figure’s 25 & 26) and erector spinae (Figure 27) activity during stance phase as velocity increased from 75% of Normal conditions relative to other conditions. This finding is interesting considering the more synchronous relationship between the trunk and pelvis that was found in the present study; however, the increased activity of the lumbopelvichip (LPH) musculature may have arisen from two separate areas. As stride length and velocity increased, step width (not reported) would have decreased, placing additional demands on the respective GM to assist in maintaining frontal plane pelvic neutrality during single-limb support (Perry, 1992; Whittle, 2003). Increased LPH activity may have also resulted from the phase-locking and increased co-contraction of lumbopelvichip musculature, which has been previously hypothesized to occur during overground walking at greater than normal velocities (Crosbie & Vachalathiti, 1997). The present study has found that the trunk and pelvis remain more in-phase as velocity (normalized stride length and cadence) increased. This more in-phase relationship also supports the supposition that the body is partially using a forceful toe-off to meet the demands of the imposed velocity rather than additional pelvic rotation a velocities above Normal, which would cause the upper body to counter rotate to conserve total body angular momentum.

Section 2: Walking Cadence Effects

The purpose of this section of the study was to investigate the effect of altered cadence on spatiotemporal variables, relative phase differences, and electromyography at three levels (Normal, 125% of Normal, 75% of Normal cadence). The kinematic variables of interest were height normalized stride length (SL), walking velocity, relative phase mean differences for the trunk-pelvis, trunk-leg, and pelvis leg (TP_{mean}, TL_{mean}, PL_{mean}, respectively), relative phase standard deviation differences for the trunk-pelvis, trunk-leg, and pelvis-leg (TP_{sd}, TL_{sd}, PL_{sd}, respectively), and angular positions of the pelvis, trunk, and leg at foot strike. Table 16 illustrates the spatiotemporal variables and relative phase differences for the altered cadence conditions as compared to the normal cadence condition.

Table 16 - Effect of Cadence on all Kinematic Variables as Compared to Normal

Variable	125%	75%
SL	↑	↓
Velocity	↑	↓
TP_{mean}	↔	↑
TP_{sd}	↔	↔
TL_{mean}	↑	↑
TL_{sd}	↔	↔
PL_{mean}	↑	↑
PL_{sd}	↔	↔
θ_{pelvis}	↑	↔
θ_{trunk}	↑	↔
θ_{leg}	↑	↔

Note. ↔ = no significant difference, ↑ = increase, ↓ = decrease

All indicated differences were significantly different ($p < 0.05$)

The sEMG variables of interest were the normalized peak activity during the stance phase of the peroneus longus ($PerL_{peak}$), soleus (Sol_{peak}), biceps femoris (BF_{peak}), right gluteus medius (rGM_{peak}), left gluteus medius (lGM_{peak}), and erector spinae (ES_{peak}). Furthermore, the time to onset muscle activity for the same six muscles at foot strike ($PerL_{onset}$, Sol_{onset}, BF_{onset}, rGM_{onset}, lGM_{onset}, and ES_{onset}) were also investigated. Table 17 illustrates the sEMG (normalized peak and time to onset for the respective six muscles) differences for the altered cadence conditions as compared to the normal cadence condition.

Table 17 - Effect of Cadence on all sEMG Variables as Compared to Normal

Variable	125%	75%
PerL $_{peak}$	↑	↔
Sol $_{peak}$	↑	↓
BF $_{peak}$	↑	↔
rGM $_{peak}$	↑	↔
lGM $_{peak}$	↑	↔
ES $_{peak}$	↑	↔
PerL $_{onset}$	↔	↔
Sol $_{onset}$	↔	↔
BF $_{onset}$	↔	↔
rGM $_{onset}$	↔	↔
lGM $_{onset}$	↔	↔
ES $_{onset}$	↔	↔

Note. ↔ = no significant difference, ↑ = increase, ↓ = decrease

All indicated differences were significantly different (p <0.05)

Condition Differences in Spatiotemporal & Kinematic Variables During Altered Cadence

The present study found a significant difference in velocity between all cadence conditions (Figure 28), indicating that as velocity was increased from 75% of Normal, to Normal, to 125% of Normal, participants adjusted velocity by continuing to walk faster. As with the altered velocity conditions, significant differences in height normalized stride length between cadence conditions (Figure 29) were found, indicating that as cadence was increased from 75% of Normal, to Normal, to 125% of Normal, participants adjusted stride length by taking longer height normalized strides. Similar to when velocity was altered, the present section demonstrated significant increases in PL $_{peak}$ (Figure 37) and Soleus $_{peak}$ (Figure 38) as cadence increased. A similar hypothesis as proposed to explain the changes noted during the imposed velocity condition is believed to have contributed to the significant difference in the behavior of the peroneus longus and soleus during the stance phase. It is believed that the significant increases in the peroneus longus, acting as a plantarflexor and evertor of the foot during the stance phase, and the soleus, performing plantarflexion of the foot during the stance phase, assisted in an increase in propulsion and as a result a greater excursion of the center of mass. Specifically, it is believed the participant would need to rapidly evert the foot to better load the head of the metatarsals (e.g. ball of the foot) and the hallux (e.g. big toe) during toe-off. As velocity (and cadence) increased, the rate of eversion would occur faster due to shorter absolute ground contact times, thereby explaining the increases in peroneus longus activity as the higher cadence levels. Similarly, the strong plantarflexion moment created by the soleus during the stance phase of gait, particularly terminal stance, would assist in the center of mass excursion.

Tables 18, 19, and 20 display the height normalized stride length, velocity, and cadence, respectively for the altered velocity and altered cadence conditions. Though differences in the dependent variables between conditions (i.e. velocity vs. cadence) were not tested for significance, values are displayed for comparison purposes. For instance, while normalized stride lengths were almost equal during self-selected trials (Normal), differences were observed at the 75% and 125% trials, with an almost 2% difference in normalized stride length between 125% of Normal Velocity and 125% of Normal Cadence (Table 18).

Table 18 - Changes in Normalized Stride Length During Altered Velocity & Altered Cadence

	Velocity Conditions	Cadence Conditions
125% of Normal	90.5%	88.6%
Normal	80.7%	80.8%
75% of Normal	71.3%	70.6%

Note. Reported values are a percentage of standing height

When considering the preferred walking velocity (Normal) of the participants, velocity was once again almost identical between testing sessions; however, large differences were found between the 125% and 75% conditions, with the largest differences exhibited between 125% of Normal Velocity and 125% of Normal Cadence (Table 19). Similar to the findings for normalized stride length, this suggests participants reacted differently when participants were free to choose a cadence and stride length combination (i.e. altered velocity) to achieve the required velocity than when cadence was pre-determined. Furthermore, when cadence was altered from 75% of Normal to 125% of Normal, a considerably greater range (from 75% of Normal to 125% of Normal) is noted.

Table 19 - Changes in Velocity During Altered Velocity & Altered Cadence

	Velocity Conditions	Cadence Conditions
125% of Normal	1.59	1.76
Normal	1.28	1.30
75% of Normal	0.96	0.87

Note. Reported values are in ms^{-1}

As with velocity and cadence, the preferred walking cadence (Normal) was similar between testing sessions, large differences were found between the 125% and 75% conditions, with the largest differences exhibited between 125% of Normal Velocity and 125% of Normal Cadence (Table 20). Rather than adjusting stride length to a greater extent to adjust for the larger variation in cadence during the cadence conditions, participants appeared to not adjust stride length outside of an internally selected range, creating larger fluctuations in velocity (Table 19) than was observed during the altered velocity conditions; however, a limit may have been reached at the upper limits (125% of Normal Cadence) as participants displayed a 2% lower normalized stride length during the faster condition (Table 18).

Table 20 - Changes in Cadence During Altered Velocity & Altered Cadence

	Velocity Conditions	Cadence Conditions
125% of Normal	122.9	139.6
Normal	111.3	112.5
75% of Normal	93.7	84.7

*Note. Reported values are in steps*minute^{-1}*

It is hypothesized that the dual-task nature of walking and coordinating walking patterns to a metronome altered walking patterns as compared to the single-task nature of the altered velocity conditions. While the altered cadence condition was successful in achieving the 125% and 75% of Normal cadences, greater variability in the kinematics of the trunk, pelvis, and right thigh at foot strike were noted. As opposed to the altered velocity conditions, significant main effects were found during the altered cadence for all six kinematic variables of interest. For instance, during the altered cadence condition, the trunk and pelvis were found to be most synchronous (Figure 30) and least variable (Figure 31) during the 125% of Normal condition. Conversely, the relationship between the trunk and leg (Figure 32), as well as the pelvis and leg (Figure 33), indicated greater asynchronicity during the 125% of Normal and 75% of Normal conditions as compared to Normal. Therefore, it is hypothesized that the increasing velocity that occurred as cadence was increased, and accompanied increases in normalized stride length, altered the relationship between the trunk and pelvis during overground walking. Conversely, the larger variation in cadence that occurred during the altered cadence conditions altered the relationship between the trunk and leg as well as the pelvis and leg.

These findings are interesting in light of the of the other kinematic results which indicated that the pelvis exhibited right pelvic girdle rotation and left trunk rotation at foot strike, as opposed to left pelvic girdle rotation (Figure 34) and right trunk rotation (Figure 35) as during the Normal and 75% conditions. It is believed that the dual-task activity of walking to a metronome, which concurrently fixed a primary spatiotemporal variable, significantly altered the timing of gait events and joint kinematics. For instance, no metronome was utilized during the Normal conditions as these trials served as baseline trials. Therefore, the only trials in which the metronome was utilized was during the 125% of Normal and 75% of Normal cadence conditions. Conversely, results of the present section indicated that upper leg displayed greater external rotation at foot strike during the 125% of Normal cadence condition as compared to Normal and 75% of Normal (Figure 36), a finding similar to the altered velocity section. It is believed that the altered trunk and pelvis positions at foot strike necessitated considerably larger external rotation of the upper leg during the higher cadence level due to the increased velocity and narrower step width which occurs as velocity increases (Perry, 1992; Whittle, 2003).

The present study also found a significant increase in LPH musculature (e.g. biceps femoris, bilateral gluteus medius, erector spinae) activity during the 125% of Normal conditions relative to other conditions as cadence was increased (Figure’s 39-42). This finding is similar to the increased LPH that was observed as velocity increased as discussed in Section 1. It is believed that the increased LPH activity may have arisen from two separate areas. The primary hypothesis is that as stride length and cadence increased, step width (not reported) would have also decreased, placing additional demands on the respective gluteus medius to assist in maintaining frontal plane pelvic neutrality during single-limb support (Perry, 1992; Whittle, 2003). Support for this theory also comes from the significantly altered relationship between the pelvis and leg during the altered cadence conditions relative to Normal. Increased gluteus medius and erector spinae activity may have also resulted from phase-locking of the trunk and pelvis, which may have resulted in increased co-contraction of LPH musculature. Phase locking of the pelvis has been previously hypothesized to occur during overground walking at greater than normal velocities (Crosbie & Vachalathiti, 1997).

Section 3: Footwear Effects During Walking

Previous research on relative phase differences has primarily focused on changes in synchronicity as a result of alterations in velocity, the author is unaware of any study that investigated the effect of footwear on these variables (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; van Emmerik & Wagenaar, 1996). This may be important due to recent research highlights of significant mechanical differences in shod gait relative to barefoot walking (Hamill, Russell, Gruber, & Miller, 2011; Lieberman et al., 2010; Majumdar, Banerjee, Pal, Kumar, & Selvamurthy, 2006; Morio et al., 2009; Wegener et al., 2011). Though the reasons for these differences are not entirely understood, the literature appears to support a neural interaction (i.e. reduced tactile feedback) rather than a purely mechanical (i.e. increased mass of the distal segment) (Divert et al., 2005; Divert et al., 2008; Maki et al., 1999; Manor & Li, 2009; Nurse et al., 2005; Nurse & Nigg, 2001). The mass of the footwear utilized in the present study was an average per foot of 143g for SH, 162g for INSH, and 31.5g for IN. While the shod conditions (SH, INSH) were similar to the lightest footwear option utilized by Divert and colleagues (2008), the insole only condition was considerably lighter than anything presently reported in the literature.

Previous research has suggested it may be possible to alter the behavior of the musculature through the manipulation of footwear components or enhancement of tactile feedback (Murley et al., 2009; Nurse et al., 2005; Nurse & Nigg, 2001; Scott et al., 2012). Therefore, the purpose of this portion of the study was to investigate the effect of footwear on spatiotemporal variables, relative phase differences, and muscle activity during walking. Normal walking velocity and cadence were chosen for four randomized footwear conditions (SH, BaFt, IN, INSH) to examine differences in the chosen dependent variables. The kinematic variables of interest were height normalized stride length (SL), cadence, walking velocity, relative phase mean differences for the trunk-pelvis, trunk-leg, and pelvis leg (TP_{mean}, TL_{mean}, PL_{mean}, respectively), relative phase standard deviation differences for the trunk-pelvis, trunk-leg, and pelvis-leg (TP_{sd}, TL_{sd}, PL_{sd}, respectively), and angular positions of the pelvis, trunk, and leg at foot strike. Table 21 illustrates the spatiotemporal variables and relative phase differences as compared to the SH condition.

Table 21 - Effect of Footwear on all Kinematic Variables as Compared to Shod

Variable	BF	IN	INSH
SL	↓	↓	↔
Cadence	↑	↑	↔
Velocity	↓	↓	↔
TP_{mean}	↔	↔	↔
TP_{sd}	↔	↔	↔
TL_{mean}	↔	↔	↔
TL_{sd}	↔	↔	↔
PL_{mean}	↑	↑	↔
PL_{sd}	↔	↔	↔
θ_{pelvis}	↔	↔	↔
θ_{thorax}	↔	↔	↔
θ_{leg}	↔	↔	↔

Note. ↔ = no significant difference, ↑ = increase, ↓ = decrease

All indicated differences were significantly different ($p < 0.05$)

The sEMG variables of interest was the normalized peak activity during the stance phase of the peroneus longus ($PerL_{peak}$), soleus (Sol_{peak}), biceps femoris (BF_{peak}), right gluteus medius (rGM_{peak}), left gluteus medius (lGM_{peak}), and erector spinae (ES_{peak}). Furthermore, the time to onset muscle activity for the same six muscles at foot strike ($PerL_{onset}$, Sol_{onset}, BF_{onset}, rGM_{onset}, lGM_{onset}, and ES_{onset}) were also considered. Table 22 illustrates the sEMG (normalized peak and time to onset) differences as compared to the SH condition.

Table 22 - Effect of Footwear on all sEMG Variables as Compared to Shod

Variable	BF	IN	INSH
PerL $_{peak}$	↔	↔	↔
Sol $_{peak}$	↔	↔	↔
BF $_{peak}$	↔	↔	↔
rGM $_{peak}$	↔	↔	↔
lGM $_{peak}$	↔	↔	↔
ES $_{peak}$	↔	↔	↔
PerL $_{onset}$	↔	↔	↔
Sol $_{onset}$	↔	↔	↔
BF $_{onset}$	↔	↔	↔
rGM $_{onset}$	↔	↔	↔
lGM $_{onset}$	↔	↔	↔
ES $_{onset}$	↔	↔	↔

Note. ↔ = no significant difference, ↑ = increase, ↓ = decrease

All indicated differences were significantly different (p <0.05)

Footwear effects on Spatiotemporal & Kinematic Variables

While footwear was found to significantly affect walking velocity, with BaFt and IN conditions exhibiting a significantly lower velocity relative to shod matched condition (i.e. SH, INSH), this difference may be of minimal clinical or sport application has noted by the small effect size (Figure 44). Conversely, height normalized stride length was found to be significantly shorter, and cadence significantly higher, during the BaFt and IN condition as compared to SH and INSH. While the difference in these spatiotemporal parameters for barefoot and shod walking is supported by previous research, previous research has indicated enhanced tactile feedback to significantly alter postural stability and muscle activity during standing and overground walking (Majumdar et al., 2006; Maki et al., 1999; Nurse & Nigg, 2001; Romer et al., 2012; Wegener et al., 2011). The lack of a significant difference in normalized stride length between insole matched conditions (BaFt vs. IN; SH vs. INSH) appears to lend support for a possible inertial interaction of footwear on gait mechanics rather than reduced afferent feedback as the mass of the insole was negligible in the current design. Likewise, the significantly higher cadence during the BaFt conditions relative to the IN condition lends support for a mass effect of footwear, though admittedly the mass of the IN was minimal (31.5g).

Though significant footwear effects were noted in the walking velocity conditions, the small changes in velocity were a result of a significantly shorter normalized stride length of the BaFt and IN conditions which were accompanied by slight, significant increases in cadence. Wegener et al. (2011) reported similar differences in spatiotemporal variables (e.g. velocity, cadence, stride length) when examining barefoot vs. shod gait, while also reporting significant reductions in base of support, contact time, and double support time during barefoot walking. Therefore, it is believed the BaFt and IN conditions resulted in shorter relative stance phase and reduced ground contact time. As during the altered cadence conditions, which resulted in increased pelvis-leg asynchronicity at higher cadence levels, it is hypothesized that the altered ground contact time and cadence of the BaFt and IN conditions resulted in a significant difference in synchronicity between the pelvis and leg.

While footwear was found to not significantly alter the kinematics of any of the other segments investigated or the sEMG of the muscles investigated, the present section has advanced this area of research by examining the effects of footwear on relative phase differences and segment kinematics during normal, overground walking. The present study indicated that while footwear did not alter the level of synchronicity between the trunk-pelvis and trunk-leg, footwear did alter the level of pelvis-leg synchronicity. Furthermore, no differences in trunk, pelvis, or leg rotation were noted as a result of footwear, which may indicate that footwear does not have a significant effect on normal, overground walking. Instead, footwear effects on joint kinematics at foot strike may be more primary constrained to differences in hip, knee, and ankle sagittal plane kinematics which were not investigated in the present study.

Conversely, the shoe utilized for the present study (Adidas AdiPure Adapt, Portland, OR, USA) was of a minimalist, zero heel height differential which were designed by Adidas to resemble a bare foot. Therefore, the shoes utilized in the present study are dissimilar from most commercially available footwear, of which typically relies on thicker sole material, considerable heel height differentials, and relatively inflexible midsole material which reduces natural foot motion through the gait cycle. Though the Adidas AdiPure Adapt were chosen specifically to reduce the effect of typical footwear design on the dependent variables, additional investigation is needed to understand the effects of heel height differentials and midsole flexibility on joint kinematics and muscle activity during walking.

Gender Differences in Spatiotemporal & Kinematic Variables

Similar to the previous section which examined footwear effects, gender differences were noted when examining the effect of altered footwear during normal walking. Specifically, the present study found a significant gender difference for height normalized stride length when footwear was altered (Figure 46) with females exhibiting a significantly longer normalized stride length than males. Furthermore, the difference in normalized stride length between genders equaled approximately 3.5% of standing height. Therefore, it appears that females take longer height normalized strides than males during self-selected walking. The lack of a significant gender effect for trunk or pelvis joint kinematics suggests that females achieved the greater stride length through increased center of mass excursion, though no significant gender differences were noted for the sEMG of any of the muscles.

Significant gender differences for cadence (Figure 48) were also found, with females exhibiting a significantly higher cadence than males. Specifically, the difference in cadence between genders was approximately 8.4 steps*min^{-1}. Finally, females walked slightly, faster (1.29 ms^{-1}) than males (1.25 ms^{-1}); however, the difference in walking velocity was exceptionally small and was non-significant. Rather, it is believed that the significant difference in cadence is a result of more profound differences, particularly differences in anthropometrics (e.g. shorter leg length of females).

Results from the present study indicated a significant gender difference for TP_{SD} when footwear was altered (Figure 49), with males exhibiting a greater TP_{SD}, indicating males exhibited greater variability in the relationship between trunk and pelvis movements. Though TP_{mean} approached significance, with males exhibiting greater asynchronicity than females, this value did fail to reach the *apriori* alpha level. Nonetheless, the results of the present study suggest that males may be more variable than females in the movement of the LPH complex during normal walking.

Section 4: Summary of the Discussion

The purpose of this section is to summarize the findings of the present study while addressing the individual research questions. This summary includes three parts:

- Part 1: Summarize the discussion comparing the effects of altering velocity and cadence on spatiotemporal variables, relative phase measures, and sEMG.
- Part 2: Summarize the discussion comparing the effects of footwear on the spatiotemporal variables, relative phase measures, and sEMG.
- Part 3: Summarize the discussion of gender differences in spatiotemporal variables and relative phase measures.

Summary: Part 1

Spatiotemporal & Kinematic Variables

As velocity and cadence were controlled at 75% of Normal, Normal, and 125% of Normal there was a significant increase in the normalized stride length of the participants. Increases in stride length as velocity increased was not necessarily an unexpected result given that walking velocity is the product of stride length and cadence (Whittle, 2003; Wyke, 1967). Therefore, concurrent increases in stride length and cadence would be expected as velocity is increased (Blanc et al., 1999; Perry, 1992). For instance, van Emmerik and Wagenaar (1996) found an increase in stride length as walking velocity was increased from 0.3 ms^{-1} to $1.3ms^{-1}$. Likewise, Crosbie et al. (1997b) found significantly longer height normalized step lengths when comparing freely chosen velocities at two levels: Normal and Faster than Normal (participants were instructed to walk as if they were late to a meeting).

In the present study, both velocity and cadence increased when manipulated from 75% of Normal to 125% of Normal, indicating that the author was successful in manipulating the independent variables for each session. However, both velocity and cadence increased significantly on the sessions in which it was a dependent variable as well. The latter results may have confounded the overall results, though it would have likely been impossible to simultaneously control for both spatiotemporal variables during overground walking. Therefore this is a limitation of the present study.

When examining the effect of altered velocity or altered cadence, significant differences were found for TP_{mean}, TP_{SD}, TL_{mean}, TL_{SD}, PL_{mean}, PL_{SD}, Pelvic Rotation, Trunk Rotation, and Leg Rotation; however the influence of velocity appeared to be smaller in which significant differences were found only for TP_{mean}, Pelvic Rotation, and Trunk Rotation. Furthermore, findings from the present study appear to contradict the findings of earlier studies, which found a decrease in the TP mean, or increase in the synchronicity of the trunk and pelvis, as velocity was increased. Of noteworthy importance though is the considerable difference in velocity ranges of the present study as compared to previous studies. For instance, van Emmerik and Wagennar's (1996) had participants walk on a treadmill at a pre-determined velocity, beginning at 0.3 ms^{-1} and ascending to 1.3 ms^{-1} in 0.1 ms^{-1} intervals before decreasing back to the original velocity. Conversely, the participants in the present study exhibited an average walking velocity between 0.87 $ms^{-}1$ and 1.76 ms^{-1}. Interestingly, the maximum walking velocity in the van Emmerick and Wagennar study (1996) was approximately the same as the Normal conditions in the present study. Lamoth, Beek, and Meijer (2002) published a similar paper examining trunk-pelvic coordination in the transverse plane during gait. Beginning at 0.39 ms^{-1}, walking velocity was increased in 0.22 ms^{-1} until reaching a maximum of velocity of 1.5 ms^{-1}, the authors reported a significant increase in the TP_{mean} as velocity increased, contradicting results of the present study. Again, though, both Lamoth et al. (2002) and van Emmerick and Wagennar (1996) performed their respective studies on treadmills using relatively rudimentary motion capture technology as compared to the present study.

Another component that may explain the differences in results is the methodological approach used in the present study, which relied on a randomization of high (125%) and low (75%) velocities and cadences based on the a participants self-selected gait patterns. Therefore, velocities were not pre-determined, nor were they altered in a predictable format as was performed by almost all of the previous researchers (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; Wagenaar & van Emmerik, 2000). Another possibility is the difference in kinematic collections between past research, which primarily relied on LED or selspot style systems, whereas the present project utilized a passive motion capture system and a marker set consisting of 44 markers (Table 2). Finally, while many of the previous articles utilized a discrete Fourier transform or non-normalized data in the construction of the relative phase plots, the present study normalized angular displacements and velocities prior to the construction of all relative phase plots (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002). Kurz & Stergiou (2002) reported that the normalization of phase angle calculations resulted in significant and profound differences in the determination of the relative phase values when compared to non-normalized data. Specifically, the authors reported RMS differences of between 22.7° to 70.6° when comparing non-normalized relative phase angles to normalized phase angles (Kurz & Stergiou, 2002). Additional investigations should be completed utilizing a standardized data reduction process, such as the one utilized in the present study, so that results are able to be more equally compared and discussed.

Finally, previous research has relied exclusively on motorized treadmills to manipulate walking velocities. Furthermore, these studies have manipulated walking velocity without reference to a participant's preferred walking strategy. As previously stated, several studies have noted significant biomechanical differences, particularly at the knee and hip, in treadmill walking relative to overground walking (Alton et al., 1998; Dingwell et al., 2001; S. J. Lee & Hidler, 2008; Parvataneni et al., 2009). Furthermore, previous research has demonstrated that treadmill walking resulted in changes to both variability and local stability relative to overground walking (Dingwell et al., 2001). Another importance difference in overground walking has been previously noted by Crosbie and Vachalathiti (1997), which reported that walking at velocities above normal tended to produce a tighter phase locking of movement patterns. Therefore, if the previous research failed to adequately manipulate the velocities of the participants, as the references were absolute and not scaled to the individual, the maximal treadmill walking speed may have barely reached normal walking speeds for the participants.

Surface Electromyography Variables

As hypothesized, normalized peak sEMG during the stance phase increased as velocity and cadence increased from 75% of Normal, to Normal, to 125% of Normal for most muscles. Interestingly, from the present study were the lack of significant difference between many of the 75% of Normal and Normal conditions, specifically the soleus during the altered velocity conditions, and the peroneous longus, biceps femoris, bilateral gluteus medius, and erector spinae during the altered cadence conditions. The results may indicate the level of muscle activity required during the gait cycle, from weight acceptance to toe-off, is not significantly different during the lower conditions (75%, Normal) due to a minimum amount of muscle activity required to perform the requisite tasks. Conversely, the dual-task of the altered cadence conditions may have significantly altered single and support times, as suggested by Tables 18 – 20, altering muscle activity during 75% cadence trials.

It is believed the increase in biceps femoris activity and erector spinae activity during the stance phase is partially due to the increased center of mass excursion that is occurring at higher velocities, chiefly an increase in normalized stride length (Table 18). Conversely, the increase in the muscle activity for the aforementioned multi-articular muscles may be due to the significant alterations in segment synchronization and rotational amplitude as velocity and cadence increased. Specifically, though rotational amplitudes for the trunk and pelvis tended to increase as velocity and cadence increased, there was also a corresponding increase in the synchronization of the trunk-pelvis. It is believed that this increase synchronization may have occurred through a type of phase locking, which Crosbie and Vachalathiti (1997) has hypothesized to occur at greater than normal walking velocities. The present author asserts that this phase locking may be increasing the co-contraction of the musculature of the LPH complex.

Finally, no significant difference was noted in the onset of muscle activity for any of the six muscles observed in the present study. Future research may want to examine the time-to-peak muscle activity, which may be more sensitive to variation in loading rate at foot strike when velocity and cadence are altered.

Summary: Part 2

Spatiotemporal & Kinematic Variables

Results indicated that footwear had a significant effect on the preferred walking velocity, cadence, and normalized stride length of the participants. Specifically, the barefoot and barefoot-like condition (IN) exhibited a significantly shorter normalized stride length relative to both SH and INSH. Furthermore, the matched footwear conditions (BaFt and IN, SH and INSH), exhibited no significant difference between each other, indicating the textured insole had no significant effect on normalized stride length. Cadence was greatest for the BaFt condition, which exhibited significant differences from SH, IN, and INSH. Finally velocity was lowest for the barefoot and barefoot-like condition (IN), as compared to both SH and INSH. Results therefore would appear to support the increased mass of the distal segment when wearing shoes is the cause for an increased stride length relative to barefoot (Majumdar et al., 2006); however, other studies appear to discount this hypothesis (Divert et al., 2005; Divert et al., 2008; Maki et al., 1999; Nurse et al., 2005; Shroyer & Weimar, 2010). While the mass of the shod conditions in the present study were minimal (SH = 143g, INSH = 162g), even this slight difference may be enough to result in alterations in the timing of key gait events, particularly foot strike. Future research should be expanded to include greater manipulation of the mass of a shoe, variation in texture designs, and the effect of heel height differentials. Additionally, future research and analyses should examine ground reaction forces (e.g. impact transients) as well as lower extremity joint kinematics as the use of stride length alone may be hiding true mass or sensory effects.

The only kinematic variable to display a significant footwear effect was PL_{mean} with BaFt and IN exhibiting greater asynchronicity than the SH and INSH conditions. Though the present study is the first known study to examine the footwear effects on relative phase differences, the results appear to support research by Huang and colleagues (2010), which found that participants exhibited a more-in phase PL_{mean} when participants took significantly longer strides, with a concurrent reduction in cadence, while walking at a fixed velocity on a treadmill. Therefore, results from the present study suggest that an increase in stride length and reduction in cadence will create a more in-phase pelvis-leg relationship.

Surface Electromyography Variables

Interestingly, no significant effects were found for normalized peak muscle or onset of muscle activity for any of the six muscles observed during the study. The lack of significant difference in normalized peak muscle activities was unexpected considering previous researchers have noted significant footwear effects on peak muscle activity (Bird et al., 2003; Gefen et al., 2002; Li & Hong, 2007; Murley et al., 2009; Nurse & Nigg, 2001). The results may have been confounded by the minimalist nature of the footwear design, which is extremely flexible, light, and marketed as a minimalist running shoe with a zero heel height differential.

Finally, no significant effect was found for footwear for the time to onset of muscle activity for any of the muscles observed. Nurse et al. (2005) has previously shown a significant difference in the time to onset when comparing a smooth vs. textured insole during walking, however, the authors in that study relied on a wavelet analysis of the EMG signal (Nurse et al., 2005). Conversely, the present studied maintain the EMG signal in the temporal component, utilizing a double-threshold to determine onset. As suggested by Murley et al. (2009), it is difficult to compare findings from wavelet analyses to traditional EMG analysis. However, several other authors have noted a significant effect of increased afferent feedback in altering postural or gait kinematics (Bird et al., 2003; Kelleher et al., 2010; Maki et al., 1999; Nurse et al., 2005; Nurse & Nigg, 2001). For instance, Maki and colleagues (1999) reported that the enhanced plantar facilitation on the periphery of the foot reduced COP trajectory and time to stabilization in response to unpredicted postural perturbations. In a study utilizing the same textured insoles as the present study, Hatton et al. (2011) reported a 9.2% reduction in mediolateral sway in older adults during a static, eyes closed test. Future research should be directed towards texture variation as well as the examination of adaptation periods to better understand the augmented afferent feedback on postural stability, gait kinematics, and muscle activity.

Summary: Part 3

Spatiotemporal & Kinematic Variables

In the present study, females exhibited a significantly longer normalized stride length than males for the altered footwear condition only. Though unusual, Crosbie and colleagues (1997b) reported a significant gender difference in height normalized step length during a faster than normal walking condition, with females exhibiting a significantly longer step than males. Similarly, females were found to display a significantly higher cadence than males altered footwear conditions. Noteworthy though is the lack of a significant difference in walking velocity, suggesting that the significantly longer normalized strides of females may be resulting from differences in center of mass excursion techniques. Likewise, as discussed in Section 3, the difference in normalized stride length was small, with a small effect size, and moderate power.

Relative phase differences were only found to be significantly different during the footwear conditions, with males exhibiting significantly greater TP_{SD} indicating that the trunk and pelvis of males was significantly more variable than for females when footwear was altered. As previously stated, the findings are especially interesting considering that women exhibited a greater normalized stride length, which has been previously hypothesized to result in a more-out-of phase relationship between the trunk and pelvis; however, all of those previous research studies have been conducted utilizing various differences in methodology including treadmill walking and velocities not scaled to the preferred spatiotemporal variables (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; van Emmerik & Wagenaar, 1996). Finally, no significant gender differences were found for the angular position of the trunk, pelvis, or leg at foot strike.

The present study is the first known study to examine gender differences in the aforementioned kinematic variables. Unique to the present study is the finding that healthy males tend to display greater asynchronicity between the trunk and pelvis, which may be related to an increased variability of the pelvis during overground walking, though the source for these gender differences are not yet understood. Future research should be directed towards understanding if this relationship is similar across various footwear designs and age groups (i.e. adolescents, older adults, elderly). Additionally, future research should be directed towards studying the source of the gender differences in the relative phase measures, which may aid in understanding the clinical relevance of the measures.

Section 5: Future Research

Future Research for Velocity & Cadence Effects

When examining relative phase differences, the present focus within the literature has been on the influence of velocity on the relationship between trunk, pelvis, and leg movements (Bruijn et al., 2008; Y. Huang et al., 2010; C. J. Lamoth, Beek, et al., 2002; van Emmerik & Wagenaar, 1996). Though it has long been hypothesized that longer stride lengths would result in more asynchronous movements, only recently have researchers attempted to separate the variables that compose velocity (Y. Huang et al., 2010). Similarly, the present project manipulated velocity and cadence during a previously unstudied task, overground walking, in an attempt to gain a better understanding of the segmental rotations during walking. While stride length does appear to be associated with an altered trunk-pelvis relationship, cadence also appears to be an important factor in the trunk-leg and pelvis-leg movements. Further researcher should be directed towards the manipulation of stride length and cadence to better gauge the independent contributions of these variables to segmental synchronicity. Additionally, future research should be directed towards gaining a better understanding of the clinical relevance of the relative phase measures. Finally, there is a need to develop an improved methodological approach for the manipulation of velocity, cadence, and stride length, thereby minimizing the cognitive demands possibly associated with altering spatiotemporal variables during overground walking.

Future Research for Footwear Effects

The present study was one of the first to research the effect of footwear on relative phase differences, during treadmill or overground walking. Minimal differences were found except for spatiotemporal differences which have generally been accepted within the literature (Bird et al., 2003; Lieberman et al., 2010; Morio et al., 2009; Murley et al., 2009; Sato et al., 2012; Scott et al., 2012; Wegener et al., 2011). In an attempt to isolate only the influence of the foot in a shoe and not characteristics of the shoe, a minimalist, zero-drop shoe was chosen. Unfortunately, this footwear condition may not have had the necessary inertia or sole rigidity to produce previously noted footwear differences. On the contrary, it may be that while shoes result in significant differences in lower extremity and some muscle activity, these differences are not realized in a variable such as relative phase measures or time to onset of muscle activity. Though the shoe chosen for the present study was identified for the minimalist nature, constraints to use for the general population or many occupational tasks are contraindicated. Therefore, many questions remain as to whether a more traditional shoe design will significantly alter the relationship between trunk, pelvic, and hip movements or the efferent response of the lower extremity.

Future Research for Gender Differences

Interestingly, males and females appeared to display the greatest differences when footwear effects were examined. Specifically, males appeared to be more asynchronous and variable in trunk and pelvic. Past research has primarily focused on pathological populations, such as individuals with low back pain, or females with pelvic girdle related pain (Crosbie & Vachalathiti, 1997; Crosbie et al., 1997a; Y. Huang et al., 2010; Y. P. Huang et al., 2011; C. J. Lamoth, Meijer, et al., 2002; W. Wu et al., 2002; W. Wu et al., 2004; W. H. Wu et al., 2008). In the present study, though females displayed a longer normalized stride length and reduced variability in trunk-pelvic movements. Therefore future research should be directed towards gaining a better understanding of the gender differences and the significance of the differences in healthy populations before attempting to extrapolate to clinical populations.

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Appendix A

Health Screening Questionnaire

Please read each question carefully and answer honestly. If you do not understand the question, please ask the investigator for clarification. Check the appropriate answer.

Participant Number: _____________

<u>**Yes**</u>	<u>**No**</u>	
____	____	1) Are you under the age of 19?
____	____	2) Are you over the age of 35?
____	____	3) Do you currently have an injury that prevents or hinders your ability to walk normally in shoes?
____	____	4) Do you currently have an injury that prevents or hinders your ability to walk normally when barefoot?
____	____	5) Have you had any injuries or surgeries to your lower extremity, pelvis, low back, or trunk during the past year?
____	____	6) Do you currently have any pain in your lower extremity, pelvis, low back, or trunk?
____	____	7) Do you have any reason to believe that your participation in this investigation may put your health or well-being at risk?
_____	_____	8) Have you ever been told you have an inner ear disorder?
_____	_____	9) Has your doctor ever said that you have heart trouble?
_____	_____	10) Are you currently taking any medication that you think might influence your ability to participate in this study?
_____	_____	11) Are you allergic to adhesives?

Appendix B

(NOTE: DO NOT SIGN THIS DOCUMENT UNLESS AN IRB APPROVAL STAMP WITH CURRENT DATES HAS BEEN APPLIED TO THIS DOCUMENT.)

INFORMED CONSENT
for a Research Study entitled
"Influence of Tactile Feedback on Trunk Coordination and Electromyography During Walking"

You are invited to participate in a research study to investigate the influence of tactile feedback on trunk coordination and electromyography during walking. The study is being conducted by Braden Romer under the supervision of Dr. Wendi Weimar, Associate Professor, in the Auburn University Department of Kinesiology. You were selected as a potential participant because you are between the ages 19 and 35, and your health condition might, through a pre-screening health questionnaire to follow, permit you to perform the conditions safely and successfully. The purpose of this study is to gain a greater scientific understanding of the variables responsible for alterations in trunk coordination. Furthermore, the effect of variable sensory inputs to the foot on changes in walking patterns will be investigated.

What will be involved if you participate? If you decide to participate in this research study, you will be asked to come to the Sports Biomechanics Laboratory (Memorial Coliseum 1127) for two testing sessions, lasting approximately 60 minutes each. Paperwork will only need to be filled out during the initial session. After meeting participation criteria, your mass, height, leg length and arch height will be measured for normalization purposes. Next you will be fitted for the proper shoe and insole size. During both sessions, you will then be asked to change into spandex shorts and spandex tank tops to minimize any potential interference from personal clothing. Researchers will need to identify six separate muscles (right erector spinae, right and left gluteus medius, right biceps femoris, right soleus, and right peroneus longus). Identification of the pertinent muscles will occur through a combination of palpation and manual muscle testing, specific to each muscle:

- The right erector spinae will be identified during a prone (face down), lumbar extension (bending backwards) task.
- The right gluteus medius will be identified with you lying on your left side while concurrently attempting a hip abduction, similar to a jumping jack, task.
- The left gluteus medius will be identified with you lying on your right side while concurrently attempting a hip abduction, similar to a jumping jack, task.
- The right biceps femoris will be identified during a prone (face down), hip extension (pushing your leg backwards) task.

Participant's Initials__________

- The soleus will be identified during a seated, plantarflexion (pushing toes down) task with your knees bent to 90°.
- The peroneus longus will be identified during a task involving seated plantarflexion and eversion (pushing toes down while flattening your foot).

Once all muscles have been appropriately identified, the skin area around the electrode placement site will be shaved, abraded, and cleaned with a 70% alcohol solution to reduce the likelihood of electrical impedance. Electrodes will be placed near the center of the muscle belly, and parallel to the direction of the muscle fibers for the muscles of interest. A ground electrode will be placed on the body aspect of the tibial tuberosity (upper shin). A thin, nylon type wrap will be placed over the electrodes and limb segment to assist in preventing electrode movement. A maximum voluntary isometric (MVIC) will then be performed and used for normalization purposes.

- An MVIC for the erector spinae will be recorded during a prone lumbar extension task (bending backwards).
- An MVIC for the right gluteus medius will be recorded during a left side-laying, hip abduction task (similar to a jumping jack).
- An MVIC for the left gluteus medius will be recorded during a right side-laying, hip abduction task (similar to a jumping jack).
- An MVIC for the biceps femoris will be recorded during supine (face up), hip extension task (pushing leg backwards).
- An MVIC for the soleus and peroneus longus will be recorded simultaneously during a seated, plantarflexion (pushing toes down) task with knees bent to 90°.

The final step of preparation will be to affix 37 markers to the skin and clothing of the participant with double-sided tape. Marker placement will be in accordance with the Vicon Plug-In-Gait model. Next, a static capture will be performed with you standing in the center of the capture volume, upright, and arms out to the side. During the initial visit, ordering of conditions will be randomized based on cadence (the number of steps per minute) or velocity (how fast you walk). During the first testing session (day), only one variable will be manipulated, while during the second testing session (day) the other factor will be manipulated. Next, footwear order will be randomized (barefoot, shoe, textured insole, shoe with textured insole). After a five minute footwear acclimation period, you will be asked to perform three trials of your preferred walking style through the center of the capture volume. Next, you will be asked to perform three additional trials at either 125% or 75% of free-walking cadence or velocity. After those three trials have been successfully performed, you will be asked to complete three additional trials at the other parameter (125% or 75%). A rest period of a minimum of five minutes will be provided to you after each the three conditions. After the rest period, your footwear will be changed and you will be provided a minimum of five minutes in which to acclimate to the specific footwear condition. The same data collection procedures (free-walking, 125% and 75% of velocity or cadence) will take place until successful completion for each of the four footwear conditions. The second day of testing will follow the exact same protocol as the first day, except there will not be a time period allocated to the completion of paperwork.

Participant's Initials__________

Are there any risks or discomfort? It is possible that you may sustain muscle soreness or a muscle injury. However, the risks associated with this study should not be any more than normally encountered in normal activities of daily living. Other risks include a possible adverse response to the adhesive on the electrodes and reflective markers; however, these risks are similar to risks associated with applying a Band-Aid. In addition, these risks should be minimized through the use of a health screening questionnaire. Last, the MVIC could result in a musculoskeletal injury such as a strain, sprain or muscle soreness, which is possible in any type of lifting or athletic activity. In the unlikely event that you sustain an injury from participation in this study, you will be required to assume full financial responsibility for your own medical care. Participants are responsible for any and all medical cost resulting from injury during the study.

Are there any benefits to yourself or others? There is no direct benefit to you; however, you may request a copy of the written results upon the completion of the study.

If you change your mind about participating, you can withdraw at any time during the study. Your participation is completely voluntary. If you choose to withdraw, your data can be withdrawn as long as it is identifiable. Your decision about whether or not to participate or stop participating will not jeopardize your future relations with Auburn University, Department of Kinesiology, or the Sport Biomechanics Laboratory.

You privacy will be protected. Any information obtained in connection with this study will remain confidential as the data will correspond solely to an identification number. Information obtained through your participation may be published in a scientific journal and/ or presented at a scholarly conference.

If you have questions about this study, *please ask them now* or contact Braden Romer at bhr0002@auburn.edu, Dr. Wendi Weimar at weimawh@auburn.edu, or by calling the Sport Biomechanics Laboratory at 334.844.1468. A copy of this document will be given to you to keep.

If you have questions about your rights as a research participant, you may contact the Auburn University of Human Subject Research of the Institutional Review Board by phone (334) 844-5966 or email at hsubjec@auburn.edu or IRBChair@auburn.edu.

HAVING READ THE INFORMATION PROVIDED, YOU MUST DECIDE WHETHER OR NOT YOU WISH TO PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATED YOUR WILLINGNESS TO PARTICIPATE.

____________________________	___________________________________
Printed Name Date	Investigator obtaining consent Date
________________________	___________________________________
Participant's signature	Printed Name