The Influence of High Heel Shoes and Toe Walking on Gait Kinematics and Kinetics

by

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Abstract

For generations, many women included high heel shoes as a part of their fashion wardrobe. Despite the claims of deleterious effects from wearing high heel, such as foot deformities, leg and back pain, and tendonitis, women still continue to buy and wear high heel shoes. Past research has reported that walking in high heel shoes affected spatiotemporal, kinematic, kinetic variables, as well as gastrocnemius pennation angles. Over the years, many inventors devised products aimed at relieving the effects of wearing high heel shoes. One product, Insolia®, claims to alter the position of the foot in the shoe so as to improve the posture of the individual while walking in high heel shoes. Non-pathological toe walking is similar to walking in high heel shoes in that an individual is walking with the foot in a plantar flexed position. The main difference between toe walking and high heel walking is that during toe walking, the heel of the individual does not make contact with the ground. While walking in high heel, an individual has the opportunity to make heel contact with the floor, if only through the heel of the shoe. This is the first study, to this author's knowledge, that will compare spatiotemporal, kinematic, and kinetic variables while walking under the following conditions: barefoot, high heel, high heel with insert, and toe walking. In addition, to the author's knowledge, this is the first study to compare the pennation angle of the medial gastrocnemius while standing under the above mentioned conditions. The purpose of this study was to: (1) investigate how walking with an elevated heel, such as walking in high heel shoes, walking in high heel shoes outfitted with Insolia® inserts, or toe walking affect the kinematic and kinetic variables of gait; (2) investigate if wearing high heel shoes with Insolia® inserts during gait brings electromyographic variables closer to those variables observed while engaged in barefoot gait; (3) investigate if wearing high heel shoes, wearing high heel shoes outfitted with Insolia® inserts, or standing in plantar flexion, as observed with toe walking, alters the pennation angle of the gastrocnemius.

Examination of footwear effects indicated that footwear had a significant effect on knee and hip flexion, distance from center of gravity to foot at initial contact, stride length, change in height, absolute change in height, braking in the anterior/posterior direction, and vertical ground reaction forces at heel strike transient, foot loading, and toe off. Specifically, the toe walking condition exhibited significantly smaller knee flexion at initial contact and through midstance than the barefoot or high heel and high heel with insert conditions. Results indicated no significant differences between barefoot and toe walking conditions when measuring hip flexion at initial contact and through midstance. However, barefoot and toe walking hip flexion was significantly smaller than that detected with the high heel and high heel with insert condition. Measurements of the horizontal distance from the center of gravity to the foot at initial contact revealed that the barefoot condition was significantly smaller than the high heel, high heel with insert, and toe walking conditions. Additionally, it was also revealed that the high heel and high heel with insert conditions afforded significantly smaller distance from center of gravity to foot at initial contact than the toe walking condition. Stride length was significantly longer in the barefoot condition than the other three footwear conditions. Change in height and absolute change in height were significantly smaller in the barefoot condition when compared to the other footwear conditions.

Results indicate that ground reaction forces in the anterior/posterior direction during the barefoot condition exhibiting significantly less braking when compared to high heel with insert and toe walking conditions. No significant differences were found between the barefoot and high heel conditions with regard to braking. Additionally, no significant differences were found in propulsion in the anterior/posterior direction. Vertical ground reaction force results indicated that the barefoot condition exhibited a significantly smaller heel strike transient than the high heel or high heel with insert conditions. Foot loading in the barefoot condition resulted in significantly ground reaction force than the other four footwear conditions. At toe off, the barefoot condition exhibited significantly less propulsive force than the toe walking condition.

The current study found that mean and peak muscle activation displayed similar results. Gastrocnemius muscle activation was significantly higher in the toe walking condition than in the high heel and high heel with insert conditions. The biceps femoris exhibited significantly higher muscle activation than that detected in the high heel condition. However, no significant differences were found between toe walking and high heel with insert. No significant differences in muscle activation were found in the tibialis anterior, rectus femoris, rectus abdominis, or erector spinae.

Finally, when examining the pennation angle of the medial gastrocnemius while standing under all four conditions, it was revealed that the barefoot condition exhibited a significantly smaller pennation angle than that observed while standing in high heel, high heel with insert, or toe walking condition. The toe walking condition exhibited a significantly larger pennation angle than the high heel or high heel with insert conditions. Interestingly, the high heel and high heel with insert conditions did not reveal significant differences amongst all of the variables measured.

Overall, the Insolia® insert did not significantly alter kinematic, kinetic, electromyographic, or pennation angle variables observed while walking in high heel as no significant differences were detected between the high heel and high heel with insert conditions. However, continuums were revealed in knee flexion, hip flexion at midstance, distance from center of gravity to foot contact, and gastrocnemius pennation angle. The observance of the aforementioned continuums indicate that altering the position of the foot while walking, through either high heel or toe walking, some kinematic variables and pennation angle of the gastrocnemius change when compared to barefoot gait. Toe walking appears to illicit more alterations in gait than when walking in high heel, thus, indicating that the toe walking condition may be more taxing to the body than the high heel condition. Additionally, the results from the current study can add to the body of knowledge for gait studies.

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ANOVA	Analysis of Variance
cm	centimeter
COG	Center of gravity
COGFC	Center of gravity at foot contact
COM	Center of mass
deg	degree
EMG	Electromyography
FC	Foot contact
GRF	Ground reaction force
IC	Initial contact
IS	Initial swing
kg	kilogram
LR	Loading response
m	meter
ms	millisecond
MSt	Mid-stance
MSw	Mid-Swing
MVIC	Maximum voluntary isometric contraction
PCSA	Physiologic cross-sectional area
PS	Pre-swing

TSt	Terminal stance
TSw	Terminal swing
VO_2	Volume of oxygen consumed

Chapter I

Introduction

If the amount of money spent on high heel shoes in 2011 is any indication, women love their high heel shoes to the tune of $38.5 billion. Furthermore, more than half of the shoes bought by women had a 3” heel or higher (Binkley, 2012). High heel shoes have been a part of the fashion industry for generations. Their popularity first came to the fore when Catherine de’ Medici married Henry II of France in 1533 (Smith, 1999). Except for a short period during the French Revolution, high heel shoes have remained, for the most part, a fixture in popular culture in one form or another with men and women (Linder & Saltzman, 1998).

Despite the popularity of high heel shoes, multiple researchers have argued that high heel shoes cause physical ailments such as foot deformities, (bunions and hammer toes), leg and back pain, and tendonitis (Linder & Saltzman, 1998; Snow, Williams & Holmes, 1992; Snow & Williams, 1994). These warnings did not appear to dampen the desire of women to wear high heel shoes. One study conducted at the turn of the 21st century described the importance that high heel shoes held with women (Joyce, 2000). This study reported that the women surveyed believed shoes were an important component of their appearance. Furthermore, a number of the women reported wearing uncomfortable shoes because of a perceived expectation to wear a particular style of footwear (Joyce, 2000). The survey also reported a small percentage of women placed fashion over function because they would wear uncomfortable shoes as long as the appearance was desirable (Joyce, 2000). While more than half of the women surveyed reported comfort trumped appearance, only a small percentage of those polled were willing to don unattractive shoes to help relieve foot maladies (Joyce, 2000). This emphasis on fashion leads one to consider that wearing high heel shoes is a cultural practice, while others believe that the height of their high heel shoes mirror their power or status in society (Smith, 1999). Economics may also play a role in who wears high heel shoes. According to an online survey conducted in 2013 by Beso.com in honor of Shoe Week 2013, 21% of the women who earned $150,000 a year wore high heel shoes on a daily basis to work while 71% of women who earned less than $40,000 a year never wore high heel shoes to work (Huffington Post, 2013).

This study was designed to investigate kinematic, kinetic, and neuromuscular variables associated with gait and to determine if there is was a continuum in various shod conditions. Specifically, this study observed barefoot, high heel, high heel with an orthotic insert, and toe walking. Therefore, walking, changes in neuromechanical variables observed when wearing high heeled shoes, pennation angle, use of orthotic inserts with high heeled shoes, and toe walking gait will be reviewed.

Walking

A major form of locomotion for humans is walking. Walking involves complex movements, including synchronization of limbs, which alternate between balance and propulsion of the body; all orchestrated for the purpose of forward progression (Perry & Burnfield, 2010; Whittle, 1996). Many endeavored to determine what allows the most efficient use of energy during gait. In 1953, Saunders et al., described how energy expenditure for normal gait was minimized through six factors: pelvic rotation, pelvic tilt, knee flexion in stance phase, foot and knee mechanisms, and lateral displacement of the pelvis. These factors, otherwise known as determinants, when working in conjunction with each other, assist the body to move forward in smooth, sinusoidal curves in the sagittal and transverse planes (Saunders, Inman, & Eberhardt, 1953).

Kinematic Alterations in Gait from Wearing High Heeled Shoes

Since high heel shoes arrived on the fashion scene more than 400 years ago, scientists and physicians researched the effects of wearing this type of footwear (Linder & Saltzman, 1998). Many of these studies explored kinematic and kinetic changes that occur while wearing high heel shoes by comparing stride length, cadence, velocity, foot pressure, posture, and ground reaction forces with contradictory results. For example, during over ground walking, when wearing high heel shoes it was found that self-selected walking speed decreased (Esenyel, et al., 2003; Barkema, Derrick, & Martin, 2012; Opila-Correia, 1990 (a)) and stride length decreased when compared to walking in low heeled shoes (Esenyel, et al., 2003; Opila-Correia, 1990 (a)). However, when walking over ground at a fixed-speed, Simonesen and associates (2012) found no significant differences between high heel and BF walking with regard to velocity, stride length, or cadence. Other studies that explored high heel gait while walking on a treadmill reported shorter stride lengths and higher cadence while walking in higher heels than when walking in low heels (Barkema, Derrick, & Martin, 2012) and BF (Simonsen,et al., 2012). Further, Snow and Williams (1994) found, when comparing multiple high heel heights while walking at a fixed speed, significant differences in stride length were not as expected. They found stride lengths were significantly shorter between medium and low heel height, but not between medium and high or low and high heel height (Snow &

Williams, 1994). In addition, a study by Eisenhardt and associates (1996) found stance phase was longer when participants were shod and walking over ground at a fixed rate. Heel height, however, did not appear to influence the amount of time in stance phase (Eisenhardt, et al., 1996). Opila-Correia (1990 (a)) also reported increased stance time while walking at a preferred speed and wearing high heel shoes when compared to low heeled shoes. Simonsen and associates (2012), though, found no significant differences between heel height and duration of stance phase while engaged in over ground walking while barefoot or walking in high heel. Little research exists which explores the distance between an individual's center of gravity and heel at foot contact (COGFC) while wearing high heel shoes. One study did find, however, significant differences between barefoot and high heel, barefoot and toe walking, and high heel and toe walking with toe walking exhibiting the shortest distance from COG and heel at FC, followed by barefoot and high heel (Smallwood, et al., 2016). Based on the aforementioned literature, it has also been found that there were no significant differences in stride length between the conditions because the change in foot position brought about by the footwear negated the inertial properties of the shod condition (Smallwood et al., 2016). Based on the Smallwood (2016) study, it was concluded that the horizontal distance between the individual's COG and foot at initial contact of the gait cycle will be significantly different when comparing barefoot, high heel, HHI, and toe walking conditions as this was observed in previous research to be a more sensitive variable (Smallwood et al., 2016).

Beyond the spatio-temporal variables, peak pressures have also been observed in those who wear high heel. When measuring peak pressure while walking in high heel shoes, it was reported that peak pressure increased with heel height in the central and medial forefoot and loading at the midfoot and under the heel was reduced (Speksnijder, et al., 2005). Speksnijder and associates (2005) reasoned that a smaller sole and heel contact area found with high heel shoes would translate to a reduction of midfoot and heel loading. It was also reported that as heel height increased, peak pressure under the fifth metatarsal head decreased (Eisenhardt, et al., 1996). In addition, the duration of weight bearing on the heel decreased when the heel height exceeded 3.2 cm which Eisenhardt and associates (1996) postulated was due to the smaller heel surface area. However, it was also reported that as heel height increased, the maximum peak pressure became more uniformly distributed among the metatarsals and forefoot as loading increased (Snow, Williams, & Holmes, Jr., 1992; Snow & Williams, 1994). Snow and Williams (1994) theorized the increased forefoot loading resulted from the anterior displacement of center of mass caused by wearing high heel shoes.

A common argument has been wearing high heel shoes results in lordosis (Linder & Saltzman, 1998). However, studies show mixed results with either a decrease (Lee, Jeong, & Freivalds, 2001; Opila-Correia, 1990 (b)), no significant change (Snow & Williams, 1994; Russell, Muhlenakamp, Hoiriis, & DeSimone, 2012), or increase (Opila-Correia, 1990 (b); de Oliveira Pezzan, Joao, Ribeiro, & Manfio, 2011) in lumbar curvature while wearing high heel shoes. Opila-Correia reported individuals with less experience wearing high heel shoes did exhibit increased lordosis when wearing high heel shoes (Opila-Correia, 1990 (b)). A study by deOliveira Pezzan and colleagues (2011) revealed that adolescents who chronically wore high heel shoes did exhibit increased lordosis when compared to adolescents who did not regularly wear high heel shoes while standing in high heel shoes.

In addition to the aforementioned kinematic changes, additional kinematics of the knee and hip have also been reported when wearing high heel shoes. Specifically, Esenyel (2003) reported increased hip flexion during the first half of stance phase of gait when wearing high heel shoes, while Snow and Williams (1994) reported that trunk angle increased with increasing heel height. In contrast, Ebbeling and associates (1994) found no significant difference in hip flexion when comparing barefoot to high heel gait. With regard to knee flexion, various studies have found knee flexion increased during the first half of stance phase as heel height increased (Esenyel, Walsh, Walden, & Gitter, 2003; Opila-Correia, 1990 (a); Opila-Correia, 1990 (b); Ebbeling, Hamill, & Crussemeyer, 1994; Mika, Oleksy, Mika, Marchewka, & Clark, 2012). Opila-Correia (1990 (b)) researched knee flexion in those experienced with wearing high heel shoes to those who rarely wore high heel shoes and found those who were more experienced wearing high heel shoes exhibited increased knee flexion than the non-experienced high heel shoe wearers. However, Simonson and colleagues found no significant differences in knee flexion when comparing age and wearing experience (Simonsen, et al., 2012). Esenyel and colleagues (2003) researched knee flexion in general and postulated the increase in knee flexion was compensation for the increase in heel height. Stefanyshyn and associates (2000) went further and theorized increased knee flexion was a mechanism utilized by the body to counteract the anterior displacement of center of mass. Both Stefanyshyn and colleagues (2000) and Snow and Williams (1994) have reported the individual's center of mass moved anteriorly as heel height increased.

In addition, to spatio-temporal, plantar pressures and kinematics, researchers have also considered kinetic variables such as vertical and anterior-posterior ground reaction forces (GRF). Studies observing GRF found that as heel height increases, vertical and anterior-posterior GRF also increases (Snow & Williams, 1994; Ebbeling, Hamill, & Crussemeyer, 1994). Interestingly, Stefanyshyn et al. (2000), found that maximum vertical GRFs were significantly larger while wearing 3.7 cm and 5.4 cm heeled shoes when compared to flat or 8.5 cm heeled shoes. Snow and Williams (1994) also reported a heel strike transient, or sharp inflection point, which is a short spike of force detected in vertical GRF between initial foot contact and weight acceptance, while wearing high heel shoes. By contrast, a smoother trajectory and smaller inflection point was observed in vertical GRF when wearing low heeled shoes (Snow & Williams 1994). These differences led Snow and Williams (1994) to postulate that a change in the person's perception of stability while wearing high heel shoes could be the reason behind the distinct inflection point detected while wearing the higher heeled shoe.

Neuromechanical Changes in Gait from Wearing High Heeled Shoes

A review of literature found no research addressing the acute effects of wearing high heel shoes once the shoes were removed. A number of studies, however, have explored the long-term effects of wearing high heel shoes. For example, Cronin et al. (2012) and Csapo and associates (2010) reported neuromechanical alterations in gait from wearing high heel shoes. These projects have suggested that when the foot is chronically placed in plantarflexion, such as when a person wears high heel shoes for an extended period of time, the musculotendinous units as well as the gastrocnemii muscle fascicles are susceptible to shortening as well as the Achilles tendon increasing in stiffness and size (Csapo et al. 2010).

Two major categories define muscle fiber arrangement: parallel and pennate. Muscles where mainly parallel fibers traverse the length of the muscle are known for speed production whereas pennate muscles, which are arranged obliquely to the tendon, translate into larger physiologic cross-sectional area, are known to produce greater force (Nordin & Frankel, 2012). Utilizing ultrasound, Cronin and associates (2012) measured medial gastrocnemius muscle fascicle length and activity in women while wearing high heel shoes. It was found that medial gastrocnemius muscle fascicles were shorter while standing in high heel shoes (Cronin, Barrett, & Carty, 2012). In addition, higher muscle activation during stance phase of gait was observed in those who habitually wore high heel shoes for more than two years when compared to those who rarely wore high heel shoes (Cronin, Barrett, & Carty, 2012). To further the work of Cronin and associates (2012), the present study will also consider the pennation angle of the medial gastrocnemius while standing. This is an effort to determine if the act of wearing high heel shoes immediately alters the medial gastrocnemius pennation angle.

The last variables of consideration during high heel gait is muscle activity. Muscle activation while walking barefoot and wearing various heel heights have also produced varied outcomes. While wearing high heel shoes, increased medial gastrocnemius activity, but not soleus or tibialis anterior activity, was detected during stance phase of gait, among those who wore high heel shoes 40 hours a week when compared to those who wore high heel shoes less than 10 hours a week (Cronin, Barrett, & Carty 2012). Mika and colleagues (2011) described a graded response to wearing high heel shoes where medial gastrocnemius and rectus femoris activity increased as the height of the heel increased however, no significant differences were found in biceps femoris activity. Stefanyshyn and colleagues (2000), however, disagreed with Mika and colleagues and reported no significant increase in medial gastrocnemius while walking in high heel shoes. Instead, they found soleus activity increased while walking in high heel shoes (Stefanyshyn et al. 2000). These findings were corroborated by Simonsen and colleagues (2012) who reported that when using electromyography (EMG), when compared to barefoot gait, there was increased peak EMG activity with the soleus, rectus femoris, biceps femoris, and vastus lateralis while walking in high heel shoes amongst those who wore high heel shoes an average of 3.8 times a week compared to those who wore high heel shoes for half a day a week. With regard to the postural muscles erector spinae and rectus abdominis, it was reported that erector spinae activity increased while walking in high heel shoes (Park et al., 2016; Barton, Coyle & Tinley, 2009; Cromwell et al., 2001; Mika et al., 2012 (b); & Lee, Jeong & Freivalds, 2001). In addition, Nascimento and colleagues (2014) reported increased rectus abdominis activity while walking in high heel shoes when compared to barefoot walking.

Use of Orthotic Inserts and High Heel Shoes

Perhaps one of the most famous shoe designers in the 20th century, Salvatore Ferragamo set out to design high heel shoes that were both stylish and comfortable. Realizing that high heel shoes altered the way the weight of the body was distributed, he designed metal arch supports that were incorporated into the high heel shoe (Bergstein, 2012; Salvatore Ferragamo patents and inventions, 2017). Ferragamo is also credited with the reintroduction of the wedge heel; first as an orthotic and then as a fashion statement during the era prior to WWII (Salvatore Ferragamo patents and inventions, 2017). Just prior to WWII, Italy was under a trade ban, which necessitated Ferragamo develop shoes without a steel shank. Ferragamo overcame this obstacle by using cork as a substitute for the arch support and heel of the shoe and designed a wedge heel made of cork (Salvatore Ferragamo patents and inventions, 2017).

Today, in order to attenuate the effects of wearing high heel shoes, a myriad of companies manufacture various products purporting to bring comfort to the wearer. The effects of utilizing products such as metatarsal pads, arch supports, heel cups, and custom total contact inserts as well as perceived comfort were studied by Yung-Hui and Wei-Hsien (2005). This study found that comfort decreased as heel height increased but that the addition of a total contact insert increased comfort perception (Yung-Hui & Wei-Hsien, 2005). It was also reported that the addition of a heel cup or total contact insert to the shoe attenuated the impact force at heel strike (Yung-Hui & Wei-Hsien, 2005).

In an attempt to design a product that would allow women to wear high heel shoes with fewer deleterious effects, Howard Dananberg, DPM, developed a shoe insert that purportedly creates a foot environment that allows the foot to function better while in plantarflexion (Dananberg & Trachtenberg, 2000). Dananberg and Trachtenberg (2000) went on to report that while using his product, named Insolia®, the amount of additional forefoot pressure experienced when wearing high heel shoes was reduced. Another study, exploring energy consumption while wearing high heel shoes and wearing high heel shoes with prefabricated foot orthoses, found that individuals wearing high heel shoes outfitted with Insolia® inserts had improved energy efficiency than when wearing high heel shoes alone (Curran, Holliday, & Watkeys, 2010). Curran and associates (2010) measured oxygen consumption (VO_2), respiration exchange ratio, heart rate, number of steps taken, and physiological cost index while walking under five footwear conditions:

1.5cm heel height, 4.5cm heel height, and 7.0cm heel height without over-the-counter orthotics, 7.0cm heel height with Insolia® orthosis, and 7.0cm heel height with McConnell® orthosis, It was reported that VO_2, number of steps taken, and heart rate was lower when walking while wearing high heel shoes with Insolia® insert than when walking in the high heel shoes alone (Curran, Holliday, & Watkeys, 2010). This is the first study, to the knowledge of this author, which will examine the effects of the Insolia® insert on high heel gait with regard to kinematics, kinetics, and muscle activity.

Toe Walking Gait

Toe walking is usually considered a gait abnormality in which an individual lacks the normal heel contact with the floor at initial contact. In some cases, toe walking is idiopathic and the individual exhibits increased plantar flexion and in other cases, such as that observed with cerebral palsy, toe walking is a result of increased knee flexion (Whittle, 1996). However, individuals do employ toe walking under intermittent, non-pathological circumstances. A paucity of literature exists which examines non-pathological toe walking. A major difference between high heel gait and toe walking is that while a person is walking in a forced plantar flexed position during both conditions, the person engaging in toe walking does not have heel contact with the ground during the gait cycle. A person walking in high heel shoes, however does have the opportunity to utilize a heel strike during the gait cycle even if the heel is positioned above the base of the heel of the shoe. In the present study, toe walking via plantar flexion will be examined to determine if there is a continuum of knee flexion and muscle activity when comparing barefoot, high heel, high heel with an over-the-counter insert, and toe walking gait. No matter the outcome, this research will add to the body of knowledge regarding gait variables.

Purpose of the Study

The purpose of this investigation was: (1) to investigate how walking with an elevated heel, such as walking in high heel shoes, walking in high heel shoes outfitted with Insolia® inserts, or toe walking, affect the kinematic and kinetic variables of gait; (2) to investigate if wearing high heel shoes with Insolia® inserts during gait brings electromyographic variables closer to those variables observed while engaged in barefoot gait; (3) to investigate if wearing high heel shoes, wearing high heel shoes outfitted with Insolia® inserts, or standing in plantar flexion, as observed with toe walking, alters the pennation angle of the gastrocnemius.

Hypotheses

The following hypotheses for this study are listed below:

H_1: When comparing barefoot, high heel, high heel with insert, and toe walking gait, knee and hip flexion during the first half of the stance phase of gait will be the least while walking barefoot and comparable to toe walking followed in order from smallest knee flexion to largest, high heel with insert, and high heel.

H_2: When comparing barefoot, high heel, high heel with insert, and toe walking gait, the horizontal distance between COG to heel at foot contact order from shortest distance to longest will be detected with toe walking, followed by high heel, high heel with insert, and barefoot.

H_3: When comparing barefoot, high heel, high heel with insert, and toe walking gait, the stride length from shortest to longest will be toe walking, followed by high heel, high heel with insert, and barefoot.

H_4: When comparing barefoot, high heel, high heel with insert, and toe walking gait, change in height and normalized change in height will be from smallest to largest barefoot, toe walking, high heel with insert, and high heel.

H_5: When comparing barefoot, high heel, high heel with insert, and toe walking gait, ground reaction forces during stance phase of gait will be smallest during barefoot and toe walking gait, high heel with insert, and high heel gait.

H_6: When comparing barefoot, high heel, high heel with insert, and toe walking gait, gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, rectus abdominis, and erector spinae muscle activation from initial contact through the stance phase of gait, barefoot walking will exhibit the least muscle activation, followed by high heel with insert, high heel, and toe walking gait.

H_7: When comparing barefoot, high heel, high heel with insert, and toe walking, the pennation angle observed in the gastrocnemius will be largest while standing in plantar flexion, as displayed in toe walking. Pennation angles will decrease in the following order: standing in high heel shoes, standing in high heel with insert, and standing barefoot when measured 5 minutes after walking in the above mentioned conditions.

Limitations

The limitations for the present study are:

1. Participants will be relied upon to self-report their health status.
2. Data collection will occur in a gait laboratory which can alter a person’s normal gait pattern.
3. Participants will be between the ages of 19 and 45.
4. Participants will be excluded if they are allergic to adhesives.
5. Participants will be excluded if they are pregnant.
6. Participants will not be excluded due to athletic experience.

Delimitations

The delimitations for the present study are:

1. Participants will wear retroreflective markers.
2. Surface electrodes will be applied to collect muscle activation parameters on the medial gastrocnemius, rectus femoris, tibialis anterior, and erector spinae on the dominant side of the body.
3. Individuals with chronic back pain and/or lower limb injuries will be excluded from the study.
4. The footwear and inserts utilized for this study will be new.
5. Participants will be assessed in the Auburn University Sports Biomechanics Laboratory which is designed specifically for gait analysis.

Chapter II

Literature Review

Shoes and gait are intimately connected. While a person can walk without shoes, there are an infinite number of circumstances in modern society where shoes are required. Women have worn high heeled shoes for generations. Their popularity first came to the fore when Catherine de' Medici wore them for her marriage to Henry II of France in 1533 (Smith, 1999). Since then, high heel shoes withstood ebbs and flows of popularity. Throughout the years, researchers have argued that high heel shoes cause foot deformities in the form of hallux valgus and callouses, as well as other physical maladies such as leg and back pain, and tendonitis (Linder & Saltzman, 1998; Snow, Williams & Holmes, 1992; Snow & Williams, 1994). Despite these warnings women still chose to wear high heel shoes. In 2000, a survey found that women viewed shoes as an important part of their appearance (Joyce, 2000). About 1/5 of the sample stated their reasoning for wearing uncomfortable shoes stemmed from the belief that wearing high heel shoes was an expected norm (Joyce, 2000). This survey also noted that a small percentage of women would wear uncomfortable shoes as long as their appearance was acceptable (Joyce, 2000). While more than half of the women surveyed felt comfort outweighed appearance, only a small percentage of them would wear unattractive shoes to help alleviate a foot problem (Joyce, 2000). These actions leads one to consider wearing high heel shoes as a cultural practice and that some believe the change of appearance, or optical illusion, gained by wearing high heel shoes (i.e., shorter feet, more slender ankles, and longer legs) allowed the person wearing the shoe to improve their power or status (Smith, 1999).

While the present study does not investigate the social and/or psychological issues surrounding the practice of wearing high heel shoes, it is realistic to note that this practice will continue well into the future. As a result, it is reasonable to explore modalities which may ameliorate some of the ill effects of wearing high heel shoes. Therefore, the purpose of this project is to evaluate the influence of wearing high heels on gait parameters. Specifically, this study will observe gait while barefoot, toe walking while barefoot, wearing high heel shoes, and wearing high heel shoes with Insolia® insert. Variables that will be observed include gait kinematics, such as knee and hip flexion from initial contact through mid-stance, vertical and anterior/posterior ground reaction forces (GRF), activation patterns of the gastrocnemius, rectus femoris, tibialis anterior, biceps femoris, rectus abdominis, and erector spinae, horizontal distance from the participant's center of gravity and foot at the point of foot contact (COGFC), and stride length. Gastrocnemius pennation angles will also be measured under the four above mentioned footwear conditions while standing. In light of the goals of this project this chapter will address the pertinent literature and is arranged into the following sections: Normal gait; Influence of heel height on human gait; Muscle pennation angles; Summary.

Normal Gait

Before dissecting the influences of footwear on gait, one must first explore the intricacies of normal gait. Normal gait is an efficient mode of locomotion (Perry & Burnfield, 2010). As Saunders, Inman, and Eberhart (1953) point out, normal gait undergoes sinusoidal curves in the sagittal and transverse planes. In the sagittal plane, the sinusoidal curve occurs twice per gait cycle with peaks at 25 and 75% of the gait cycle, which corresponds to the mid-stance (MSt) phase for each limb. The lowest level of the vertical sinusoidal curve is detected at 50% of the gait cycle, which corresponds to the pre-swing (PS) phase of gait. During the PS phase, both feet are in contact with the ground, in a double support condition. In the transverse plane, the body's center of gravity (COG) moves toward the supporting, or weight bearing limb during gait, to ensure that the COG remains within the base of support. During normal gait, movement in the sagittal and transverse planes can be described as smooth curves, with no sharp or sudden changes in any direction, such as observed with a sine wave pattern (Figure 1) (Saunders, Inman, & Eberhart, 1953). Utilizing a compass model where the lower extremities do not bend at the foot, ankle, or knee, Saunders and colleagues (1953) described the arc formed by the body's COG during normal gait. This type of locomotion, they argued however, would take a considerable amount of energy to accomplish. To counter this increased cost of locomotion and make it more efficient, Saunders and colleagues proposed that pelvic rotation, pelvic tilt, and knee flexion during the stance phase of gait contributed to the flattening of the sinusoidal curves displayed during normal gait (Saunders, Inman, & Eberhart, 1953). Saunders and associates explained that the pelvis rotated approximately 4° from the center to the side of progression while walking (Figure 2). This would translate to a total of an 8° pelvic rotation during one gait cycle. The result of this pelvic rotation is a flattening of the arc formed by the COG during the gait cycle (Saunders, Inman, & Eberhart, 1953). Pelvic tilt was also credited by Saunders and colleagues with lowering the COG of the body during gait. This was accomplished by a 5° pelvic tilt (Figure 3) toward the non-weight bearing limb which contributed to the lateral list exhibited during MSt (Saunders, Inman, & Eberhart, 1953; Perry & Burnfield, 2010). Additionally, knee flexion occurs in the stance limb after initial contact (IC) and continues to flex ~15° until MSt at which time the knee moves into extension (Saunders, Inman, & Eberhart, 1953; Perry & Burnfield, 2010; Kaufman & Sutherland, 2006). These three actions work in concert to flatten the arc travelled by the COG during the gait cycle (Figure 1).

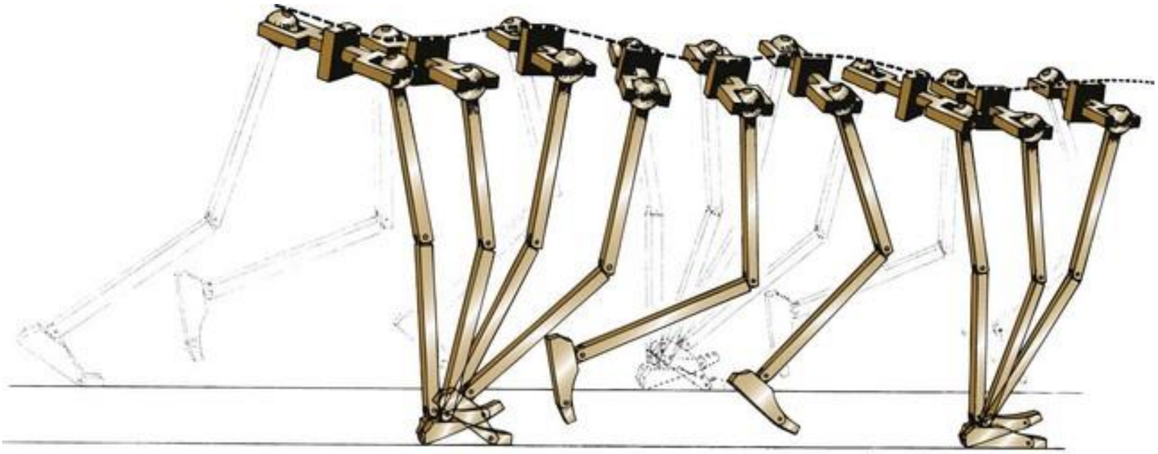

Figure 1. Sinusoidal path of COG during gait cycle (Saunders, Inman, & Eberhart, 1953).

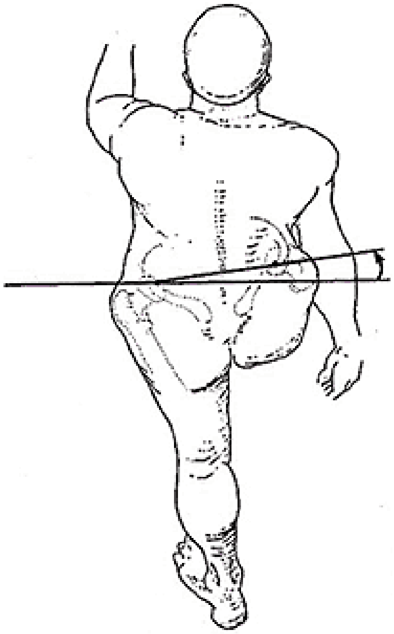

Figure 2. Pelvic rotation from Six Determinants of Gait (Ayyappa, 1997).

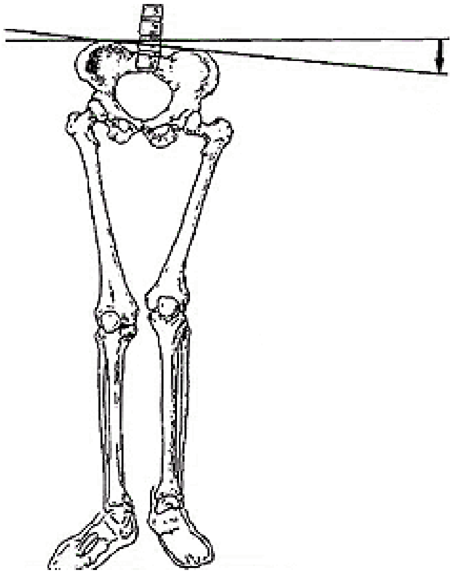

Figure 3. Pelvic Tilt from Six Determinants of Gait (Ayyappa, 1997).

When a person undertakes normal walking, they utilize a gait cycle where the person's limbs work in concert to alternate between forward progression and stability within the base of support. The gait cycle is divided into periods, phases, and tasks. One complete gait cycle is called a stride and is considered the distance between the same two events of the ipsilateral leg (for example, initial contact of the right foot, to initial contact of the right foot). A stride is divided into two periods: stance and swing. Stance is further divided into five phases: initial contact (IC), loading response (LR), mid-stance (MSt), terminal stance (TSt), and pre-swing (PS) and comprises approximately 60% of the total gait cycle. The swing period contains three phases: initial swing (IS), mid-swing (MSw), and terminal swing (TSw) and this period lasts for approximately 40% of the gait cycle. From these divisions, three tasks must be undertaken in order to walk: weight acceptance, single limb support, and swing limb advancement. Weight acceptance occurs during IC and LR. Single limb support ensues during MSt and TSt. Swing limb advancement happens over PS, IS, MSw, and TSw (Figure 4.) (Perry & Burnfield, 2010). Each task will now be addressed in more detail.

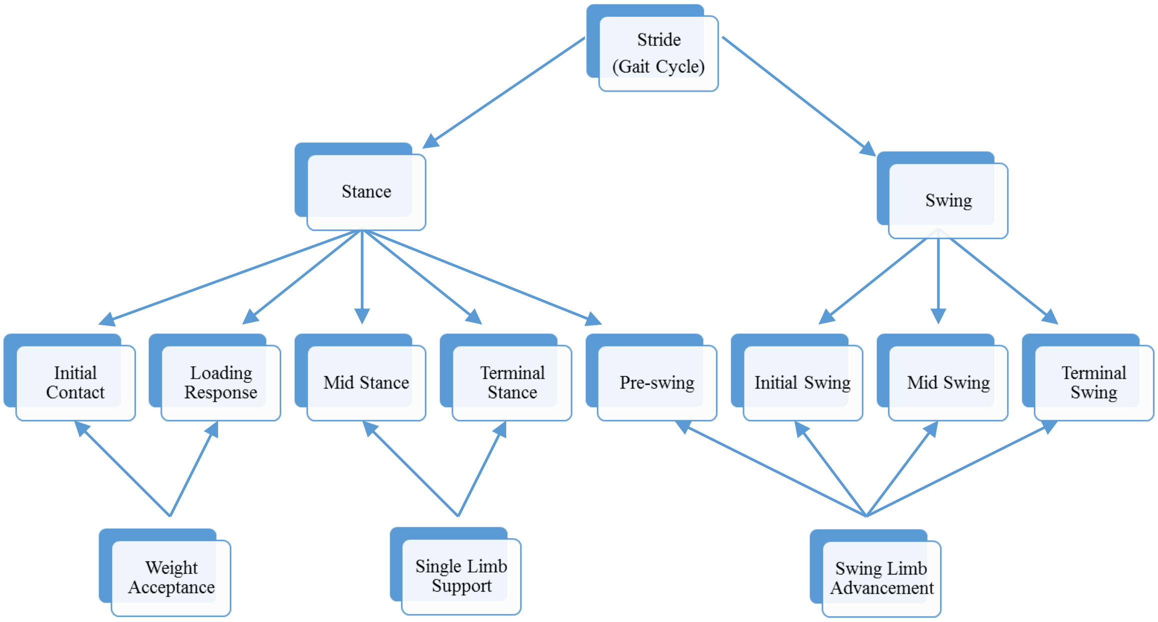

Figure 4. Gait cycle divisions after Perry and Burnfield (2010).

Weight acceptance

During weight acceptance, the body must accept and absorb the shock from the sudden transfer of weight, as well as stabilize that transfer of weight from one limb to the next, all while continuing forward progression (Perry & Burnfield, 2010). Weight acceptance occurs during the stance period of gait over the IC and LR phases. At the onset of IC, the heel of the foot normally makes first contact with the ground. During this time, the hip flexes while the knee extends. The amount of hip flexion reported at IC varies between 20° (Perry & Burnfield, 2010) to ~35° (Gage, Deluca, & Renshaw, 1995). The body must also absorb shock produced by the abrupt transfer of weight from the trailing limb to the advancing limb. During the transfer of weight, the body undergoes a short interval of free fall because the body moves past the base of stability of the trailing foot and moves forward to the leading foot (Perry & Burnfield, 2010). At IC, the ankle moves from a neutral position to a 5° arc of plantar flexion. This occurs just prior to dorsiflexor muscles eccentrically contracting, thus controlling the descent of the foot to the floor and reducing impact (Gage, Deluca, & Renshaw, 1995; Perry & Burnfield, 2010) while mitigating the vertical forces present at ground contact (Ebbeling, Hamill, & Crussemeyer, 1994). The tibialis anterior displays the most activation out of all of the dorsiflexion muscles at this point and therefore will be considered in the present project (Perry & Burnfield, 2010).

A heel strike transient, or inflection point, may occur between IC and weight acceptance and is exhibited as a short spike of force in the vertical ground reaction force (GRF) (Figure 5) that lasts usually 10 – 20 ms (Whittle, 2007; Whittle, 1999). Simon and associates (1981) explain that the inflection point results from an impulse load occurring at high frequency during heel strike. This peak force can reach magnitudes between 0.5 to 1.25 times the individual's body weight (Simon, et al., 1981).

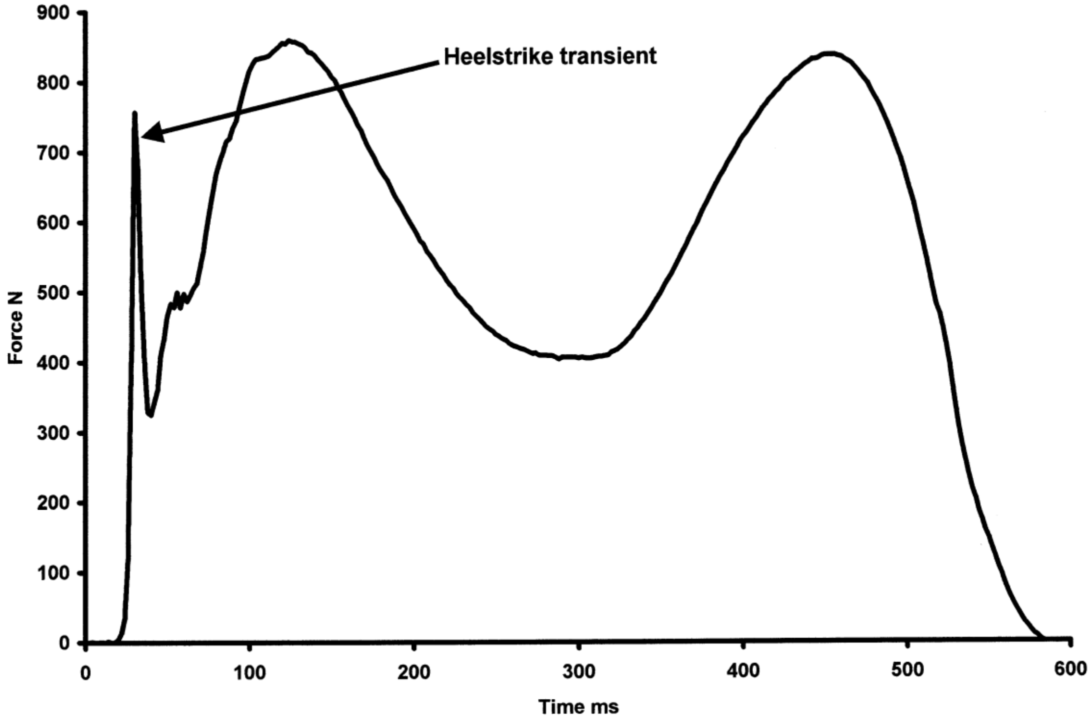

Figure 5. Plot of vertical GRF of individual walking while barefoot (Whittle, 1999).

Jefferson and colleagues (1990) considered two extreme forms of heel strike transient. At one extreme, there is a large, sharp heel strike transient which occurs prior to weight acceptance. At the other extreme, no heel strike transient is perceived prior to weight acceptance. Jefferson and associates (1990) studied two individuals who demonstrated the extremes: one individual exhibited larger heel strike transients while presenting the usual biphasic sagittal plane motions of flexion and extension while walking barefoot; and at the other end of the spectrum, an individual that exhibited no heel strike transient and presented only knee flexion in the sagittal plane while walking barefoot (Figure 6). Differences were reported between the two individuals regarding vertical foot velocity just before foot contact. The individual with the larger heel strike transient displayed a larger vertical velocity (+32%) than the individual with a smaller heel strike transient. It was also found that the individual with the larger heel strike transient exhibited EMG activity in the hamstrings, quadriceps, and gastrocnemius while the individual with the smaller heel strike transient only exhibited EMG activity in the hamstrings and quadriceps. In addition, constant angular acceleration was observed just prior to heel contact in the individual with the smaller heel strike transient while the other individual exhibited increased deceleration at heel strike (Jefferson, Collins, Whittle, Radin, & O'Connor, 1990). These results indicated to Jefferson and colleagues that the person with the larger heel strike transient used the ground to help slow down his limb, while the person with the smaller heel strike transient did not use the ground as a brake and slowed his limb prior to contact with the ground through muscle activity (Jefferson et al., 1990). This study concluded that different strategies were implemented by the individuals at IC. The researchers reported that the individual with the smaller heel strike transient utilized their quadriceps to negatively accelerate the lower limb before contact with the ground and the individual with the larger heel strike transient utilized their gastrocnemius and hamstrings to arrive at the same outcome (Jefferson et al., 1990). These results indicate that quadriceps activation during IC can ameliorate the intensity of the heel strike transient.

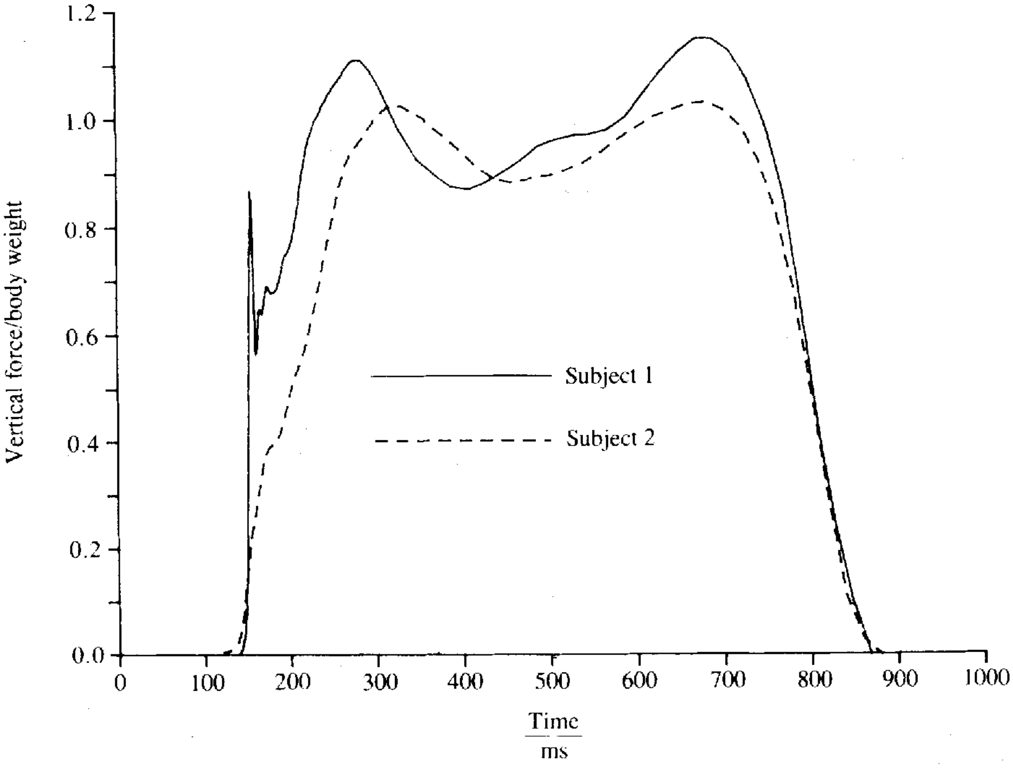

Figure 6. Plot of GRF for two individuals during normal walking. First peak near the beginning of sampling for subject one indicates heel strike transient (Jefferson et al., 1990).

In 2006, Verdini and associates reported findings that indicated there are three types of heel strike transients: a short lived sharp heel strike transient, a longer lived smooth heel strike transient, and no heel strike transient. Verdini and associates observed 25 recreationally active adults while walking barefoot and found 76% of the participants exhibited a sharp heel strike transient, 13.3% exhibited a smooth heel strike transient, and 10.7% exhibited no heel strike transient (Verdini, Marcucci, Benedetti, & Leo, 2006). It was further reported that the timing of tibialis anterior, vastus medialis, and rectus femoris activity influenced the intensity of the heel strike transient. If all of the previously mentioned muscles activated at the appropriate time to slow and stabilize the limb in preparation for ground contact, no heel strike transient would be displayed (Verdini et al., 2006). If one of the observed muscles exhibited an altered activation pattern, a smooth heel strike transient would result. Finally, if two or more of the above mentioned muscles displayed altered muscle activation, then a sharp heel strike transient would be present during the first 10% of the gait cycle (Verdini et al., 2006).

After the heel strike transient, transfer of body weight continues to the leading leg resulting in the first peak displayed in the vertical GRF, which indicates the end of the loading response (Perry & Burnfield, 2010). At the same time, knee flexion also occurs to assist with shock absorption. While the dorsiflexors eccentrically contract to slow the fall of the foot, the tibia and fibula move anteriorly, resulting in increased knee flexion. The quadriceps also activate to reduce the speed of knee flexion, thus allowing increased shock absorption (Perry & Burnfield, 2010; Ebbeling, Hamill, & Crussemeyer, 1994). The advancing limb abductor muscles also contribute to minimizing the impact force because their activation offsets the pelvic drop of the trailing limb, which may result from the sudden transfer of support (Perry & Burnfield, 2010; Whittle, 2007). Furthermore, activity by the lower gluteus maximus, adductor magnus, semimembranosus, semitendinosus, and long head of the biceps femoris contribute to maintaining stability in the hip by preventing further hip flexion (Perry & Burnfield, 2010).

The erector spinae, also known as the anti-gravity muscle, mainly assists in maintaining the body’s upright position. Cromwell and associates (1989, 2001) and Murray and colleagues (1984) reported erector spinae activation immediately after IC. Researchers put forth that erector spinae activation is necessary to negatively accelerate the trunk during gait as well as to sustain the trunk’s upright position (Cromwell, Schultz, Beck, & Warwick, 1989; Cromwell, Aadland-Monahan, Nelson, Stern-Sylvestre, & Seder, 2001). Cromwell and colleagues (1989) went further and reported that erector spinae activity on the contralateral side to heel strike was larger than erector spinae activity on the ipsilateral side of heel strike. It was reasoned that the rise in erector spinae activity on the contralateral side occurred as a mechanism to inhibit disproportionate lateral trunk movement while the trunk's COG moved over the supporting limb (Cromwell, Schultz, Beck, & Warwick, 1989).

Loading response is the last phase of gait contributing to weight acceptance. Objectives of LR mirrors that of the IC. That is, a continuation of shock absorption, weight bearing stability, as well as insuring the body progresses forward. A 5° plantar flexion slows the advancement of the tibia and reduces the speed at which the knee flexes (Perry & Burnfield, 2010). Dorsiflexors also slow the progression of the foot to the floor, resulting in a smoother transition instead of a foot drop or slap. The subtalar joint undergoes eversion, which unlocks the midtarsal joint (Backus, Brown, & Barr, 2012). This action allows the midfoot to become more flexible and adapt to the supporting surface (Perry & Burnfield, 2010; Backus, Brown, & Barr, 2012). As the forefoot comes into contact with the floor, it undergoes pronation and the tibia internally rotates (Whittle, 2007). Shock absorption is accomplished via knee flexion as the quadriceps eccentrically contract to slow the speed of flexion (Whittle, 2007). At this juncture, the body weight moves to the forward limb from the trailing limb. While the tibialis anterior, extensor hallicus longus and extensor digitorum longus eccentrically contract during IC, these same muscles now undergo concentric contractions during the last half of LR as the tibia moves forward (Perry & Burnfield, 2010). Tibialis anterior, extensor hallicus longus, and extensor digitorum longus activity subsides near the completion of the LR (Perry & Burnfield, 2010). During LR, the body's center of mass is at its lowest vertical point during the gait cycle (Perry & Burnfield, 2010; Whittle, 2007; Kaufman & Sutherland, 2006).

Single Limb Support

The main objective of single limb support is to move the body over and beyond the base of the supporting foot while maintaining stability in the trunk and limbs and providing for the forward motion of the advancing limb. Single limb support occurs during MSt and continues through TSt (Figure 7). At MSt, the contralateral foot is lifted from the ground in dorsiflexion which allows for ground clearance and advances the contralateral limb toward the swing period. The knee of the stance limb has reached its maximum flexion and is now beginning to extend. This is accomplished through activation of the quadriceps (Whittle, 2007). Midstance lasts until the individual's center of mass and forefoot align. During Mst, the body alignment is similar to that observed with stable quiet standing (Perry & Burnfield, 2010), with the body appearing to be stacked upon itself. At TSt, the stationary foot initiates a heel rise while the contralateral leg concludes its TSw (Figure 7) . During this time, the body weight advances ahead of the forefoot of the now trailing limb (Perry & Burnfield, 2010). Throughout the stance period, the objective of the muscles are to slow down the effects of gravity and momentum. This is accomplished at limb loading via extensor moments at the hip and knee and through an ankle plantar flexor moment (Perry & Burnfield, 2010).

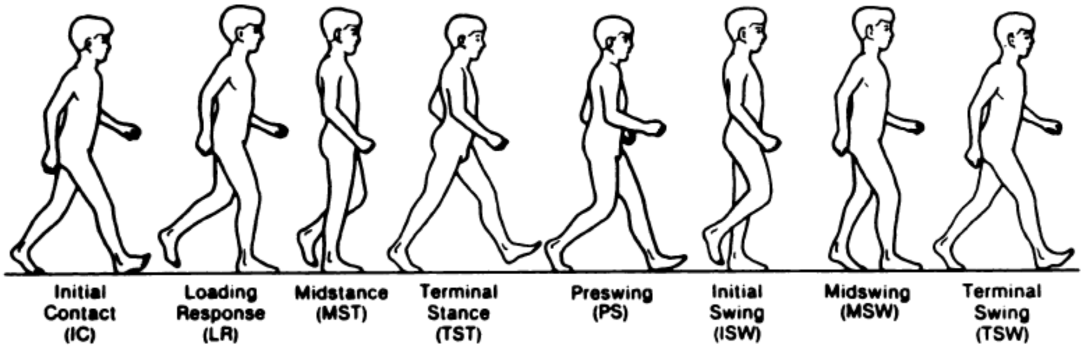

Figure 7. Illustration of individual during gait cycle (Gage, Deluca, & Renshaw, 1995).

Swing Limb Advancement

During swing limb advancement, the body must prepare the leg to swing without colliding with the floor, advance the limb forward and prepare it for the stance period. Swing limb advancement occurs through the PS, IS, MSw, and TSw. Preswing is also considered part of the final phase of the stance period. During PS, the opposite foot is in IC, creating terminal double limb support. The foot in PS increases plantarflexion and knee flexion while decreasing hip extension (Perry & Burnfield, 2010). These movements contribute to the forward progression by directing the force of the foot into the floor and, and in accordance with Newton’s Third Law of Motion, the floor will push back on the foot allowing for forward progression of the limb moving in PS. Preswing concludes when the toe leaves the floor. From PS, the limb proceeds to IS where the foot lifts above the floor and concludes at the point where the advancing foot is opposite the stance foot. At this point, MSw begins and continues until the swinging limb is forward and the tibia is positioned vertical to the ground, and the process begins again.

Though the walking gait pattern is familiar, it is quite complicated with many components. Previous literature has provided the science community divisions which breaks the gait cycle into managable components. These components were utilized in the present study to identify the influence of high heel shoes and toe walking.

Influence of Heel Height on Human Gait

Designer Manolo Blahnik is credited with saying that putting on high heels changes a person (Annal, 2013). Numerous studies have considered the kinematic and kinetic effects of wearing high heel shoes. Opila-Correia (1990 (b)) observed 14 women of various ages and level of experience wearing high heel shoes. Four groups were formed which compared younger (21 – 33 years of age) to older (35 – 54 years of age) women and experienced versus inexperienced wearers of high heel shoes. Kinematic analysis of gait trials in the participant’s own high heeled and low heeled shoes at preferred walking speeds found increased knee flexion during the stance phase of high heel gait among those experienced in wearing high heel shoes when compared to those with less experience under the same condition (Table 1). Younger participants exhibited increased lordosis and a more anterior pelvic tilt while older participants displayed decreased lordosis, or flattening of the lower back, along with a more posterior tilt of the pelvis (Opila-Correia, 1990 (b)). Opila-Correia (1990 (b)) postulated these kinematic alterations of increased knee flexion and lordosis were an effort to dampen the vertical loads exerted on the body during high heel gait. It was suggested that the flattening of the lordotic curve was performed in an attempt to avoid low back pain in the older participants (Opila-Correia, 1990 (b)).

Ebbeling, Hamill, and Crussemeyer (1994) endeavored to determine at what heel height the shoe becomes biomechanically and metabolically detrimental to the person wearing that shoe. To that point, Ebbeling and colleagues studied 15 women, some experienced and some inexperienced wearing high heel shoes, walking at a set speed of 4.2 km/hr. on a walkway and treadmill in a variety of heel heights (1.25cm, 3.81cm. 5.08cm, and 7.62cm). No significant differences were found for any of the variables between the experienced/inexperienced groups. In addition, no significant differences in hip flexion were observed (Ebbeling, Hamill, & Crussemeyer, 1994). Knee flexion did increase as heel height increased, which Ebbeling and colleagues (1994) attributed to the body's attempt to absorb the impact of the heel strike during IC (Table 1). Also measured were GRFs while walking under the aforementioned conditions. Ebbeling et al., (1994) reported an increased first and second maximal force in the vertical direction when comparing the three lowest heel heights to the highest heel height. Shock attenuation is usually accomplished through the eccentric action of the dorsiflexors when the foot moves through plantarflexion during the support phase of gait. According to Ebbeling and colleagues (1994), shock attenuation is reduced while wearing high heel shoes. This reduction occurs due to the foot being placed in an increased plantarflexion position while in the high heel shoes (Ebbeling, Hamill, & Crussemeyer, 1994). This increased plantarflexion inhibits the range of motion that the ankle usually has during weight acceptance and also deters the dorsiflexors from eccentrically engaging during this action, thus limiting the amount of shock absorption that can occur (Ebbeling, Hamill, & Crussemeyer, 1994). The anteroposterior GRF represent the braking and propulsive forces during gait. As heel height increased, Ebbelling et al., (1994) reported an increase in braking force. It was reasoned that this increase in braking force was a result of increased ankle plantarflexion that also altered the ankle angle at touch down to a more plantar-flexed position.

Snow and Williams (1994) observed eleven participants who were experienced in wearing high heel shoes to ascertain if higher heel height alters center of mass (COM) position, posture, sagittal kinematics, and GRF. Shoe heel heights studied included 1.91cm, 3.81cm, and 7.62cm. This study found that while standing, an increase in heel height resulted in increased forefoot loading, increased anterior displacement of the COM, and increased trunk flexion (Snow & Williams, 1994). These alterations were described as compensatory measures to offset the increase in heel height. Snow and Williams (1994) also reported no significant change in the lumbar curve or pelvic tilt while standing in high heel shoes. No significant differences were reported regarding knee and hip flexion at IC or toe off (Table 1) (Snow & Williams, 1994). Maximum knee flexion during swing and knee extension velocity was greater with lower heeled shoes, however Snow and Williams (1994) did not feel these variables would impart great influence on the gait cycle. Comparisons of vertical GRF revealed differences in the initial rise of GRF. Lower heeled shoes displayed a smooth rise to the first maxima while high heel shoes revealed a sharp inflection point, sometimes referred to as a heel strike transient, prior to reaching first maxima (Snow & Williams, 1994). It was postulated that the addition of the inflection point while walking in high heel shoes could be due to perceived stability, or lack thereof, of the shoe at IC (Snow & Williams, 1994). Results from anteroposterior GRF also indicated increased force associated with increased heel height. Snow and Williams (1994) reasoned that because the foot is placed in plantarflexion while wearing high heel shoes, the foot tends to stay supinated, thus limiting pronation. This, according to Snow and Williams (1994), may lead to increased force in vertical and anteroposterior GRF during the first half of stance phase in gait.

Stefanyshyn and colleagues (2000) continued the investigation of the effects of wearing high heel on gait by observing 13 women while walking under four different heel heights (1.4cm, 3.7cm, 5.4cm, and 8.5cm). This study reported that as heel height increased, plantarflexion increased, resulting in the COM moving forward (Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). Stefanyshyn et al., (2000) reasoned that with the forward movement of COM an increase in knee flexion occurred as a mechanism to counteract this change (Table 1). This increase in knee flexion, postulated by Stefanyshyn and colleagues (2000), also resulted in an increase in rectus femoris activation (Table 2), which was reasoned to be a mechanism utilized to stabilize the knee. Soleus activation also increased as heel height increased (Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). Gastrocnemius, tibialis anterior, semitendinosus, biceps femoris and vastus medialis, however, did not display increased activity as heel height increased (Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). Stefanyshyn et al., (2000) also reported increased anteroposterior GRF which was attributed to increases in vertical peak acceleration and deceleration forces. Interestingly, the maximum vertical impact force did not follow a linear pattern. Instead, the maximum vertical impact forces were higher in the 3.7cm and 5.4cm heel, but the 8.5cm heel produced the lowest vertical impact force (Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). This led Stefanyshyn and colleagues (2000) to opine that the decrease in vertical impact force may be a compensatory mechanism to lower the impact from wearing the higher height of high heel shoe.

Mika and colleagues (2011) studied kinematic and EMG variables during high heel gait, between two age groups of individuals, walking at a self-selected pace, under three footwear conditions: barefoot, 4 cm heels (which is considered lower heel), and 10 cm heels (which is considered higher heel). Comparisons between barefoot and higher heel knee flexion at IC and during the first half of stance phase revealed increased knee flexion while walking in the higher heels (Table 1). Mika and colleagues (2011) did not find significant differences between age groups regarding knee flexion (Table 1). However, when comparing younger adult women (20 to 25 years) to older, middle aged women (45 to 55 years), knee and ankle range of motion in the older, middle age group decreased (Mika, Oleksy, Mika, Marchewka, & Clark, 2011). Significant increases in medial gastrocnemius, tibialis anterior, and rectus femoris muscle activity was observed when comparing barefoot to higher heel gait (Table 2). However, no significant age differences were found when analyzing the aforementioned muscles (Mika, Oleksy, Mika, Marchewka, & Clark, 2011). Mika, et al., (2011) agreed with Ebbeling and colleagues (1994) and further posited that the increase in knee flexion occurred so as to ameliorate the effects from increased ground reaction forces at IC displayed with wearing high heel shoes. Mika and colleagues (2011) additionally cautioned that these increases in medial gastrocnemius, tibialis anterior, and rectus femoris activity could further aggravate muscle fatigue and possibly lead to muscle overuse and repetitive strain injuries.

Simonsen and colleagues (2012) explored the effects of wearing high heel shoes on kinematics, kinetics, and muscle activity on 14 women. No significant differences were found between those who frequently wore high heel shoes and those who rarely wore high heel shoes. When walking over ground at a fixed-speed, Simonesen and associates found no significant differences between heel height and velocity, stride length, or cadence. In addition, no significant differences between heel height and duration of stance phase while engaged in over ground walking were found. Simonson and colleagues (2012) reported increased knee flexion but not hip flexion when comparing barefoot to high heel gait.

Esenyel and associates examined the differences between wearing low heeled (1 cm) sports shoes and high heel (6 cm) shoes during over ground walking. Results from Esenyel and associates study found decreased stride length and walking speed when walking at a self-selected pace while wearing high heel shoes (Esenyel, Walsh, Walden, & Gitter, 2003). Increased knee and hip flexion during the first half of stance was also reported by Esenyel and colleagues (2003). It was postulated that the increase in knee and hip flexion occurred as a compensatory mechanism due to the ankle being in a forced plantar flexed position (Esenyel, Walsh, Walden, & Gitter, 2003).

Table 1. Summary of studies investigating knee and hip flexion while wearing high heel shoes.

Author	Knee Flexion	Hip Flexion
Opila-Correia, 1990 (b)	↑ (experienced)	
Ebbeling, Hamill, & Crussemeyer, 1994	↑	↔
Snow & Williams, 1994	↔	↔
Stefanyshyn, et al., 2000	↑	
Mika, et al., 2011	↑	
Simsonen, et al., 2012	↑	↔
Esenyel, et al., 2003	↑	↑

With increased heel height comes increased vertical and anterior-posterior ground reaction force (GRF), but only to a point. Snow and Williams (1994) reported when heel height increased, significant increases in vertical and anterior-posterior GRF occurred.

Ebbeling and colleagues (1994) confirmed this finding by reporting similar results. Different results, however, were reported by Stefanyshyn et al. (2000), when they reported that maximum vertical GRF were significantly larger while wearing 3.7 cm and 5.4 cm heeled shoes when compared to flat or 8.5 cm heeled shoes. It was also noted that while wearing 8.5 cm heeled shoes, maximum vertical impact forces were lower than in the other conditions. A difference in protocols between Stefanyshyn et al., (2000) and Snow and Williams (1994) along with Ebbeling et al. (1994) may shed some light on the dissimilar results. Snow and Williams (1994) examined individuals while wearing heel heights of 1.91 cm, 3.81 cm, and 7.62 cm while Ebbeling and colleagues (1994) examined those wearing heel heights of 1.25 cm, 3.81 cm, 5.08 cm, and 7.62 cm. The aforementioned researchers only explored individuals wearing heel heights up to 7.62 cm, while Stefanyshyn and colleagues (2000) observed individuals wearing a higher heel height of 8.5 cm. This led Stefanyshyn et al. (2000) to question if there is a point concerning heel height where the body enacts a defense mechanism that causes vertical forces to decrease in order to avoid injury. When one wears high heel shoes, a sharp inflection point can be found in the vertical GRF between heel contact and weight acceptance (Snow & Williams, 1994). By contrast, a smoother trajectory and smaller inflection point is detected in vertical GRF when wearing low heeled shoes (Snow & Williams 1994). These differences led Snow and Williams (1994) to postulate that the person’s perception of stability while wearing high heel shoes may explain the distinct inflection point observed while wearing the higher heeled shoe. Previous research which compared toe walking to barefoot and high heel gait revealed no inflection point in vertical GRF when toe walking (Smallwood, et al., 2016).

Increases in vertical and anterior-posterior forces found in those who wear high heel shoes were thought to result from the foot’s inability to fully pronate during the support phase while wearing high heel shoes (Snow and Williams, 1994). Ebbeling, Hamill, and Crussemeyer (1994), hypothesized that since the ankle is in forced plantarflexion when it makes contact with the ground, the ankle cannot fully utilize dorsiflexion. This action hinders eccentric contraction of the dorsiflexors that would normally assist in attenuating ground reaction force during weight acceptance (Ebbeling, Hamill, & Crussemeyer, 1994). Stefanyshyn and colleagues (2000) theorize that the larger braking forces detected when wearing high heel shoes led to larger negative acceleration of the center of mass than when wearing lower heel or flat heel shoes. This in turn, necessitated a larger propulsive force at toe off in order to accelerate the center of mass forward.

Muscle activation while walking barefoot and walking in shoes with various heel heights have yielded mixed results. Cronin, Barrett, and Carty (2012) compared the high heel gait of those who wore high heel shoes for 40 hours a week to those who wore high heel shoes less than 10 hours a week. Using electromyography (EMG) to study muscle activation, Cronin, Barrett and Carty (2012) found those who wore high heel shoes 40 hours a week exhibited increased EMG activation in the medial gastrocnemius, but not tibialis anterior, during the early stance phase of gait while walking barefoot (Table 2). Those who wore high heel shoes less than 10 hours a week did not exhibit this pattern of activation (Cronin, Barrett, & Carty, 2012). Interestingly, when comparing barefoot to high heel gait of those who wore high heel shoes habitually, the results were opposite with an increase in tibialis anterior and soleus activity and no increase in medial gastrocnemius activity (Table 2). However, it should be noted that while medial gastrocnemius activity was not larger, it did occur over a longer period of time, leading Cronin and associates (2012) to opine that this was an adaptation resulting from instability and increased energy costs which occur when wearing high heel shoes(Cronin, Barrett, & Carty, 2012). Mika and colleagues (2011) also compared high heel gait to barefoot gait. They found EMG activity in medial gastrocnemius, rectus femoris, and tibialis anterior increased at initial contact when walking in high heel shoes. Stefanyshyn and colleagues (2000), however, reported medial gastrocnemius activity did not significantly increase when high heel shoes were worn. Instead, they found increased soleus activity during high heel gait (Stefanyshyn et al. 2000). Simonsen and colleagues (2012) reported when comparing high heel gait to barefoot gait, there was increased peak EMG activity in the rectus femoris, vastus lateralis, and biceps femoris, during the first half of stance while walking in high heel shoes. They further reported that medial gastrocnemius and tibialis anterior EMG activity was not significantly higher when comparing high heel gait to barefoot gait (Simonsen, et al., 2012). Simonsen and associates (2012) attributed the increase in EMG activity to increased extensor moment due to increased knee flexion. They also indicated that the controlled speed during walking trials may have contributed to the increased EMG activity. A comparison of barefoot and high heel gait of 20 females by Nascimento and colleagues (2014) found that in addition to increased gastrocnemius, tibialis anterior, rectus femoris and rectus abdominis activity over the entire gait cycle, the timing of onset of muscle activation was altered (Nascimento et al., 2014). While walking in high heel shoes, gastrocnemius and rectus femoris muscle activation occurred later in the gait cycle and lasted longer than when walking barefoot (Nascimento et al., 2014). It was theorized that these differences in gastrocnemius and rectus femoris activity and timing were a result of the high heel shoes placing the foot in plantarflexion (Nascimento et al., 2014).

While walking barefoot, it was noted that erector spinae activation dramatically increased at or immediately after heel contact and was followed by the trunk moving from flexion to extension (Cromwell et al., 2001). Lee and colleagues (2001) reported increased erector spinae activation during gait which they postulated was in response to changes in posture associated with wearing high heel shoes (Table 2). Park and associates (2016) also investigated the effects of high heel shoes on the erector spinae and reported that during gait, as heel height increased, so did erector spinae activity (Park, Kim, Chung, & Hwang, 2016). Mika and colleagues (2012) attributed increased erector spinae activity while walking in high heel shoes on the foot being placed in plantar flexion and an increase in GRF. They further postulated that this increase in activity was an attempt by the body to ameliorate the increased load that propagated up through the skeleton at IC (Mika, Oleksy, Mika, Marchewka, & Clark, 2012). Barton and colleagues (2009), reported that a 20mm heel lift inserted into shoes owned by the participants led to an immediate increased erector spinae activation at heel contact. After two days of wearing the heel lifts, Barton and associates reported a significantly earlier onset of erector spinae activity among participants. It was postulated that this increase was an adaptation to increased GRF at heel strike (Barton, Coyle, & Tinley, 2009). Additionally, Barton and associates opined that the inserts diminished the mechanism which aids shock absorption in the foot, namely limitations in subtalar pronation and internal tibial rotation. (Barton, Coyle, & Tinley, 2009). These studies indicate that the erector spinae plays a vital role in maintaining posture and balance during gait.

Table 2. Summary of literature outcomes of muscle EMG activity while walking in high heel shoes.

Authors	Gastrocnemius	Tibialis Anterior	Rectus Femoris	Biceps Femoris	Rectus Abdominis	Erector Spinae
Simsonen et al., 2012	↔	↔	↑	↑		
Stefanyshyn et al., 2000	↔	↔	↑	↔		
Cronin, Barrett, & Carty, 2012	↑(habitual)* ↔**	↔*(non-habitual) ↑**				
Nascimento et al., 2014	↑	↑	↑		↑	
Park et al., 2016						↑
Barton, Coyle & Tinley, 2009						↑
Cromwell et al., 2001						↑
Mika et al., 2011 (b)						↑
Mika et al., 2011 (a)	↑	↑	↑	↔		
Lee, Jeong & Freivalds 2001		↔				↑

Note: ↔ = no significant difference, ↑ = increase.

* Comparison between habitual versus control (non-habitual) while walking barefoot
** Comparison of high heel participants barefoot versus high heels

To further understand the effects of walking in high heel shoes, a number of studies investigated changes in stride length while wearing high heel shoes. When walking at a self-selected pace during over ground walking, it was found that stride length shortened when wearing high heel shoes up to 5.5 and 6.1 cm (Esenyel, Walsh, Walden, & Gitter, 2003; Opila-Correia, 1990 (a)). However, Snow and Williams (1994) reported that significant differences in stride length were only observed between medium (3.81 cm) and low (1.91 cm) heeled shoes, and between low and high (7.62 cm) or medium and high heel shoes (Snow & Williams, 1994). Changes in center of gravity while wearing high heel shoes have also been studied. Bendix and colleagues (1984) reported that when heel height was increased to 4.5 cm, the line of gravity moved backward. However, Snow and Williams (1994) and Stefanyshyn et al., (2000) reported that when heel height increased, COM moved anteriorly. A dearth of literature exists which explores the distance between an individual’s center of gravity and heel at foot contact (COGFC) while wearing high heel shoes. One study that compared stride length and COGFC found no signficant differences between footwear condition with regard to stride length, but did find, however, significant differences with COGFC between barefoot and high heel, barefoot and toe walking, and high heel and toe walking with toe walking exhibiting the shortest distance from COG and heel at FC, followed by barefoot and high heel (Smallwood, et al., 2016). The above literature caused the author to hypothesize that no significant differences were detected in stride length between the conditions because the change in foot position brought about by the footwear would negate the inertial properties of the shod condition (Smallwood et al., 2016). However, it appeared more plausible that the distance between COGFC will be significantly different when comparing high heel, and high heel with insert, barefoot, and toe walking conditions as this was reported in previous research to be a more sensitive variable (Smallwood et al., 2016).

Over the years, entrepreneurs have developed various products in an effort to ameliorate some of the effects of wearing high heel shoes. These products ranged from the inexpensive over-the-counter shoe inserts to expensive custom-made total contact shoe inserts. These products provided limited relief where heel cups and total contact shoe inserts did attenuate the magnitude of impact transients, but metatarsal pads did not (Yung-Hui & Wei-Hsien, 2005). Yung-Hun and Wei-Hsien (2005) went on to report that an arch support used in their study did show reduction in medial forefoot peak pressures while increasing midfoot pressure. Recently, podiatrist Howard Dananberg developed a shoe insert, branded Insolia®, which he asserts moves pressure from the forefoot while wearing high heel shoes to the heel (Insolia Science for Shoes, n.d.). Claimed alterations in gait include reduced hip and trunk flexion resulting in improved posture while wearing high heel shoes with Insolia® insert (Insolia Science for Shoes, n.d.). A study by Curran and associates (2010) tested this product and found that those who wore the Insolia® and another over-the-counter insert expended less energy (heart rate, oxygen consumption, and respiration exchange ratio) while wearing the aforementioned inserts when compared to high heel gait without inserts.

While numerous studies have explored the pathologies surrounding toe walking, a dearth of literature exists which compares normal gait to toe walking. In an attempt to further the study of gait analysis, Bovi and colleagues (2011) proposed a protocol which observed a variety of locomotive tasks including self-selected, increased, and decreased gait speed, toe walking, heel walking, and ascending and descending steps. Data from the Bovi and colleagues (2011) study found that gait speed was slower and stride length shorter while toe walking when compared to normal walking. Additionally, medial gastrocnemius, soleus, and peroneus longus activity was apparent from IC through MSt and also during part of TSt. While rectus femoris, vastus medialis, and biceps femoris activity increased, tibialis anterior activity was reduced throughout the gait cycle (Bovi, Rabuffetti, Mazzoleni, & Ferrarin, 2011). No inflection point was detected with toe walking when compared to normal walking. This outcome is understandable because there is no heel contact with the ground during the toe walking gait cycle due to the ankle being in plantarflexion. Perry and associates (2003) also compared toe walking to barefoot gait and found no significant differences in knee flexion during stance phase, except during pre-swing, where knee flexion was greater during normal (heel-toe) walking. The results of these studies, therefore, leads the author to believe that the high heel condition will yield larger knee flexion than the high heel with insert condition. In addition, the author also believes that knee flexion during the first half of stance phase during toe walking will be comparable to the knee flexion exhibited during the same phase of barefoot gait. Based on the available literature, this author believes that the high heel condition will result in a larger inflection point between initial contact and weight acceptance than the high heel with insert condition. The author also believes that the high heel with insert condition will exhibit an inflection point between initial contact and weight acceptance that is larger than the barefoot condition. Due to the fact that toe walking does not utilize a heel-toe mechanism, no inflection point should be observed between IC and weight acceptance. Perry and associates (2003) reported timing of the soleus and medial gastrocnemius activity was significantly different when comparing toe walking to heel-toe gait. The onset of soleus and medial gastrocnemius activity occurred much earlier during toe walking versus heel-toe gait. Specifically, earlier muscle activity occurred during MSw while toe walking and just after heel strike during heel-toe gait (Perry, Burnfield, Gronley, & Mulroy, 2003). In addition, the soleus and medial gastrocnemius exhibited much higher levels of mean EMG activity when comparing toe walking to heel-toe gait. Perry and colleagues (2003) theorized this increase in EMG activity was a mechanism to help stabilize the ankle while walking in plantarflexion.

The above literature leads the author to expect to observe increased medial gastrocnemius, rectus femoris, tibialis anterior, biceps femoris, rectus abdominis, and erector spinae activity while wearing high heel shoes when compared to barefoot and high heel with insert conditions. In addition, the author anticipates barefoot gait will exhibit the least medial gastrocnemius, rectus femoris, tibialis anterior, biceps femoris, rectus abdominis, and erector spinae activity when compared to the other conditions. Toe walking muscle activity is predicted to follow the results reported by Perry and associates (2003).Therefore, one would expect to see a continuum of medial gastrocnemius, rectus femoris, tibialis anterior, biceps femoris, rectus abdominis, and erector spinae activity starting with barefoot gait exhibiting the least activity, and increasing from high heel with insert to high heel, and finally toe walking gait.

Muscle Pennation Angles

Muscle fiber arrangement falls into two major categories: parallel and pennate. Parallel muscles contain fibers oriented parallel along the length of the muscle. These fibers are usually long, such as that appearing in the sartorius muscle. Pennate muscles contain one or more tendons that span the length of the whole muscle. Three subcategories comprise the pennate muscle group, depending on how many tendons penetrate the muscle. Unipennate muscles are composed of muscle fibers that attach to one side of a tendon, as found in the biceps femoris. Bipennate muscles exhibit muscle fibers attached obliquely on both sides of a centrally located tendon, as observed in the rectus femoris or flexor hallucis longus. Multipennate muscles are comprised of muscles with several tendons containing muscle fibers running between them, as found in the deltoid (Floyd, 2012).

Muscle architecture plays a role in the amount of force produced through the arrangement of its contractile components. Sarcomeres arranged in series translate into longer myofibrils which can affect a muscle's shortening velocity and the distance the muscle can shorten. In other words, sarcomeres arranged in series can shorten faster and over a greater distance. Sarcomeres arranged obliquely, or like a fan, that occur in pennate muscles, translate into larger physiologic cross-sectional area (PCSA) which allows the muscle to produce greater force (Nordin & Frankel, 2012).

In pennate muscles, fascicles usually attach to the aponeurosis of the muscle. The angle in which the fascicle lays in relation to the muscle's aponeurosis is called the fascicle angle. A muscle's aponeurosis lies at an angle in relation to the tendon, forming the aponeurosis angle. Pennation angle is found by subtracting the aponeurosis angle from the fascicle angle (Nordin & Frankel, 2012) and force is transmitted to the tendon from the fascicles through the aponeurosis (Enoka, 2015). A muscle's pennation angle influences the amount of force it can produce. Larger fascicle angles can translate to larger PCSA, which may result in increased force generation (Oatis, 2009).

In order to view the pennation angle and length of fascicles during an action, some investigators have utilized diagnostic ultrasound as their methodology. This means of inquiry is minimally invasive to the participant. Diagnostic ultrasound has been utilized to view the pennation angle of the medial gastrocnemius during various activities, including walking.

To that end, Lieber and Fridén (2000) reported medial gastrocnemius pennation angle increased ~20 - 45° during low levels of voluntary muscle contraction. In the same vein, Narici and associates (1996) reported that as joint angle increased, medial gastrocnemius pennation increased, whether or not the person's ankle underwent passive rotation or while transitioning from rest to MVIC. With regard to gait, Fukunaga and colleagues (2001) reported that pennation angle remained constant during gait and that during double support and single support phases, they also reported, the tendon of the medial gastrocnemius stretched during the stance phase and recoiled at toe off. This led Fukunaga and associates (2001) to conclude that the tendon behaved like a spring and produced elastic strain energy (Fukunaga, et al., 2001). Csapo and associates (2010) also utilized ultrasound to compare the pennation angle of the medial gastrocnemius of those who chronically wore high heel shoes to those who usually wore flat shoes. No significant differences were reported between the two groups when pennation angle was measured while the participants were prone. This indicated to the authors that the fascicles adjusted to their shortened position resulting from the wearing of high heel shoes and were at their optimal resting force-velocity length (Csapo, Maganaris, Seynnes, & Narici, 2010). Csapo and colleagues went on to further state that increased Achilles' tendon stiffness detected with those who habitually wear high heel shoes most likely assists in maintaining optimal muscle tendon unit tension necessary for force transmission while walking in high heel shoes (Csapo, Maganaris, Seynnes, & Narici, 2010).

Walking is normally described as an activity that uses the body's energy and muscular effort efficiently (Perry & Burnfield, 2010). If someone continually loaded their muscles through chronic wearing of high heel shoes, one would expect to see a change in fascicle pennation angle. However, Csapo and associates (2010) reported no significant differences in medial gastrocnemius pennation angles when comparing chronic high heel shoe wearers to non-high heel shoe wearers. One must note that these pennation angles were measured while the participant was prone and not engaged in gait or other weight bearing activity. This leads the author to question that while studies have shown that as heel height increases so does GRF (Snow & Williams, 1990; Ebbeling, Hamill, & Crussemeyer, 1994; Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000), is this increase enough to alter pennation angle? Or, if there is a change in pennation angle, is it just transitory and the muscle remodels itself over time to reflect alterations in use? This author believes that medial gastrocnemius pennation angles would be different while standing and will exhibit larger medial gastrocnemius pennation angles.

Summary of the Literature

High heel shoes have been a staple in many women's wardrobe for generations and most likely for future generations. Even with the prospect of maladies such as leg and back pain, tendonitis, and foot deformations (Linder & Saltzman, 1998; Snow, Williams & Holmes, 1992; Snow & Williams, 1994), women still chose to wear high heel shoes (Joyce, 2000). To improve the condition of walking in high heel shoes, products, such as over-the-counter shoe inserts or total contact shoe inserts, have been developed with claims of relieving foot pressure and pain. Some of these products, such as heel cups and metatarsal pads, were found to provide limited relief (Yung-Hui & Wei-Hsien, 2005).

This study will investigate a shoe insert which claims to improve the posture of the person wearing the device.

During normal gait, the body undergoes sinusoidal curves which allows the body to efficiently move from one point to the other (Saunders, Inman, & Eberhart, 1953). During high heel gait, the body does not exhibit smooth transitions during the gait cycle. Instead, it has been reported that an increase in vertical ground reaction force and increase in braking occurs during IC, thus altering the smooth transitions normally observed in normal gait (Ebbeling, Hamill, & Crussemeyer, 1994). This study will observe gait under barefoot, high heel, HH $_{\text{Ins}}$, and toe walking conditions to determine if high heel with insert, does ameliorate the high heel condition to the point that it will allow the person to exhibit lower vertical ground reaction forces than those detected when walking in high heel shoes only.

Heel strike transients, with a peak force between 0.5 and 1.25 times an individual's body weight, are events that may occur between IC and weight acceptance (Simon, et al., 1981). Tibialis anterior and quadriceps activation have been implicated in heel strike transient (Simon, et al., 1981; Verdini, Marcucci, Benedetti, & Leo, 2006). It is not unusual to observe increased heel strike transients while walking in high heel shoes when compared to barefoot walking (Saunders, Inman, & Eberhart, 1953; Snow & Williams, 1994).This study will evaluate the following conditions of walking: barefoot, high heel, high heel with insert, and toe walking, to determine if a continuum exists among these four conditions. It is expected that the largest heel strike transient will be exhibited while walking in high heel condition, followed by high heel with insert condition. Walking in barefoot condition is expected to yield the smallest heel strike transient and toe walking should exhibit no heel strike transient because toe walking does not involve a heel-to-toe interaction with the ground.

Walking in shoes of different heel heights has been shown to alter kinematic and kinetic variables of gait (Ebbeling, Hammil, & Crussemeyer, 1994; Opila-Correia, 1990 (a); Mika et al., 2012; Stefanyshyn, et al., 2000; and Saunders, Inman, & Eberhart, 1953). These studies found increased knee flexion during the first half of stance as heel height increased (Ebbeling, Hammil, & Crussemeyer, 1994; Opila-Correia, 1990 (a); Mika et al., 2012;, and Stefanyshyn, et al., 2000),while Opila-Correia (1990 (b)) noted that only those who were experienced in wearing high heel shoes exhibited increased knee flexion. On the other hand, Snow and Williams (1994) found no changes in knee flexion while walking in high heel shoes. It was hypothesized that increased knee flexion while walking in high heel shoes occurs as compensation to the increased vertical ground reaction force and braking force which occurs during IC (Ebbeling, Hamill, & Crussemeyer, 1994; Mika, Oleksy, Mika, Marchewka, & Clark, 2012; Opila-Correia, 1990 (a)); Opila-Correia, 1990(b)). Another change detected with walking in high heel shoes is the location of the COM. Stefanyshyn et al., (2000) found a graded response when, as heel height increased, the COM moved anteriorly. It was theorized that knee flexion increased in order to counteract the change in the COM (Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). This author expects the results of this study will reveal decreased knee flexion at IC when walking under the high heel with insert condition.

Increased muscle activity has been attributed to walking in high heel shoes. Mika et al., (2012 (a)) and Nascimento et al., (2014) reported increased EMG activity in the gastrocnemius, rectus femoris, and tibialis anterior while Simonsen et al., (2012) and

Stefanyshyn et al., (2000) reported no significant differences in gastrocnemius and tibialis anterior activity but an increase in rectus femoris activity. However, Cronin, Barrett, and Carty (2012) reported an increase in gastrocnemius activity when comparing habitual to sporadic wearing of high heel shoes and no significant difference in gastrocnemius activity when comparing high heel to barefoot among those who habitually wear high heel shoes. In addition, when comparing habitual to sporadic wearing of high heel shoes, tibialis anterior activation was not significantly different. Yet, tibialis anterior activation increased when comparing barefoot to high heel among those who habitually wore high heel shoes (Cronin, Barrett, & Carty, 2012). The increase in rectus femoris activity was believed to be a compensation mechanism for the increase in knee flexion brought about by an increase in heel height (Simsonen, et al., 2012; Mika, et al., 2012 (a); Stefanyshyn, et al., 2000; Nascimento, et al., 2014). Nascimento and colleagues (2014) reported both an increase in gastrocnemius and rectus femoris activity as well as a change in the timing of their activation (Nascimento, Saraiva, da Cruz, Jr., Souza, & Callegari, 2014). The results of this study indicated to Nascimento and colleagues that the individual walking in high heel shoes made postural changes in an attempt to counteract any perceived instability brought on by walking in high heel shoes (Nascimento, Saraiva, da Cruz, Jr., Souza, & Callegari, 2014). It is expected that if high heel with insert condition results in muscle activation patterns similar to that observed in barefoot walking, then the knee flexion and heel strike transient of high heel with insert should be similar to that of barefoot walking.

Other kinematic characteristics of gait, such as stride length and changes in center of gravity are sometimes measured when comparing barefoot to high heel gait. Reports stated that stride length shortened (Esenyel, Walsh, Walden, & Gitter, 2003; Opila-Correia, 1990 (b)) and that the center of gravity moved anteriorly (Snow & Williams, 1994; Stefanyshyn et al., 2000) as heel height increased. A more sensitive variable, however, may come from comparing the COGFC of an individual while walking barefoot, high heel, high heel with insert, and toe walking. This method of measurement will measure distance from the center of gravity to the point of foot contact instead of measuring the distance from heel strike of one foot to heel strike of the ipsilateral foot.

In an attempt to ameliorate the effects of wearing high heel shoes, shoe designers and entrepreneurs have developed various products such as over-the-counter shoe inserts and total contact shoe orthotics. A recently developed shoe insert, Insolia®, has been advertised to improve posture while walking in high heel shoes by reducing knee, hip, and trunk flexion (Insolia Science for Shoes, n.d.). A previous study by Curran and colleagues (2010) reported that those who wore high heel shoes with the Insolia® insert did, indeed, expend less energy while walking than those who wore the high heel shoe only (Curran, Holliday, & Watkeys, 2010). This study will expand the body of research started by Curran and associates and further explore the kinematic and kinetic changes which result from walking in the high heel with insert condition.

Non-pathological toe walking does exhibit kinematic and kinetic differences when compared to normal gait. Stride length was reported to be shorter and muscle activation altered while engaged in toe walking when compared to normal gait (Bovi, Rabuffetti, Mazzoleni, & Ferrarin, 2011; Perry, Burnfield, Gronley, & Mulroy, 2003). Because the foot is in constant plantarflexion while toe walking, no inflection point or heel strike transient was detected during toe walking (Perry, Burnfield, Gronley, &

Mulroy, 2003). No differences in knee flexion were found during stance. However decreased knee flexion was detected during PSw while engaged in toe walking when compared to normal gait (Perry, Burnfield, Gronley, & Mulroy, 2003). In the present study, it is anticipated that the magnitude of the transient will progress from smallest to largest across the footwear conditions in the following order: toe walking (no transient), barefoot, high heel with insert and the largest will be noted in the high heel condition

Diagnostic ultrasound is a method whereby investigators can observe the pennation angle of and the length of fascicles. This minimally invasive diagnostic tool has been utilized to view the pennation angle during various activities, such as walking in those who habitually or rarely wore high heel shoes (Fukunaga, et al., 2001, Csapo, Maganaris, Seynnes, & Narici, 2010). While Csapo and colleagues did report increased Achilles' tendon stiffness, they did not find significant differences in pennation angle, leading the authors to speculate that the fascicles adjusted to their shortened position which resulted from wearing high heel shoes (Csapo, Maganaris, Seynnes, & Narici, 2010). This author expects to find no increased pennation angles with high heel, high heel with insert, and toe walking condition when compared to barefoot condition. This is because while the foot is placed in plantarflexion not enough time will have elapsed in order for the muscle fascicles to adjust to their shortened position.

The current study will examine the effects of walking in an elevated heel, such as high heel shoes, walking in high heel shoes with Insolia® inserts, or toe walking on the kinematic and kinetic variables of gait. Another purpose of this project will be to investigate if wearing high heel shoes outfitted with Insolia® inserts brings kinematic and kinetic variables closer to those variables observed during barefoot gait or if there is a continuum of knee flexion, GRF, and muscle activity when comparing high heel, high heel with insert, and toe walking to barefoot gait. Finally, this project will examine if standing in high heel, high heel with insert, or standing in plantarflexion modifies the pennation angle of the gastrocnemius. The present study will add to the body of knowledge concerning high heel shod and toe walking gait.

Chapter III

Methodology:

Participants

Participants between the ages of 19 - 45 years were recruited for this study. Participants indicated they were comfortable wearing high heels in order to take part in the study. Utilizing G*Power 3.1.9.2 for Windows, a power analysis was conducted (Cohen's F = 0.25, alpha = 0.05, and power = 0.80) which determined for a repeated measures analysis of variance (ANOVA), 32 participants were necessary to demonstrate significance. Thirty six participants were recruited to ensure adequate power.

Each participant read and completed a health screen survey. Participants were excluded if they affirmed any of the following: 1) experience back or lower limb injury or surgery during the previous twelve months, 2) experience any injury that would jeopardize successfully performing the required tasks, 3) allergies to adhesive or adhesive type products, 4) cannot fit into the shoes provided for the study (Ladies' US 7-9), 5) any known balance or inner ear maladies, 6) pregnancy, and/or 7) fear of wearing high heeled shoes. Willingness to take part in the study was acknowledged by the participant, voluntarily signing the Institutional Review Board approved Informed Consent document prior to the start of data collection. Participants were then fitted for the correct size high heel shoe to wear for the study.

Setting

All data collection and testing were completed in the Sports Biomechanics Laboratory at Auburn University (Room 020). This is a large laboratory outfitted with the equipment necessary to complete the study.

Instrumentation

Kinematics:

Three dimensional kinematics were collected via a 10 camera Vicon$^{®}$ MX motion analysis system (Vicon$^{®}$, Los Angeles, CA, USA) utilizing a sampling frequency of 100 Hz. Ten cameras were used to create a capture volume large enough to record participants in the three cardinal planes during one complete gait cycle (Figure 8). Retroreflective markers measuring 14 mm in diameter (MKR-6.5, B&L Engineering, Testin, CA, USA) were affixed to the participants using double sided tape (Duck $^{®}$ Indoor/Outdoor Carpet Tape, Shur Tech Brands, Avon, OH, USA). Placement of retroreflective markers are noted in Table 3 and Figure 9. A modified Plug-In-Gait model was utilized for marker placement and data reduction. Data were reduced by Visual 3D (C-Motion Inc., Germantown, MD, USA) to determine segment location and joint angles during one gait cycle. Since this study was designed to compare data within subjects, kinematic data were not normalized. This was done because all trials were performed at the same session, retroreflective markers were not removed between trials, and static trials were performed before each walking trial. If a retroreflective marker happened to fall of the participant, the marker was replaced and a new static trial was performed before resumption of walking trials.

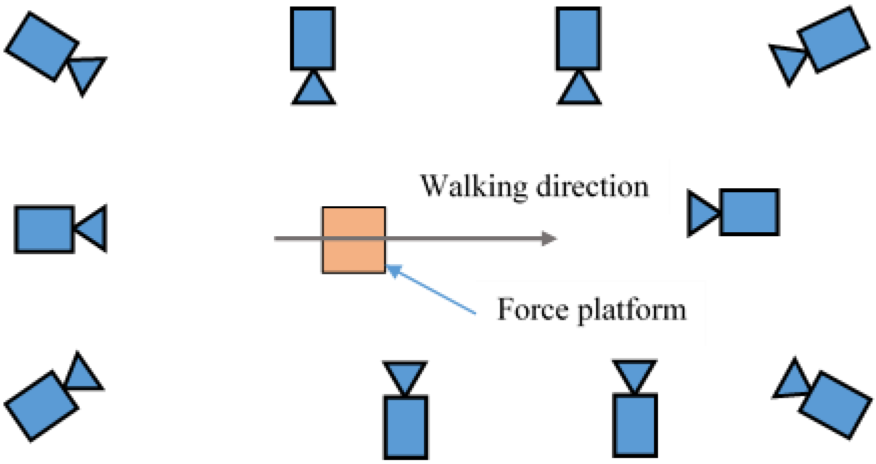

Figure 8. Vicon motion capture camera orientation with AMTI force plate orientation.

Table 3. Names and Positions of Markers Used For 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	Clavicle	Torso
C7	Seventh Cervical Vertebrae	Spine
STRN	Sternum	Torso
T10	Tenth Thoracic Vertebrae	Torso
R/LBAK	Right/Left Back	Torso
R/LSHO	Right/Left Acromioclavicular Joint	Torso
R/LASI	Right/Left Anterior Superior Iliac Spine	Pelvis
R/LPSI	Right/Left Posterior Superior Iliac Spine	Pelvis
SACR	Sacrum	Pelvis
R/LTRO	Right/Left Greater Trochanter	Leg
R/LTHI	Right/Left Lateral Aspect of the Femur	Leg
R/LKNE	Right/Left Lateral Tibiofemoral Joint	Knee
R/LMKNE	Right/Left Medial Tibiofemoral Joint (Calibration)	Knee
R/LTIB	Right/Left Lateral Aspect of the Tibia	Lower Leg
R/LANK	Right/Left Lateral Malleolus	Ankle
R/LMANK	Right/Left Medial Malleolus (Calibration)	Ankle
R/LHEE	Right/Left Heel--Heel Cup for Shoe	Heel/Shoe
R/L1MET	Right/Left Distal End of First Metatarsal	Toe/Foot
R/L5MET	Right/Left Distal End of Fifth Metatarsal	Toe/Foot
R/LHSH	Right/Left Heel of Shoe	Shoe

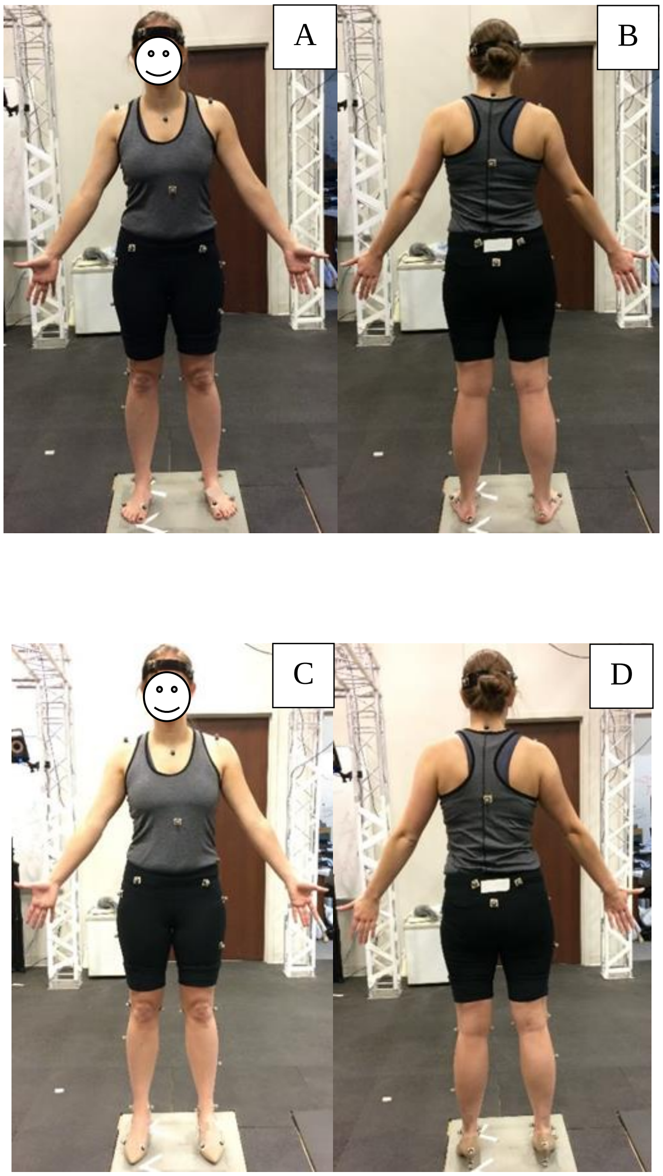

Figure 9. Placement of retroreflective markers while barefoot (A, B) and while wearing shoes (C, D).

The gait cycle for this study started at initial contact of the dominant foot with the floor and continued until the end of the second initial contact with the floor with the ipsilateral limb. This insured that a full gait cycle was observed in the capture volume and allowed for evaluation of events of interest during the gait cycle.

Kinetics

Kinetic data were obtained utilizing an AMTI OR6-1000 force platform (Advanced Mechanical Technology, Inc., Watertown, MA, USA) with MiniAmp MSA-6 amplifiers (Advanced Mechanical Technology, Inc., Watertown, MA, USA) embedded in the walkway (Figure 8.). Force data were transformed from analog to digital data and sampled at 1000 Hz and was used to compare vertical and anterior/posterior ground reaction forces (GRF) during the stance phase. Kinetic data were normalized by dividing each force component by body weight and then multiplying by 100 which resulted in percentage of body weight. Between initial contact and weight acceptance, the magnitude of the heel transient in vertical GRF was measured and compared between the barefoot, high heel, and high heel with insert conditions. In the anterior/posterior GRF, braking and propulsion forces amongst the four conditions were compared.

Surface Electromyography (sEMG)

Muscle activity of the thorax (erector spinae and rectus abdominis) and lower extremity (tibialis anterior, rectus femoris, biceps femoris, and medial gastrocnemius) was monitored using sEMG on the dominant side. These muscles were selected because the erector spinae contributes to negatively accelerating the trunk during gait and sustains the upright position of the trunk (Cromwell, Schultz, Beck, & Warwick, 1989) while the rectus abdominis contributes to stabilizing the trunk and pelvis during midstance (MSt) (Watanabe, 1996). Likewise, the tibialis anterior activation assists to control the descent of the foot to the floor (Gage, Deluca, & Renshaw, 1995; Perry & Burnfield, 2010). Rectus femoris activation during initial contact (IC) is viewed as a mechanism to negatively accelerate the lower limb and reduce the speed of knee flexion (Jefferson, Collins, Whittle, Radin, & O'Connor, 1990), while biceps femoris activity aids in avoidance of knee hyperextension (Perry & Burnfield, 2010). Medial gastrocnemius activation occurs because of the continued plantar flexion associated with walking in high heel shoes.

Six pairs of Ag-AgCl surface electrodes (Red Dot, 3M, St. Paul, MN, USA) were used to monitor muscle activity. Skin preparation for the electrodes included careful shaving, gentle abrading, and cleansing with alcohol. This action removed oil and debris from the skin and facilitated electrical contact with the electrode (DeLuca, 2006). Electrodes were placed no more than 2cm from center to center, along the longitudinal axis of the muscle of interest and midline of the muscle belly (De Luca, 1997). Leads from the electrodes were connected with a Noraxon® Telemyo 2400T-V2 wireless transmitter (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) which relayed data to a Noraxon® Telemyo 2400R-Worldwide Telemetry Receiver (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) and was recorded at a sampling rate of 1500 Hz. (Figure 10). Muscle activity was trimmed to encompass only the dominant side stance phase of the gait cycle. EMG was band-pass filtered (10 – 500 Hz), rectified and low-pass filtered (40 Hz). Peak activation and mean activation of the erector spinae, rectus abdominis, rectus femoris, biceps femoris, tibalis anterior, and medial gastrocnemius activity were compared between walking conditions. Manual muscle testing was performed to ensure proper electrode placement. Further, maximum volitional isometric contractions (MVIC’s) were collected to normalize peak and mean muscle activation during the stance phase of gait. During MVIC’s, the participant was asked to push as hard as they can for five seconds and the mean value of the middle three seconds was the magnitude to which the muscle activity was normalized (Soderberg & Knutson, 2000).

Maximum volitional isometric contractions were performed using the following procedures for the muscles listed (Table 4). Gastrocnemius: the individual was seated in a chair and positioned such that the hips were flexed, the knee of the dominant leg was extended and the foot placed in a plantar flexed position. A strap was placed around the ball of the foot and continued around the back of the chair. The investigator stood behind the participant holding the strap in order to provide resistance while the participant plantar flexed their foot. Tibialis anterior: Sitting with knee flexed, the examiner supported the leg just above the ankle joint while the individual dorsiflexed the ankle joint and invert the foot without extending the great toe. Pressure was placed against the medial dorsal surface of the foot to provide manual resistance to the movement of the individual (Kendall, McCreary, Provance, Rodgers, & Romani, 2005; Halaki & Ginn, 2012). Biceps femoris: Lying prone with the thigh on the table knee flexed between 50° and 70° with the thigh and leg in slight lateral rotation, a strap was wrapped around the ankle and held by the examiner to provide resistance. Rectus femoris: While sitting with knees over the side of the table and holding on to the table, a strap was wrapped around the ankle and held by the examiner behind the table for resistance. The individual fully extended the knee without rotating the thigh. Erector spinae: While prone and with a strap placed across the upper torso to provide resistance, the individual clasped their hands behind their head and performed a trunk extension to their full range of motion while legs were stabilized to the table (Kendall, McCreary, Provance, Rodgers, & Romani, 2005). Rectus abdominis: The individual laid supine with their arms positioned across the chest. A strap was placed across the shoulders of the participant to provide resistance while the individual performed a sit up (Drysdale, Earl, & Hertel, 2004; Halaki & Ginn, 2012;Youdas, 2014).

Table 4. Manual muscle test descriptions

Muscle Investigated	Manual Muscle Test
Gastrocnmeius	Seated, hips flexed, knee extended, foot plantar flexed
Tibialis anterior	Seated, knee flexed, foot dorsiflexed with resistance
Rectus femoris	Seated, knee flexed, perform knee extension with resistance
Biceps femoris	Lying prone, knee flexed, perform knee flexion with resistance
Erector spinae (lumbar region)	Lying prone, trunk extension to full range of motion
Rectus abdominis	Lying supine, arms across chest, perform sit up with resistance

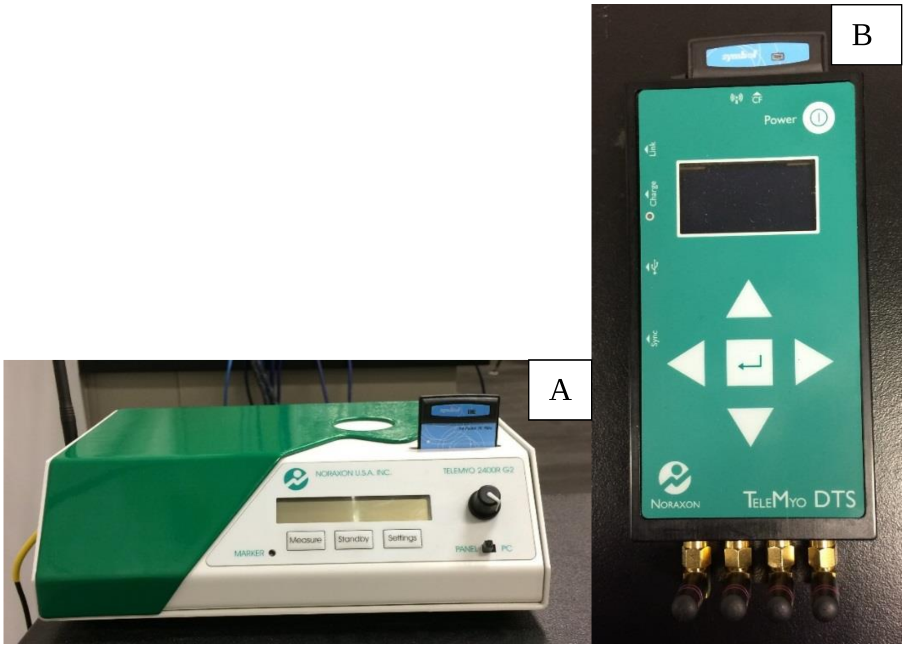

Figure 10. Noraxon® Telemyo 2400R-Worldwide Telemetry Receiver (A) and Noraxon® Telemyo 2400T-V2 wireless transmitter (B).

Unfortunately, an error occurred with data collection of gasctrocnemius MVIC for some participants which rendered the data for this muscle unusable. As a result, the mean and peak sEMG ensemble average value for each muscle over an entire gait cycle during the barefoot condition was used to normalize each sEMG variable. This method of normalization has been done previously and can indicate the amount of activity a muscle is exerting during the gait cycle relative to mean and peak activity recorded during gait (Yang & Winter, 1984; Burden, Trew, & Baltzopoulos, 2003).

Ultrasonography

In order to measure changes in gastrocnemius pennation angle while standing during each footwear condition, GE Logiq S7 Expert ultrasound (GE Healthcare, Milwaukee, WI, USA) was used with a broad spectrum linear probe (GE L-3-12-D, Milwaukee, WI, USA) at a bandwidth of 2-11 MHz (Figure 11). Position of the ultrasound probe was in the central region, along the longitudinal axis and perpendicular to the medial gastrocnemius while the participant was standing (Figure 12) (Bolsterlee, Gandevia, & Herbert, 2016). Logiq S7 software (GE Healthcare, Milwaukee, WI, USA) was utilized to determine pennation angle. The non-dominant leg was selected due to sEMG electrodes creating an obstacle to positioning the ultrasound probe on the dominant leg. In addition, since both legs underwent the same movement, the author believed the images from the non-dominant leg were representative of both legs.

Figure 11. Logiq S7 Expert and L3-12-D probe (geultrasoundpart.com).

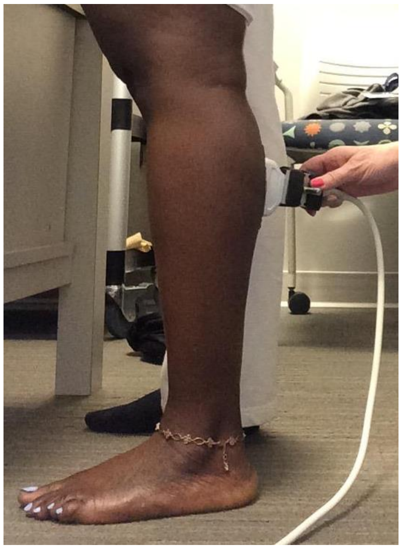

Figure 12. Position of ultrasound probe while standing.

Intraclass Correlation Coefficient

In order to provide reliability of measurements taken during this study, an intraclass correlation coefficient (ICC) protocol was conducted for sEMG and pennation angle measurements to determine the consistency of a single rater. The ICC protocol for sEMG and pennation angle measurement consisted of recruiting at least three individuals to undergo sEMG of the erector spinae, rectus abdominis, biceps femoris, rectus femoris, tibialis anterior, and medial gastrocnemius as well as ultrasound to measure pennation angle of the medial gastrocnemius while standing on two non-consecutive days. Manual muscle tests on the aforementioned muscles were conducted to insure the correct muscle was recorded. Comparisons between data collected on both days resulted in ICC (3,1) average of 0.931 for EMG and 0.8605 for ultrasound measurements. ICC values between 0.5 and .075 are considered moderate, between 0.75 and 0.9 are considered good, and above 0.9 are considered excellent reliability (Koo & Li, 2016).

Footwear

Four footwear conditions were utilized: 1) barefoot, 2) barefoot while toe walking, 3) high heel, and 4) high heel with Insolia insert affixed inside the shoe. High heel shoes with a 3” heel were used for this study. Shoe sizes ranged from Ladies’ US 7 to 9 to allow for proper fitting. To avoid the influence of degradation associated with removing the insert from the insole, two pairs of shoes in each size were utilized: one pair of shoes were worn without an insert and the other pair was outfitted with an insert.

Shoe Insert

The foot orthotic, or insert, is a commercially available pre-fabricated product called Insolia® (Insolia®, Salem, NH, USA) (Figure 13). To prevent slippage of the insert inside the shoe, double sided tape (Duck ® Indoor/Outdoor Carpet Tape, Shur Tech Brands, Avon, OH, USA) was applied to the insert. This allowed for slight adjustments to the placement of the insert inside the shoes, thus ensuring proper positioning of the insert in relation to the heel.

Figure 13. Insolia® shoe orthotics (www.pplbiomechanics.com/products/insolia-high-heel-insert).

Design and Procedures

All testing was completed during one session. Participants were requested to meet at the Sports Biomechanics Lab. Once at the lab, the participants were asked to read and sign the Auburn University Institutional Review Board approved Informed Consent form (Appendix A) and complete the health questionnaire (Appendix B). Signing the informed consent indicated the participant freely volunteered to participate in the study. The health questionnaire was used as a screening tool as it asked if the participant had: 1) experienced back or lower limb injury or surgery during the previous twelve months, 2) experienced any injury that would jeopardize successfully performing the required tasks, 3) allergies to adhesive or adhesive type products, 4) any known balance or inner ear maladies, 5) pregnancy, and/or 6) fear of wearing high heeled shoes, among other questions (Appendix B). When the informed consent signed and inclusion criteria was met, the participant was requested to try on the shoes to insure proper shoe size selection. Next, the participant's mass and height was measured. The participant's foot dominance was determined by asking them to point their toe to a box. The foot which reached toward the box was designated as the dominant foot.

Participants were asked to change into compression clothing and remove any jewelry so as to reduce noise in the motion capture system. The barefoot condition was used as a baseline from which to compare the other footwear conditions. Skin preparation in the area of sEMG electrode placement occurred in the form of shaving, gentle abrading, and cleaning with 70% isopropyl alcohol solution so as to reduce the incidence of electrical impedance. Electrodes were placed 2-4 cm from the innervation zone, between the myotendinous junctions as well as parallel to the muscle fibers of the muscle of interest, i.e., erector spinae, rectus abdominis, rectus femoris, biceps femoris, tibialis anterior and medial gastrocnemius on the dominant side for a total of six muscle assessments (Merletti, 1999; De Luca, 1997). Maximum voluntary isometric contractions (MVIC) were then performed on the muscles of interest utilizing procedures outlined previously.

Footwear condition order (i.e. toe walking, high heel, and high heel with Insolia® insert) was selected via a counter-balance design order. The order of each footwear condition where the first footwear condition were randomly selected (Table 5). Remaining footwear conditions were dependent upon the first footwear condition. For example, if barefoot was selected as the first footwear condition, then the second condition was toe walking, followed by high heel, and finally high heel with insert footwear condition. The toe walking condition involved the participant walking in plantar flexion. In other words, the participant walked on their forefoot only. Once order of footwear was determined, the participant donned the designated footwear and stood or walked for five minutes in order to acclimate themselves to that condition. After that time, ultrasonography was performed on the medial gastrocnemius while standing. A mark was placed on the medial gastrocnemius to ensure that the ultrasound probe would be placed in the same position for each condition. An ultrasound probe was then positioned on the non-dominant leg of the participant in the central region along the longitudinal axis of the medial gastrocnemius while standing. For the toe walking condition, the participant was instructed to walk on their toes. Once the participant was comfortable with the toe walking condition, then ultrasonography was performed on the medial gastrocnemius. The five minute acclimation time was not required for the toe walking trials. Comparisons were made between pennation angles while standing in each footwear condition.

Next, retroreflective markers were affixed to the participant using double sided tape (Duck ® Indoor/Outdoor Carpet Tape, Shur Tech Brands, Avon, OH, USA) on anatomical locations outlined in Table 1. The participant was asked to stand in the capture volume to obtain a static trial. Then the participant was requested to walk through the capture volume at a self-selected pace. A trial was considered successful if the participant’s dominant foot struck the force plate during initial contact. Three successful trials were collected in each condition as this was previously determined to provide reliable EMG data (Arsenault, Winter, Marteniuk, & Hayes, 1986). The participant was then allowed to rest for at least three minutes before embarking on the next footwear condition. This was done to counter any sequential or lingering effects which might have occurred from the footwear condition. After data collection was complete, the participant was thanked for their contribution to the study.

Table 5. Counter-balance Design Footwear Order

First Footwear Condition	Second Footwear Condition	Third Footwear Condition	Fourth Footwear Condition
A (Barefoot)	B (Toe Walking)	C (High Heel)	D (High Heel Insert)
B (Toe Walking)	C (High Heel)	D (High Heel Insert)	A (Barefoot)
C (High Heel)	D (High Heel Insert)	A (Barefoot)	B (Toe Walking)
D (High Heel Insert)	A (Barefoot)	B (Toe Walking)	C (High Heel)

Statistical Analysis:

Based on the intended use of the repeated measures multivariate ANOVA with one group by four conditions, within-groups factor, 32 participants were required to achieve 80% statistical power assuming Cohen's F = 0.25 and $\alpha = 0.05$, and the correlation between repeated measures = 0.5 (Holm's Sequential Bonferroni Procedure) as calculated by GPower 3.1.9.2. Independent variables were footwear conditions (barefoot, high heel, high heel with insert, and toe walking). Dependent variables were knee and hip angle at initial contact and mid-stance of gait, normalized change in height from initial contact to mid-stance, vertical and anterior/posterior ground force reactions during the first half of stance phase, peak and mean muscle activation of medial gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, rectus abdominis, and erector spinae, position of center of gravity at initial contact with regard to heel location, and gastrocnemius pennation angle.

Effect of Footwear Condition on Kinematic Variables during Gait

Powered for main effect of condition the following statistical analyses were performed using SPSS software (version 22, SPSS Inc. Chicago, IL, USA). To investigate the effects of footwear condition on relative knee and relative hip angle at initial contact and maximum flexion during stance phase of gait, a two-way repeated measures ANOVA with the independent variables being footwear condition and position, that is, initial contact and maximum flexion during the stance phase of gait, and the dependent variables being knee and hip angles, was performed. Post hoc repeated measures ANOVAs were performed for each dependent variable of the two-way repeated measures ANOVA that demonstrated significance. Follow-up pairwise comparisons using Bonferroni adjustment to avoid Type I error were then performed to determine statistical significance between independent variables. To observe the effects of change in absolute height from initial contact to midstance of gait a multivariate ANOVA with independent variables being footwear condition and the dependent variable being change in height between initial contact and midstance phases of gait was completed. Follow-up pairwise comparisons using Bonferroni adjustment to avoid Type I error were then performed. To examine the effects of footwear condition on the distance from COG to heel at initial contact a multivariate ANOVA with independent variables being footwear condition and the dependent variable being the horizontal distance between COG and heel at initial contact was performed. Follow-up pairwise comparisons using Bonferroni adjustment to avoid Type I error were then performed. An examination of stride length was performed using a multivariate ANOVA with independent variables being footwear condition and the dependent variable being the distance between the heel strike of the dominant foot and the subsequent heel strike of the same foot.

Effect of Footwear Condition on Kinetic Variables during Gait

To investigate the effects of footwear condition on vertical ground reaction force during the stance phase of the gait cycle, the kinetic data were analyzed utilizing a repeated measures multivariate ANOVA. The dependent variables were peak vertical ground reaction force data and independent variables were heel strike transient, foot loading, and toe off, and footwear condition. A two-way repeated measures ANOVA was used to examine the effects of footwear condition on anterior/posterior ground reaction force. Independent variables were peak breaking and peak propulsion during the stance phase of gait, and footwear conditions. Dependent variables were the anterior/posterior ground reaction forces. Post hoc ANOVAs were performed for each dependent variable of the two-way repeated measures ANOVA that demonstrated significance. Follow-up pairwise comparisons were using Bonferroni adjustment to avoid Type I error then performed to determine statistical significance between independent variables.

Effect of Footwear Condition on Muscle Activity during Gait

To examine the effects of footwear condition on sEMG of peak muscle activation and mean muscle activation during the stance phase during gait, two repeated measures multivariate ANOVA were conducted. Independent variables were erector spinae, rectus abdominis, rectus femoris, biceps femoris, tibalis anterior, and medial gastrocnemius, as well as footwear conditions of high heel, high heel with insert, and toe walking. Dependent variables were peak activation and mean muscle activation during the stance phase of gait. Post hoc ANOVAs were performed for each dependent variable of the repeated measures multivariate ANOVA that demonstrated significance. Follow-up pairwise comparisons using Bonferroni adjustment to avoid Type I error were then performed to determine statistical significance between independent variables.

Effect of Footwear Condition on Medial Gastrocnemius Pennation Angle while Standing

Finally, to investigate the effects of footwear condition on gastrocnemius pennation angle, a multivariate ANOVA was conducted with the independent variable being footwear condition and the dependent variable being the pennation angle of the medial gastrocnemius while standing. Follow-up pairwise comparisons using Bonferroni adjustment to avoid Type I error were then performed to determine statistical significance between independent variables.

Chapter IV

Results

The purpose of this study was to investigate kinematic, kinetic, and neuromuscular variables associated with gait to determine if there is a continuum in various shod conditions. Specifically, this study investigated: (1) how walking with an elevated heel, such as that observed when walking in high heel shoes, high heel shoes with an insert, or toe walking, affects the kinematic and kinetic variables of gait; (2) if walking in high heel shoes with inserts brings muscle activation of the medial gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, erector spinae, and rectus abdominis closer to that found while walking barefoot; and (3) the effects of wearing high heel shoes, wearing high heel shoes with insert, or standing in plantar flexion on the pennation angle of the medial gastrocnemius. The following chapter presents the results of this study in the ensuing order 1) participant demographics, 2) the effect of footwear condition on kinematic variables during gait, 3) the effect of footwear condition on kinetic variables during gait, 4) the effect of footwear condition on muscle activity during gait, and 5) the effect of footwear condition on medial gastrocnemius pennation angle while standing.

Participant Demographics

Thirty-six female participants between the ages of 19 - 45 years, who were comfortable wearing high heels, were recruited for this study (Table 6). All thirty-six participants completed all components of the data collection in one session. However, data from three participants were deemed unusable due to erroneous data, leaving thirty-three participants included in the data analysis for kinematic, kinetic, and pennation angles. In addition, three extra participants were excluded from EMG results due to outlying data Therefore, only a 30 participant EMG data sets were analyzed.

Table 6. Participant Demographics

Participants	**Mean**	**Standard Deviation**
Age (years)	22.3	2.8
Height (m)	1.6	0.048
Weight (kg)	64.9	8.9

Effect of Footwear Condition on Kinematic Variables during Gait

Examination of the effects of footwear condition and time point on knee angles at initial contact and midstance found significant interaction effects between footwear and time, F (3, 30) = 35.676, $p < 0.001$, $\eta_p^2 = 0.781$. A significant main effect for footwear was found, F (3, 30) = 30.318, $p < 0.001$, $\eta_p^2 = 0.752$, and a significant main effect for time was also found, F (1, 32) = 340.064, $p < 0.001$, $\eta_p^2 = 0.914$. To further explore the effects of footwear condition on knee angles at initial contact, a follow-up ANOVA was also conducted, F (3, 30) = 11.152, $p < 0.001$, $\eta_p^2 = 0.527$. The means and standard error for knee angles at initial contact and midstance are presented in Figure 14. Because all data were compared within subjects, standard error was selected to be shown instead of standard deviation for all statistical comparisons. Pairwise comparisons revealed knee flexion at initial contact under the barefoot condition was significantly smaller when compared to knee flexion at initial contact while walking in the high heel and high heel with insert conditions ($p < 0.05$). Knee flexion at initial contact while toe walking was also found to be significantly smaller than knee flexion found at initial contact while walking barefoot and while walking in high heel shoes and high heel with insert ($p < 0.05$). Knee flexion was computed by measuring the angle between the thigh and shank and subtracting the difference from 180°. Therefore, a smaller angle would indicate a more extended knee. No statistical significance was found between high heel and high heel with insert with regard to knee angle at initial contact (Figure 14).

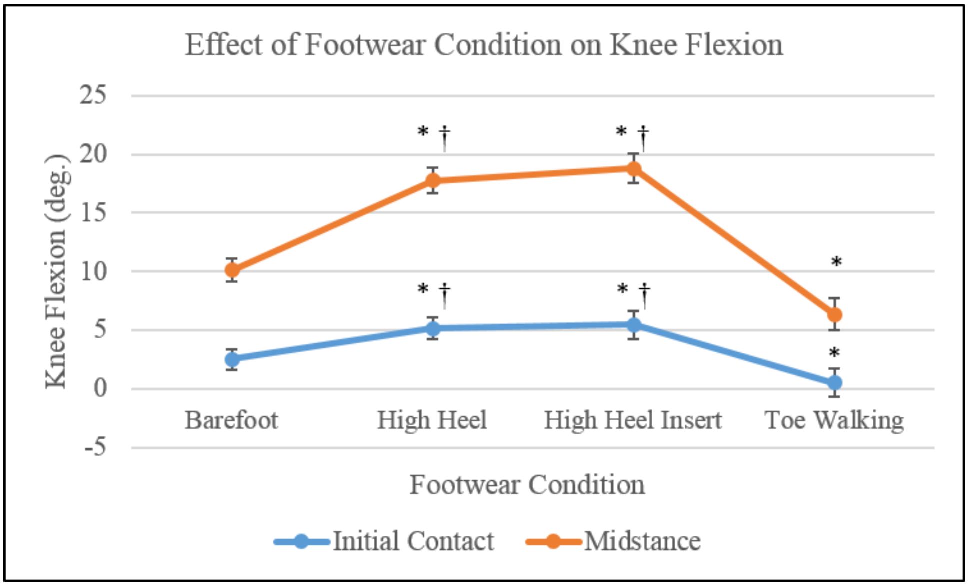

Figure 14. Effect of footwear condition on knee flexion at initial contact and midstance. Notations (*) indicates significant differences when compared to barefoot at $p < 0.05$, and notations (†) indicates significant differences when compared to toe walking at $p < 0.05$.

To further explore the effects of footwear condition on knee angles at midstance, a follow-up ANOVA was also conducted, $F(3, 30) = 41.869$, $p < 0.001$, $\eta_p^2 = 0.807$. Pairwise comparisons revealed the maximum knee angle during midstance under the barefoot condition was significantly smaller when compared to the high heel and high heel with insert conditions ($p < 0.05$). Maximum knee flexion during midstance while toe walking was also found to be significantly smaller than maximum knee flexion while walking barefoot and while walking in high heel or high heel with insert ($p < 0.05$). No statistical significance was found between high heel and high heel with insert with regard to maximum knee flexion during midstance (Figure 14).

Analysis of the effects of footwear condition on hip angle at initial contact and midstance revealed significant interaction effects between footwear and time, F (3, 30) = 6.847, $p < 0.001$, $\eta_p^2 = 0.406$. A significant main effect for footwear was observed, F (3, 30) = 33.165, $p < 0.001$, $\eta_p^2 = 0.768$, and a significant main effect for time was also found, F (1, 32) = 2164.260, $p < 0.001$, $\eta_p^2 = 0.985$. To further explore the effects of footwear condition on hip flexion at initial contact and midstance, a follow-up ANOVA was conducted and indicated a significant footwear effect, F (3, 30) = 23.246, $p < 0.001$, $\eta_p^2 = 0.699$. The means and standard error for hip flexion at initial contact and midstance are presented in Figure 15. Hip flexion was determined by the relative hip angle, that is, the angle formed by the thigh relative to the pelvis. Therefore, a larger number would indicate a more flexed hip. Pairwise comparisons revealed that hip flexion at initial contact under the barefoot and toe walking were significantly smaller than high heel and high heel with insert ($p < 0.05$), but not significantly different from each other. In addition, no significant differences were found between the high heel and high heel with insert. These relationships are presented in Figure 15.

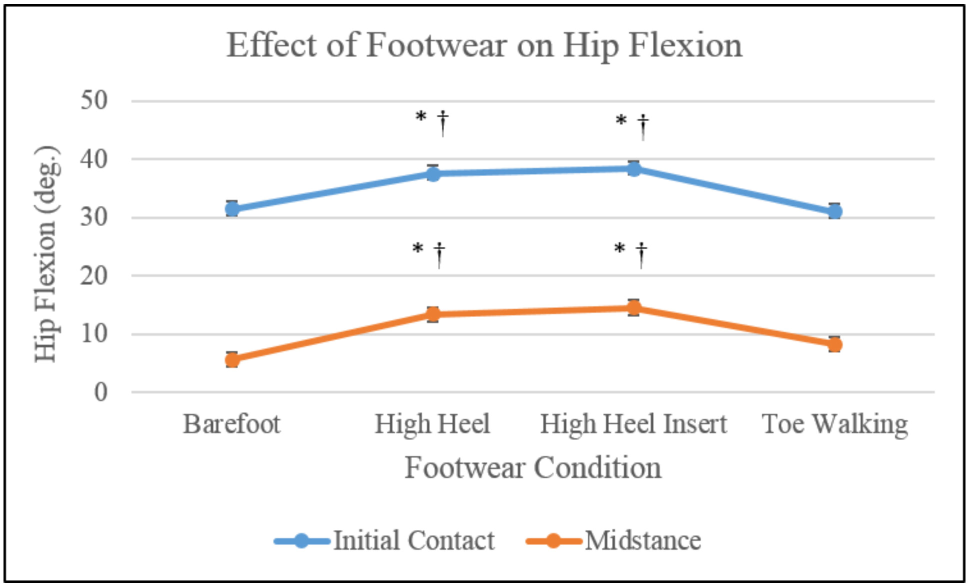

Figure 15. Effect of footwear condition on hip flexion at initial contact and midstance. Notations (*) indicates significant differences when compared to barefoot at $p < 0.05$, and notations (†) indicates significant differences when compared to toe walking at $p < 0.05$.

Further investigation of the effects of footwear condition on hip flexion at midstance was conducted via a follow-up ANOVA. The means and standard error for hip flexion are presented in Figure 15. Results indicated a significant footwear effect, F (3, 30) = 37.029, $p < 0.001$, $\eta_p^2 = 0.787$. Pairwise comparisons revealed hip flexion at midstance under the barefoot condition was significantly smaller when compared to walking in high heel, high heel with insert, and toe walking ($p < 0.05$). Hip flexion at midstance while toe walking was also found to be significantly smaller than while walking in high heel or high heel with insert ($p < 0.05$). No statistical significance was found between high heel and high heel with insert with regard to hip flexion at midstance.

Investigation of the effects of footwear condition on the position of the center of gravity to the point of foot contact (COGFC) found the results of the multivariate ANOVA indicating a significant footwear effect, F (3,30) = 39.270, $p < 0.001$, η_p^2 =

0.797. Means and standard errors for the position of the COGFC with regard to percentage of body height are presented in Figure 16. Pairwise comparisons revealed the COGFC under the barefoot condition was significantly smaller than while walking in the other three conditions ($p < 0.05$). Toe walking also elicited statistically significant greater COGFC when compared to the high heel and high heel with insert conditions ($p < 0.05$). No statistical significance was found between high heel and high heel with insert conditions with regard to COGFC.

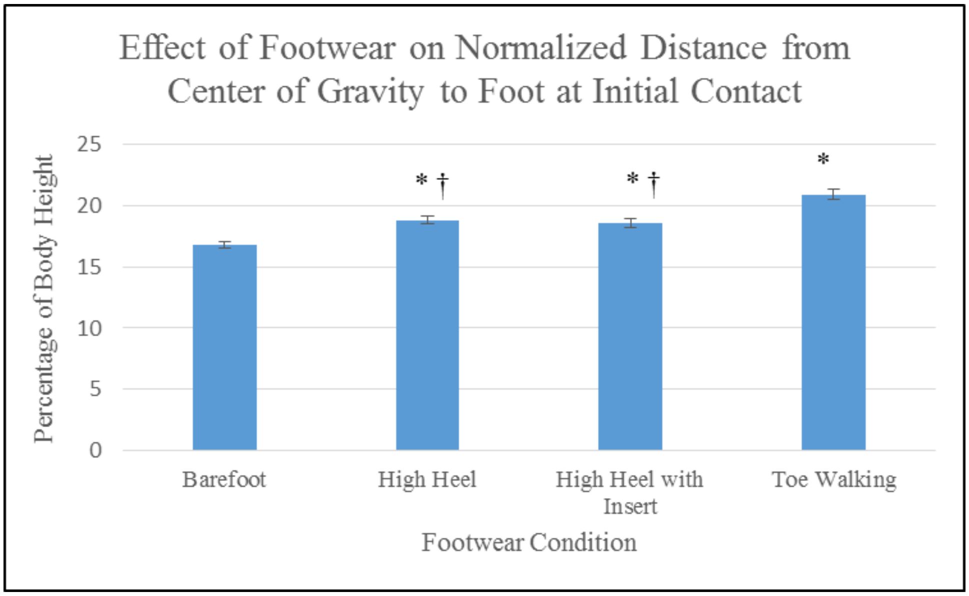

Figure 16. Effect of footwear condition on distance from center of gravity and foot at initial contact. Notations (*) indicates significant differences when compared to barefoot at $p < 0.05$, and notations (†) indicates significant differences when compared to toe walking at $p < 0.05$.

Examination of the effects of footwear condition on stride length found a significant footwear effect, F (3, 30) = 8.418, $p < 0.001$, $\eta_p^2 = 0.457$. The means and standard error for normalized stride length are presented in Figure 17. Pairwise comparisons revealed that stride length under the barefoot condition was significantly longer than while walking in the other three conditions ($p < 0.05$). No significant statistical differences were found when comparing the other footwear conditions to normalized stride length.

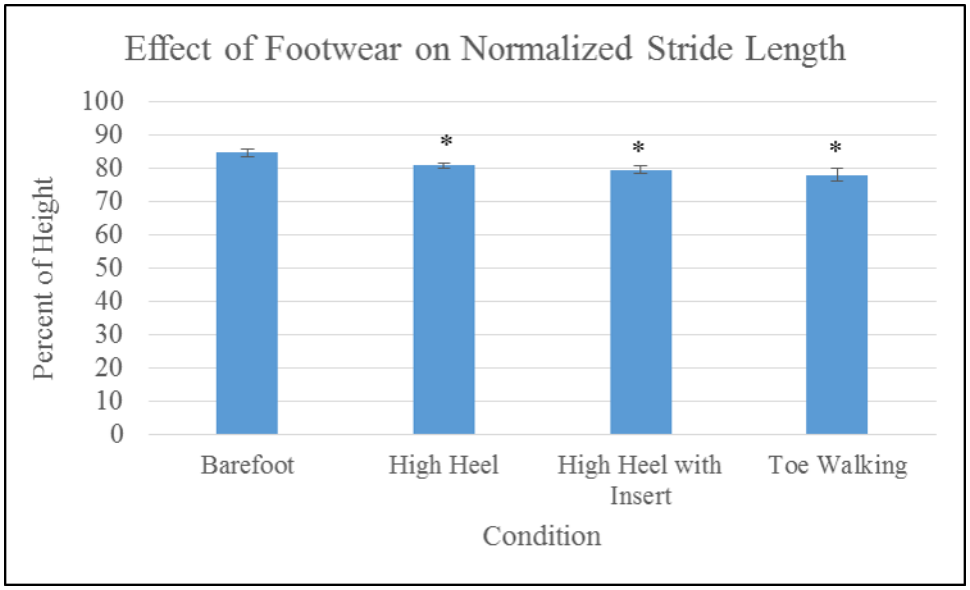

Figure 17. Effect of footwear condition on stride length. Notations (*) indicates significant differences when compared to barefoot ($p < 0.05$).

Analysis of the effects of footwear condition on change in height found a significant footwear effect, F (3, 30) = 17.447, $p < 0.001$, $\eta_p^2 = 0.363$. Pairwise comparisons revealed the change in absolute height under the barefoot condition was significantly smaller than while walking in the other three conditions ($p < 0.05$). No significant statistical differences were found when comparing the other footwear conditions to change in height. The means and standard error for change in absolute height are presented in Figure 18.

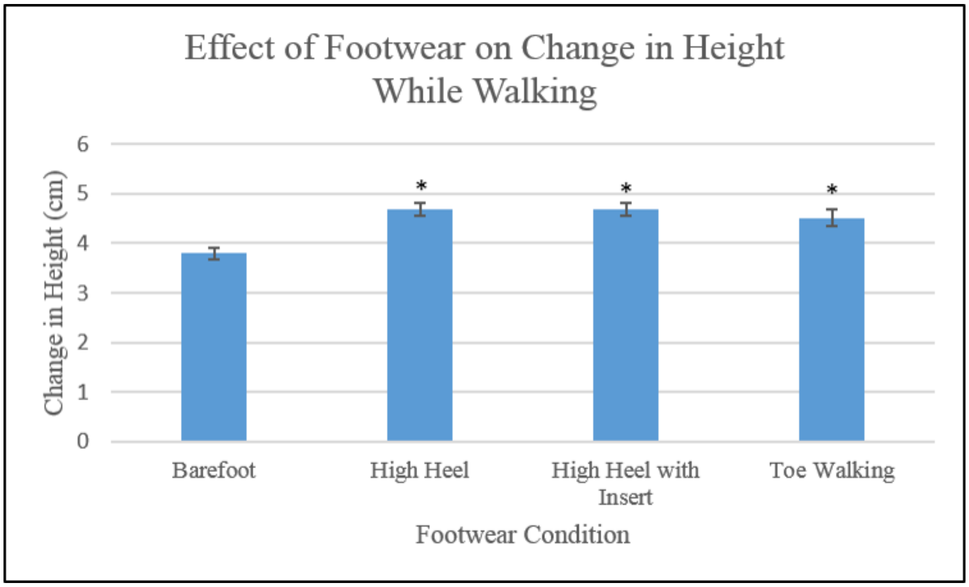

Figure 18. Effect of footwear condition on change in height while walking. Notations (*) indicate significant difference when compared to barefoot ($p < 0.05$).

In addition to exploring change in height while walking, the data were normalized in order to examine the change in percentage of body height. A significant footwear effect was detected, F (3, 30) = 17.500, $p < 0.001$, $\eta_p^2 = 0.636$. Pairwise comparisons revealed the change in normalized absolute height under the barefoot condition was significantly smaller than while walking in the other three conditions ($p < 0.05$). No significant statistical differences were found when comparing the other footwear conditions to normalized change in height. The means and standard error for normalized change in height are presented in Figure 19.

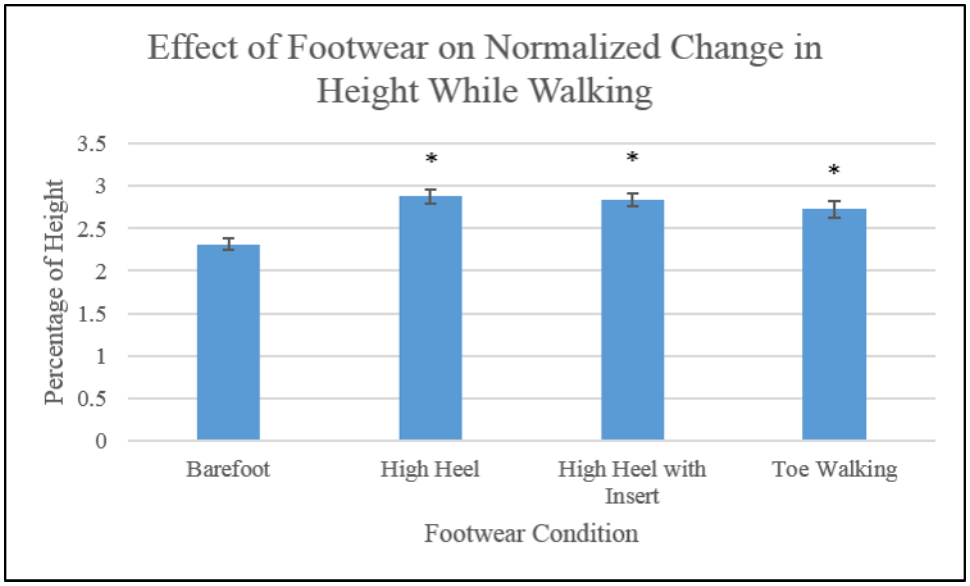

Figure 19. Effect of footwear condition on normalized change in height while walking. Notations (*) indicate significant difference when compared to barefoot ($p < 0.05$).

Effect of Footwear Condition on Kinetic Variables during Gait

Investigation of the effects of footwear condition and anterior/posterior ground reaction force (GRF), specifically investigating peak braking and peak propulsion during the stance phase of gait found significant interaction effects between footwear and time, F (3, 30) = 5.261, p =0.005, $\eta_p^2 = 0.345$. Simple main effects analysis showed a significant difference in footwear condition, F (3, 30) = 4.835, p = 0.007, $\eta^2 = 0.326$, and in anterior/posterior ground reaction forces, F (1, 32) = 90.069, $p < 0.001$, $\eta_p^2 = 0.738$. Further exploration of the effects of footwear on anterior/posterior GRF found significant footwear effect, F (3, 30) = 5.950, p = 0.003, $\eta_p^2 = 0.373$. Means and standard error for the magnitude of peak braking are presented in Figure 20. Pairwise comparisons revealed the magnitude of braking under the barefoot condition was significantly smaller than while walking in the high heel with insert and toe walking conditions ($p < 0.05$). No significant statistical differences were found between barefoot and high heel gait. In addition, no significant statistical differences were found when comparing the other footwear conditions, specifically between high heel and high heel with insert, high heel and toe walking, and high heel with insert and toe walking gait with regard to the magnitude of peak braking. Further investigation of the effects of footwear condition on anterior/posterior GRF, specifically peak propulsion, resulted with a follow up ANOVA revealing no significant footwear effect, F (3,30) = 2.553, $p = 0.074$, $\eta_p^2 = 0.203$.

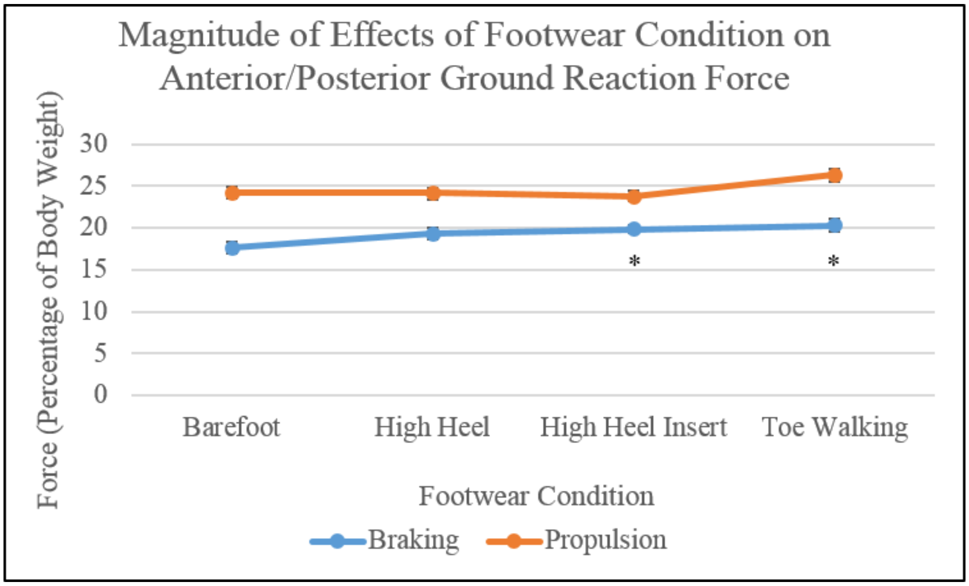

Figure 20. Effect of footwear condition on peak braking and peak propulsion in the anterior/posterior direction while walking. Notations (*) indicate significant difference when compared to barefoot ($p < 0.05$).

Analysis of the effects of footwear condition on the vertical GRF heel strike transient, foot loading, and peak propulsion while walking found significant interaction effects between footwear and time, F (6, 27) = 24.147, $p < 0.001$, $\eta_p^2 = 0.843$. Simple main effects analyses showed significant difference in footwear condition, F (3,30) = 17.046, $p < 0.001$, $\eta^2 = 0.630$ as well as significant differences in vertical GRF, F (2, 31) = 496.115, $p < 0.001$, $\eta_p^2 = 0.970$. Further exploration of the effects of footwear condition on vertical GRF, specifically investigating heel strike transient during the stance phase of gait, revealed significant footwear effect, F (3, 30) = 39.786, $p < 0.001$, $\eta_p^2 = 0.799$. The means and standard error for heel strike transient are presented in Figure 21. Pairwise comparisons revealed heel strike transient under the barefoot condition was significantly smaller than while walking in the high heel or high heel with insert condition ($p < 0.05$). No significant statistical differences were found between high heel and high heel with insert gait. Toe walking condition was not compared as the heel does not make contact with the ground when walking in this condition.

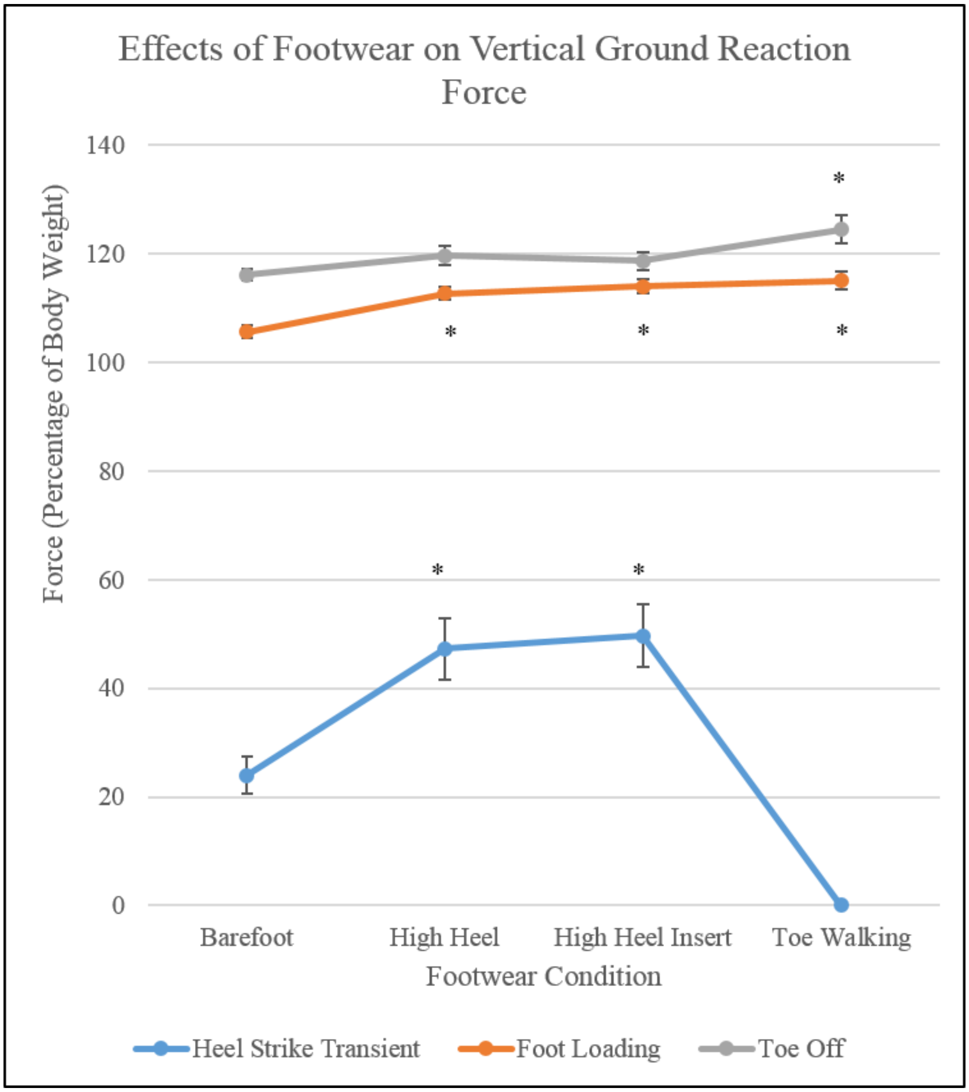

Figure 21. Effects of footwear condition on vertical ground reaction force. Notations (*) indicate significant difference when compared to barefoot ($p < 0.05$). Toe walking condition was not compared at heel strike transient as the heel does not make contact with the ground when walking in this condition.

Further investigation of the effects of footwear condition on vertical GRF, specifically investigating peak vertical force at foot loading during the stance phase of gait indicated a significant footwear effect, F (3, 30) = 27.600, $p < 0.001$, $\eta_p^2 = 0.734$. Means and standard error for peak vertical ground reaction force at foot loading are presented in Figure 21. Pairwise comparisons revealed peak vertical force at foot loading under the barefoot condition was significantly smaller while than while walking in the high heel, high heel with insert, or toe walking condition ($p < 0.05$). No significant statistical differences were found between high heel, high heel with insert, or toe walking gait.

Analysis of the effects of footwear condition on vertical GRF, specifically investigating the propulsive force at toe off during the stance phase of gait, revealed a significant footwear effect, F (3, 30) = 4.886, p = 0.007, $\eta_p^2 = 0.328$. The means and standard error for peak vertical ground reaction propulsive force are presented in Figure 21. Pairwise comparisons indicated peak propulsive vertical force under the barefoot condition was significantly smaller than toe walking ($p < 0.05$) condition only. No other significant statistical differences were found.

Effect of Footwear Condition on Muscle Activity During Gait

Exploration of the effects of footwear condition on mean muscle activation while walking revealed significant interaction effects between footwear condition and muscle of interest with an increase in mean muscle activation while in the toe walking condition, F (10, 20) = 5.637, $p = 0.001$, $\eta_p^2 = 0.738$. These results indicate that the position of the foot while walking affects the normalized mean muscle activation. Simple main effects analyses showed a significant difference in footwear condition, F (2, 28) = 15.455, $p < 0.001$, $\eta_p^2 = 0.525$ as well as mean muscle activation, F (5,25) = 17.992, $p < 0.001$, $\eta_p^2 = 0.783$.

Follow up analysis on the effects of footwear condition on normalized mean gastrocnemius activation indicated significant footwear effect, F (2, 28) = 56.313, $p <$

0.001, $\eta_p^2 = 0.801$. The means and standard error for the normalized mean gastrocnemius activation are presented in Figure 22. Pairwise comparisons indicated normalized mean gastrocnemius activation under the toe walking condition was significantly larger than while walking in high heel and high heel with insert condition ($p < 0.05$). No significant statistical differences were found between high heel and high heel with insert conditions.

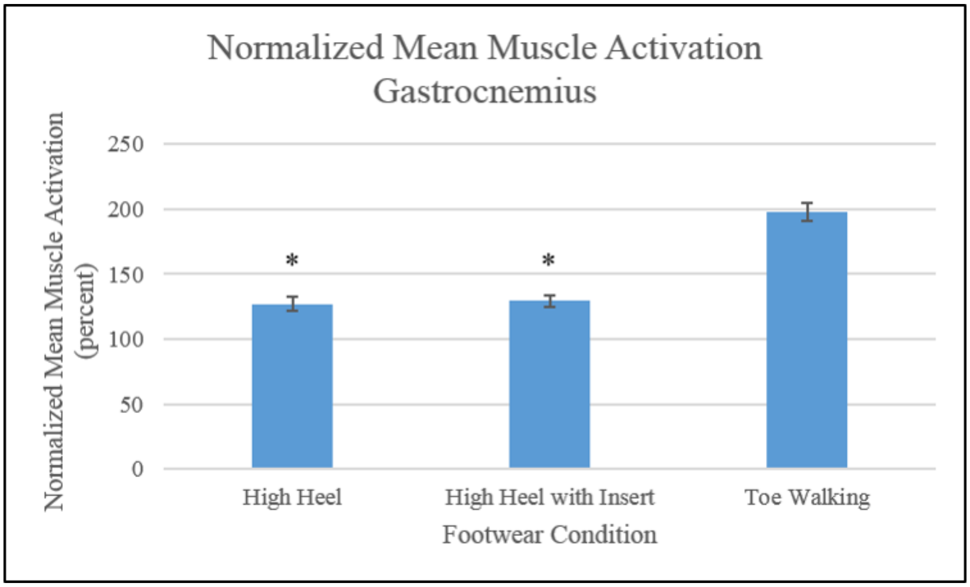

Figure 22. Effect of footwear condition on normalized mean gastrocnemius activation during stance phase of gait. Notations (*) indicate significant difference between toe walking at ($p < 0.05$).

Further assessment of the effects of footwear condition on normalized mean biceps femoris activation indicated a significant footwear effect, F (2, 28) = 4.185, p = 0.026, $\eta_p^2 = 0.230$. Means and standard error are presented in Figure 23. Pairwise comparisons revealed normalized mean biceps femoris activation under the high heel condition was significantly smaller than while walking in the toe walking condition ($p < 0.05$). No significant statistical differences were found between high heel and high heel with insert, nor between toe walking and high heel with insert.

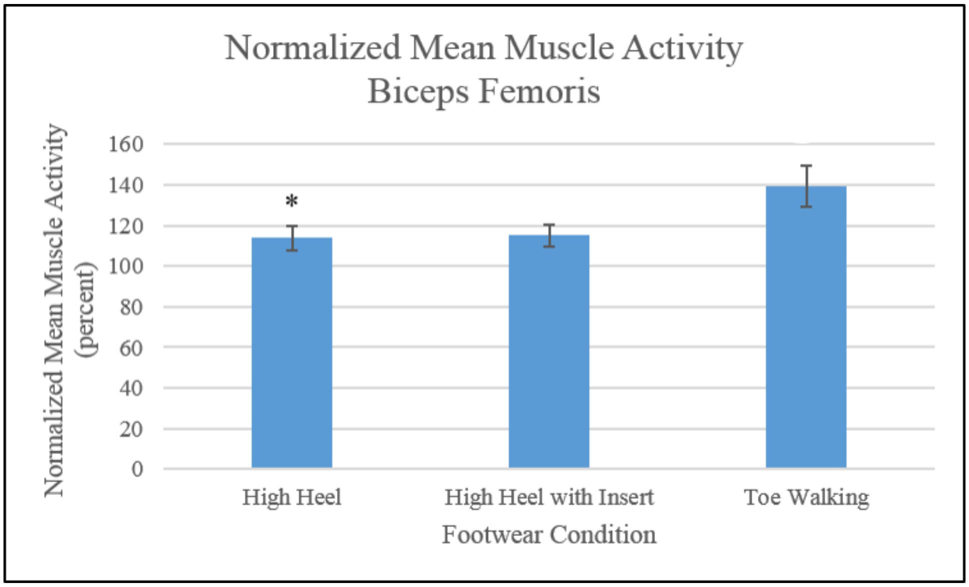

Figure 23. Effect of footwear condition on normalized mean biceps femoris activation during stance phase of gait. Notations (*) indicate significant difference between toe walking at ($p < 0.05$).

Follow-up ANOVAs found no significant condition effect for the following mean muscle activation variables (Table 7); indicating that footwear condition did not significantly affect mean muscle activation of the tibialis anterior, rectus femoris, rectus abdominis, or erector spinae.

Table 7. Non-significant condition effects for sEMG average muscle activation.

	F	df	N	p	η_p^2
Tibialis Anterior	2.456	2	30	0.104	0.149
Rectus Femoris	1.613	2	30	0.217	0.103
Rectus Abdominis	1.940	2	30	0.163	0.122
Erector Spinae	0.044	2	30	0.957	0.003

Analysis of the effects of footwear condition on peak muscle activation while walking revealed significant interaction effects between footwear condition and muscle activation with an increase in muscle activation while in the toe walking condition, *F* (10, 20) = 5.637, $p = 0.001$, $\eta_p^2 = 0.738$. These results indicate that the position of the foot while walking affects the normalized peak muscle activation. Simple main effects analyses showed a significant difference in footwear condition, *F* (2, 28) = 15.455, $p < 0.001$, $\eta_p^2 = 0.525$ as well as normalized peak muscle activation, *F* (5,25) = 17.992, $p < 0.001$, $\eta_p^2 = 0.783$. Exploration of the effects of footwear condition on normalized peak gastrocnemius activation indicated a significant footwear effect, *F* (2, 28) = 56.313, $p < 0.001$, $\eta_p^2 = 0.801$. The means and standard error for the normalized peak gastrocnemius activation are presented in Figure 24. Pairwise comparisons indicated normalized peak gastrocnemius activation under the toe walking condition was significantly larger than the high heel and high heel with insert conditions ($p < 0.05$). No significant statistical differences were found between high heel and high heel with insert conditions.

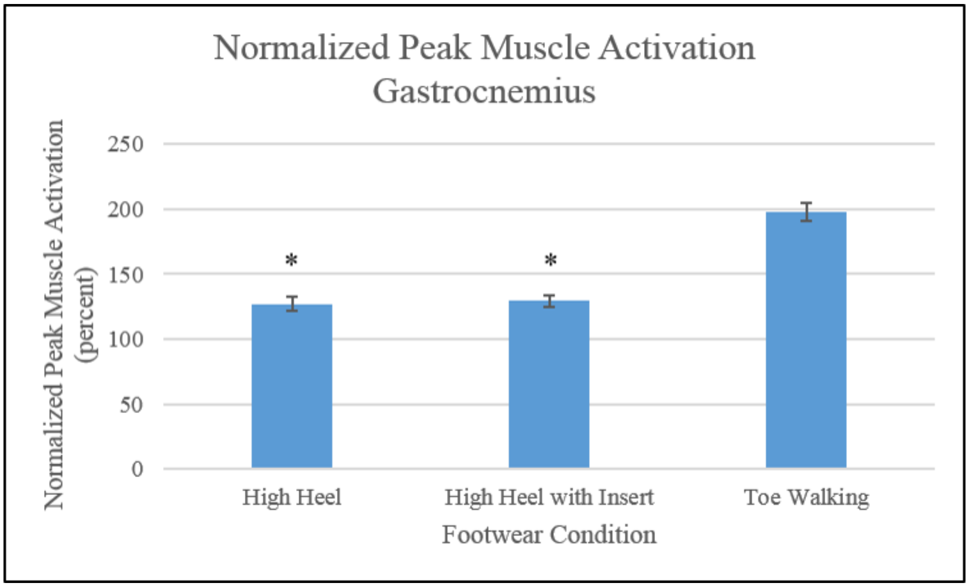

Figure 24. Effect of footwear condition on normalized peak gastrocnemius activation during stance phase of gait. Notations (*) indicate significant difference between toe walking at ($p < 0.05$).

Further examination of the effects of footwear condition on normalized peak biceps femoris activation revealed significant footwear effect, F (2, 28) = 4.185, p = 0.026, $\eta_p^2 = 0.230$. Means and standard error are presented in Figure 25. Pairwise comparisons indicated normalized peak biceps femoris activation under the toe walking condition was significantly larger than high heel ($p < 0.05$). No significant statistical differences were found between high heel and high heel with insert, nor toe walking and high heel with insert conditions.

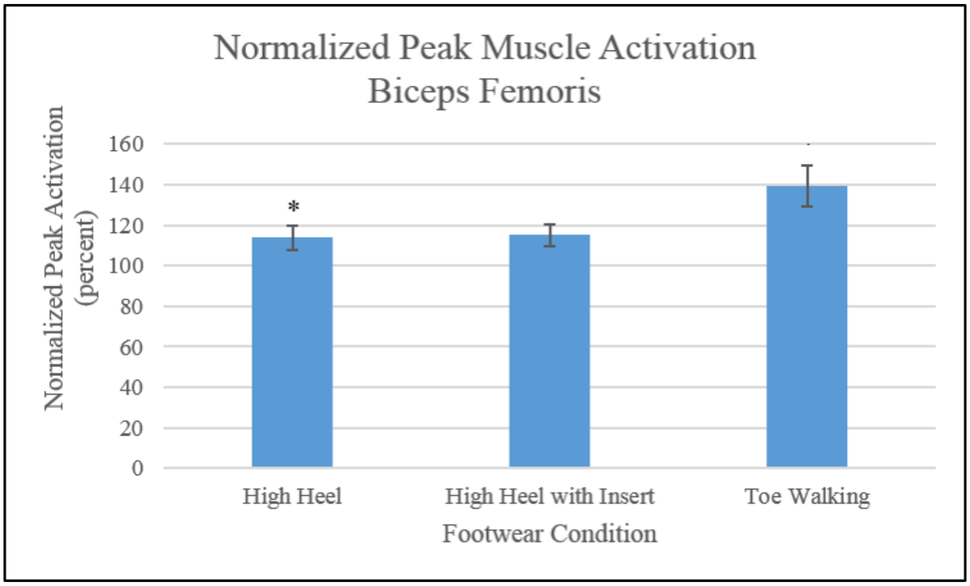

Figure 25. Effect of footwear condition on normalized peak biceps femoris activation during stance phase of gait. Notations (*) indicate significant difference between toe walking at ($p < 0.05$).

Follow-up ANOVAs found no significant condition effect for the following peak muscle activation variables: tibialis anterior, rectus femoris, rectus abdominis, or erector spinae (Table 8); indicating that footwear condition did not significantly affect peak muscle activation.

Table 8. Non-significant condition effects for sEMG peak muscle activation.

	F	df	N	p	η_p^2
Tibialis Anterior	2.456	2	30	0.104	0.149
Rectus Femoris	1.613	2	30	0.217	0.103
Rectus Abdominis	1.940	2	30	0.163	0.122
Erector Spinae	0.044	2	30	0.957	0.003

Effect of Footwear Condition on Medial Gastrocnemius Pennation Angle While Standing

Investigation of the effect of footwear on medial gastrocnemius pennation angle found significant footwear effect, $F\ (3, 30) = 178.084$, $p < 0.001$, $\eta_p^2 = 0.947$. Pennation angles were determined by measuring the angle of the medial gastrocnemius fibers from the deep aponeurosis. A larger pennation angle indicated shortening of the muscle fiber and suggested more muscle involvement. Pairwise comparisons revealed medial gastrocnemius pennation angle was significantly smaller in the barefoot condition when compared to all other conditions ($p < 0.05$) and that toe walking exhibited significantly larger medial gastrocnemius pennation angle than all other conditions ($p < 0.05$). No significant differences were detected between the high heel and high heel with insert conditions. The means and standard error for pennation angle while standing are presented in Figure 26.

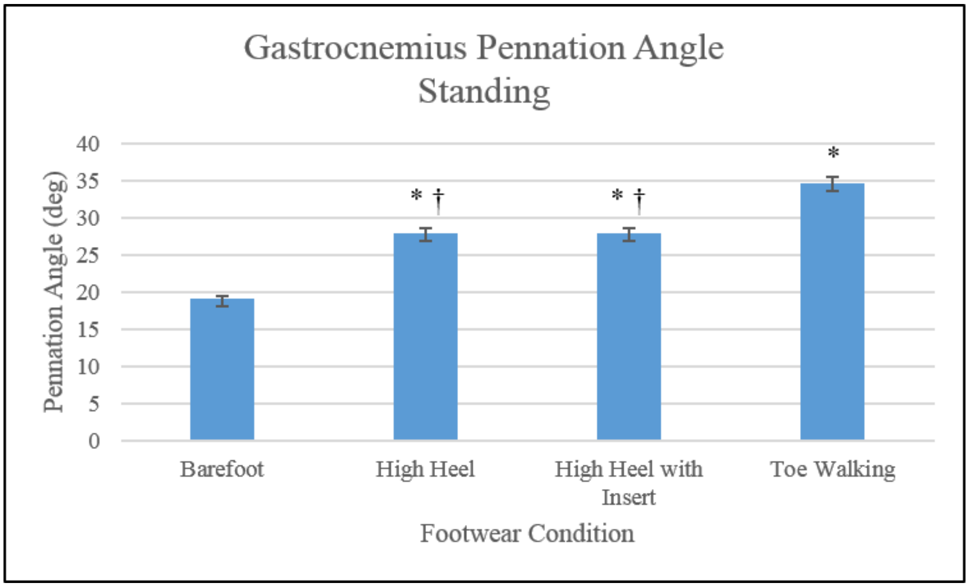

Figure 26. Effect of footwear condition on medial gastrocnemius pennation angle while standing. Notations (*) indicates significant differences when compared to barefoot at $p < 0.05$, and notations (†) indicates significant differences when compared to toe walking at $p < 0.05$.

Chapter V

Discussion

The purpose of this study was to explore kinematic, kinetic, and neuromuscular variables related to gait to ascertain if there is a continuum established with various shod conditions. Specifically, this study examined: (1) how walking with an elevated heel, such as that observed when walking in high heel shoes, high heel shoes with an insert, or toe walking, affects the kinematic and kinetic variables of gait; (2) if walking in high heel shoes with inserts brings muscle activation of the medial gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, erector spinae, and rectus abdominis closer to that found while walking barefoot; and (3) the effects of wearing high heel shoes, wearing high heel shoes with insert, or standing in plantar flexion on the pennation angle of the medial gastrocnemius. The following chapter has been divided into six sections, with the initial four sections addressing the research questions presented in the Introduction. Section 1 discusses the kinematic variables and their effect on walking in the four footwear conditions. Section 2 discusses the kinetic variables and their effect on walking in the aforementioned footwear conditions. Section 3 discusses footwear condition on muscle activation. Section 4 discusses the effect of footwear condition on gastrocnemius pennation angles while standing. Section 5 provides a summary of the findings. Section 6 discusses final conclusions from this study and recommendations for future research.

Kinematic Variable Effects

The purpose of this section of the study was to explore the effect of four footwear conditions on kinematic variables of gait. The kinematic variables of interest were knee flexion, hip flexion, distance from center of gravity to foot at initial contact (COGFC), stride length, change in height, and change in absolute height during the stance phase of gait.

Knee Flexion During Stance Phase

Knee flexion at initial contact during the stance phase of gait was measured under all four conditions resulting with the least amount of knee flexion observed while toe walking at 0.5° and the knee flexion increased in the following order, from least to greatest: barefoot (2.5°, high heel (5.1°), and high heel with insert (5.5°) (Figure 14). This is contrary to the hypothesis that the relative knee flexion would be the least in the barefoot condition, followed by toe walking, high heel with insert and high heel. It was hypothesized that knee flexion angle would be smallest during barefoot because the body would not need to counteract the anterior displacement of center of mass (COM) which occurs when walking in a plantar flexed position (Snow & Williams, 1994), whether it be high heel, high heel with insert or toe walking. Of particular interest, there were no statistical differences noted between high heel and high heel with insert. The results from the current study are in agreement with the findings of Opila-Correia (1990), Ebbeling, Hamill, and Crussemeyer (1994), Stefanyshyn, et al. (2000), and Mika, et al. (2012) and indicate a compensatory movement to ameliorate the effects of the increased vertical force at foot loading observed in the high heel and high heel with insert. With regard to toe walking, the decrease in knee flexion may be due to a change in the plantar flexor-knee extension couple that works to control the knee during gait. Brunner and Rutz (2013) describe the plantar flexor-knee extension couple as a mechanism where the knee extensors work during the first part of the loading response to control knee extension and the plantar flexors work during the second phase of the loading response to complete knee extension. It is their assertion that when the knee extensors are not engaged, toe walking occurs (Brunner & Rutz, 2013).

Maximum knee flexion during midstance followed a similar pattern to knee flexion at initial contact, where toe walking exhibited significantly less knee flexion than barefoot (6.4° and 10.1° respectively), high heel, and high heel with insert (17.8° and 18.8° respectively) (Figure16). Additionally, the barefoot condition exhibited significantly less knee flexion than high heel and high heel with insert ($p < 0.05$). However, it should be noted that there were no significant differences between high heel and high heel with insert conditions during midstance. It is speculated that the differences between the shod and non-shod conditions may be attributed to increased vertical ground reaction forces (GRF) observed while walking in high heel and high heel with insert (Figure 24) as well as the body compensating for changes in footwear condition by decreasing stride length (Figure 19) thus leading to increased maximum knee flexion found in this study. These findings seem to suggest that toe walking is more demanding than walking in high heel, high heel with insert, or barefoot. Furthermore, due to the decreased knee flexion observed in toe walking, it would appear that this could be a compensatory response for the change in foot position and lack of heel contact while walking in the toe walking condition.

Hip Flexion During Stance Phase

Hip flexion at initial contact during the stance phase of gait was measured under all four conditions resulting with the least amount hip flexion observed during toe walking, followed by barefoot, then high heel and high heel with insert condition (Figure 15). It should be noted that there were no significant differences between the toe walking and barefoot condition nor between the high heel and high heel with insert conditions with regard to hip flexion. However, significant differences were detected between the barefoot and high heel ($p < 0.05$), barefoot and high heel with insert conditions ($p < 0.05$). In addition, significant differences were found between toe walking and high heel ($p < 0.05$) and toe walking and high heel with insert conditions ($p < 0.05$). These results partially agree with the original hypothesis that hip flexion during stance phase would be smallest during barefoot, and comparable to toe walking, followed by high heel with insert, and high heel. Hip angles at initial contact are reported to vary between 20° (Perry and Burnfield, 2010) and ~35° (Gage, Deluca, & Renshaw, 1995). Data reported from this study places hip flexion at initial contact within parameters of previously published data with regard to barefoot and toe walking (31.1° to 31.4°, respectively). However, high heel and high heel with insert conditions resulted in larger hip flexion (37.5° and 38.3°, respectively), which was contrary to previous studies that reported no significant differences in hip angle with regard to heel height (Ebbeling, Hamill, & Crussemeyer, 1994; Simonsen et al., 2010; Snow & Williams, 1994). This difference in reporting may be due to Ebbeling and associates (1994) as well as Snow and Williams (1994) not comparing high heel gait to barefoot. While Simonsen and colleagues (2010) did compare barefoot to high heel gait, the height of the heel was 9 cm, not 7.62 cm which was used in this study. Furthermore, participants in the present study were allowed to walk at a preferred speed, whereas the previously mentioned studies observed gait at a set speed. These differences in variables may have influenced the results from this study. An increase in hip flexion at initial contact can be interpreted as a compensatory mechanism to the foot being forced into plantar flexion by the high heel shoe. At midstance, barefoot and toe walking conditions yielded significantly smaller hip flexion ($p < 0.05$) than when walking in either the high heel or high heel with insert condition (Figure 17). Contrary to hip flexion at initial contact, there were significant differences between barefoot and toe walking (5.7° and 8.3°, respectively) ($p < 0.05$) but no significant differences between high heel and high heel with insert (13.4° and 14.6°, respectively). This increase in hip flexion at midstance is postulated as a compensatory measure resulting from the foot being placed in plantar flexion, either it be from inserting the foot into a high heel shoe, or manually, as observed with toe walking. The increased hip flexion observed in toe walking during the current study agrees with data reported by Bovi and colleagues (2011). However, Bovi and colleagues (2011) did not indicate if natural walking was performed in a shod or unshod condition. Increased hip flexion in the high heel and high heel with insert condition observed in the current study agrees with the findings from Esenyel and colleagues (2003), yet is contrary to Ebbeling, Hamill & Crussemeyer (1994) which found no significant increase in hip flexion with high heel walking. Esenyel and associates (2003) as well as Ebbeling and colleagues (1994), Opila-Correia (1990), and Snow and Williams (1994) did not observe barefoot walking in their studies. One of the aims of the current study was to determine if the addition of the Insolia® insert to high heel shoes would bring kinematic variables closer to that found while walking barefoot.

In the case of knee and hip flexion, the addition of Insolia® inserts did not bring kinematic variables closer to those observed while walking barefoot. Instead, while statistical differences were detected between barefoot and high heel ($p < 0.05$) and barefoot and high heel with insert ($p < 0.05$), no statistically significant differences were found between high heel and high heel with insert conditions. The current study advances the body of literature by supporting the findings of Esenyel and associates (2003) as well as contributing data which compares shod and unshod conditions and reporting on the efficacy of the Insolia® insert.

Center of Gravity at Foot Contact

The horizontal distance from center of gravity (COG) of each participant to the point of foot contact (COGFC) was measured under all four footwear conditions. To the author's knowledge, only one other study explored the COGFC while wearing high heel shoes. The previous study found significant differences between barefoot and high heel ($p < 0.05$), barefoot and toe walking ($p < 0.05$), and high heel and toe walking ($p < 0.05$) with toe walking exhibiting the shortest distance from COG and heel at foot contact, followed by barefoot and high heel condition (Smallwood, et al., 2016). It was hypothesized that even though there may be no significant differences in stride length between the conditions because the change in foot position brought about by the footwear negated the inertial properties of the shod condition. However, by measuring the horizontal distance from the COG to the point of initial contact of the foot would derive a more sensitive measurement of how far an individual will reach outside of their base of support during gait (Smallwood et al., 2016). Results from the current study, however, found that the opposite occurred, with toe walking exhibiting the largest COGFC (20.9%

of body height), followed by high heel (18.8% of body height), high heel with insert (18.6% of body height), and finally barefoot (16.8% of body height) ($p < 0.05$). No significant differences were found between high heel and high heel with insert condition (Figure 18). These results are counter to what was expected. Although, the high heel with insert condition did result in a slightly smaller COGFC distance than the high heel condition, giving the appearance to trend toward the COGFC distance observed under the barefoot condition. This lack of signficance between barefoot and high heel with insert suggests that the use of the Insolia® insert may help some, but not enough to bring kinematic variables closer to that observed while walking barefoot. It should be noted that the distance measured was the horizontal distance from the COG of the individual to the heel at initial contact. Measuring to the point that the foot makes contact with the floor and not the heel may have resulted in a different outcome. Another factor to consider is that the gait trials were not controlled for speed as participants were requested to walk at a comfortable rate. Previous studies have shown an increase in speed of walking also increases knee and hip flexion as well as increasing step length (Murray, Kory, & Sepic, 1970; Schwartz, Rozumalski, & Trost, 2008). Additional investigation should be done to further clarify this variable by comparing this variable under consistent gait speed conditions.

Stride Length

Stride length under all four footwear conditions was evaluated in this study (Figure 19). Results indicate that while under the barefoot condition, stride length was significantly longer than the other conditions (84.6% of body height) ($p < 0.05$). However, no significant differences in stride length were detected between high heel

(80.7% of body height), high heel with insert (79.5% of body height), and toe walking (78.0% of body height). The results from the current study do not agree with the hypothesis that stride length would be shortest in toe walking condition followed by high heel, high heel with insert, and finally barefoot with the longest stride length. However, if high heel and high heel with insert were viewed separately, then a continuum where toe walking exhibited the shortest stride length, followed by high heel (or high heel with insert) and barefoot could be observed. Additionally, the results from the current study are in agreement with other studies that explored the effects of walking in high heel shoes which found that stride length decreased when heel height increased (Esenyel, Walsh, Walden, & Gitter, 2003; Opila-Correia, 1990). Interestingly, Snow and Williams (1994) reported that stride length significantly decreased when comparing medium height heels (3.81 cm) to low heels (1.91 cm) but no signficant differences were found when comparing stride length while walking in the highest heel (7.62 cm) to the medium or low heels. The investigators attributed this result to the decrease in walking speed observed when heel height increased (Snow & Williams, 1994). With regard to toe walking, Bovi and associates (2010) as well as Davids and colleagues (1999) reported normal walking, that is, walking barefoot in a heel-to-toe fashion, displayed longer strides lengths than toe walking. However, these results are contrary to those reported by Simonsen and colleagues (2012) who found no signficiant differences in stride length when walking at a fixed-speed over ground. The lack of signficance between high heel and high heel with insert conditions indicate that the addition of the Insolia® did not help to bring stride length closer to barefoot condition.

Except for the study performed by Bovi and associates (2010), stride lengths were reported as linear measurements and not normalized to the height of the participant. The current study normalized stride length, using the height of the participant, which made stride length comparisons between conditions and participants more equitable. Additionally, it appears that observing gait at a self-selected speed can result in larger differences in stride length between footwear conditions not normally found when observing gait at a fixed speed. Since the current study and Bovi et al (2010) study did not use a set walking speed, stride length results should be read with caution. While it is this author's opinion that observing gait at a self-selected speed will better reflect real world conditions, more research should be performed on this variable controlling for gait speed as well as normalizing to body height in order to further determine the sensitivity of this measurement.

Change in Height and Change in Absolute Height

Change in participant height and normalized change in participant height were observed under all four conditions. The least change in participant height and normalized change in participant height was observed in the barefoot condition (Figures 20 and 21) which was significantly smaller than the other conditions ($p < 0.05$). This variable is of interest because during normal gait, the body attempts to move through space utilizing the least amount of energy and minimizing change in height. During gait, the body undergoes sinusoidal curves in the sagittal and transverse plane. Flattening of the sinusoidal curve in the sagittal plane is associated with conservation of the energy used during gait (Saunders, Inman, & Eberhart, 1953; Inman, 1966). No significant differences were found between toe walking, high heel with insert, and high heel conditions in change of participant height or normalized change in participant height. This is the first study, to the author's knowledge, that addresses change in height while toe walking. These data do not agree with the hypothesis that change in height and normalized change in height would follow a progression from smallest to largest: barefoot, toe walking, high heel with insert, and high heel. Saunders and colleagues (1953) hypothesized that more energy was needed to accomplish a gait cycle if the sinusoidal curves, produced during gait, were larger. Change in absolute height can be perceived as an indicator as to whether an individual is walking efficiently. Data from this study reported height change ranging from 3.8, 4.5, 4.7, and 4.7 cm for the barefoot, toe walking, high heel with insert, and high heel conditions respectively. Although walking speed was not studied nor controlled for this investigation, it appears that change in height under barefoot and toe walking conditions as observed in the current study fall into distances previously reported by Gard and Childress (2001). However, high heel and high heel with insert conditions resulted in height changes larger than what Saunders and colleagues (1953) reported for males. Results from this study indicate that walking in either the high heel or high heel with insert condition does not lead to a more efficient gait if one was to compare change in height. The designer of the Insolia® insert claims this orthotic reduces energy consumption when worn with high heel shoes (Curran, Holliday, & Watkeys, 2010). Their investigation found that individuals walking in high heel shoes with the Insolia® insert exhibited lower heart rate, lower volume of oxygen consumed, a lower respiration exchange ratio (carbon dioxide/oxygen consumption), fewer number of steps taken, and a lower physiological cost index ((mean heart rate at work – mean heart rate at rest)/walking speed) when walking on a treadmill at 4.2 km/hour (Curran, Holliday, &

Watkeys, 2010). While physiological measurements were not taken in the present study, if change in participant height is used as a measure of efficiency while walking, then there was no difference in efficiency between the high heel and high heel with insert condition. This calls into question the effectiveness of the Insolia® insert. The increase in change in participant height when comparing toe walking to barefoot may be attributed to decreased knee flexion while in the toe walking condition. In light of these conflicting results, more research should be performed on this variable with the addition of walking speed and physiological cost index as factors for measurement to help determine the efficacy of these measurements and this orthotic insert.

Kinetic Variable Effects

The purpose of this section of the study was to explore the effect of four footwear conditions on kinetic variables of gait. Kinetic variables of interest were anterior/posterior ground reaction forces (GRF) and vertical GRF, all normalized to bodyweight, during the stance phase of gait. First, braking and propulsive forces in the anterior/posterior direction will be addressed followed by heel strike transient, loading response, and finally propulsive forces in the vertical direction.

Anterior/Posterior Ground Reaction Forces

Anterior/posterior GRF were examined to investigate braking and propulsion forces during the stance phase of gait, under all four footwear conditions. The only significant differences found were between barefoot and high heel with insert and toe walking (17.6%, 19.8%, and 20.3% of body weight, respectively) ($p < 0.05$) with the least amount of braking observed under the barefoot condition ($p < 0.05$) (Figure 22). No other significant differences were found between barefoot and high heel (19.3% of body weight) or the remaining footwear conditions. The results from the current study do not agree with the hypothesis that the least amount of braking would be detected with barefoot, followed by high heel, high heel with insert, and toe walking. In addition, the results of the current study concur with Snow and Williams (1994) who hypothesized that the foot was unable to fully pronate because the shape of the shoe forced the foot to continue in plantarflexion during the stance phase of gait. Ebbeling, Hamill, and Crussemeyer (1994) also reported similar braking when examining high heel gait and attributed this to the inability of the foot to engage in the amount of dorsiflexion observed while walking in low heeled shoes and this therefore hindered attenuating the ground reaction forces via eccentric contraction of the dorsiflexors. Stefanyshyn and colleagues (2000), reported larger braking forces with shoes with heels up to 5.4 cm (2 1/8") and smaller braking forces with shoes that had an 8.5 cm (3 1/3") heel. Since Stefanyshyn and colleagues did not employ a shoe with a 7.62 cm (3") heel, it may be inferred that defense mechanisms of the body do not engage until a heel height larger than the one used in the current study is utilized. With regard to toe walking condition, data from the current study agrees with Bovi and associates (2010) which found increased peak braking in the toe walking condition when compared to normal walking among adults. During normal gait, when braking occurs, the center of mass is behind the foot when it makes contact with the ground. However, when walking in a plantar flexed position, whether it be in high heel, high heel with insert or toe walking, the COM has moved anteriorly to that observed while walking in a lower heel shoe (Snow & Williams, 1994; Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). In addition, when walking in high heel shoes, high heel with insert, or toe walking the angle of approach the foot employs when making contact with the ground due to plantar flexion is different to that of a person walking barefoot (Ebbeling, Hamill, & Crussemeyer, 1994). This change in COM and angle of approach, as well as the inability of the foot to fully utilize dorsiflexion, possibly contributes to the increase in braking observed in the high heel with insert and toe walking condition. The lack of statistical significance between high heel and high heel with insert indicates that the orthotic insert used for this study did not sufficiently alter braking as compared to walking in high heel condition.

Footwear effects on the magnitude of propulsion forces while walking, resulted in no statistically significant differences. These results are intriguing because there were statistically significant differences with regard to braking (Table 9). When measuring braking and propulsion during normal gait, nearly symmetrical forces are expected when comparing these two variables while walking at a constant rate (Marasovic, Cecic, & Zanchi, 2009). The lack of statistically significant differences between footwear conditions during propulsion observed in the current study agrees with Ebbeling and colleagues (1994) who found no significant differences between heel height and propulsive force. In addition, Bovi and associates (2010) as well as Raha and colleagues (2006) found no significant differences in propulsive force when comparing normal walking to toe walking. However, this data ran counter to the hypothesis presented for this study as well as studies performed by Stefanyshyn et al., (2000) and Snow and Williams (1994) who reported an increased propulsive force when comparing medium height heels to low heels. Stefanyshyn and colleagues (2000) suggested that since larger forces occurred during braking to slow the COM, larger forces were necessary to accelerate the COM forward, resulting in a less fluid gait pattern while walking in high heel shoes. Absence of significant differences between high heel and high heel with insert indicates the orthotic used in this study did not alter propulsive forces while walking in high heel shoes.

Table 9. Effect of footwear condition on Anterior/Posterior GRF variables.

Parameter	**Footwear Condition** Barefoot	 High Heel	 High Heel Insert	 Toe Walking	**Main Effects** ***F***	 ***p* Value**
Braking	17.6*	19.3	19.8*	20.3*	5.010	0.005
Propulsion	24.2	24.1	23.7	26.3	2.553	0.074

* = significant difference between barefoot. All indicated differences were significantly different (p < 0 .05).

Vertical Ground Reaction Forces

The purpose of this section of the study was to examine vertical GRF, to investigate heel strike transient, foot loading, and propulsion, under all four footwear conditions. A heel strike transient may be observed in some, but not all individuals (Jefferson, Collins, Whittle, Radin, & O'Connor, 1990) and can occur in the shod or unshod conditions (Simon, et al., 1981). The heel strike transient is an inflection point which occurs between initial contact and weight acceptance during the gait cycle and is detected as a short, sharp spike in force in vertical GRF that lasts approximately 10 – 20 ms (Whittle, 2007; Whittle, 1999) (Figure 31). Larger heel strike transients are thought to occur as a result of the individual utilizing the ground to slow down the limb, while smaller heel strike transients are considered to be a result of an individual using muscle activity to slow down the limb just prior to ground contact (Jefferson, Collins, Whittle, Radin, & O'Connor, 1990). It has been suggested in literature that there may be a correlation between heel strike transients and joint degeneration, specifically osteoarthritis and lower back pain (Collins & Whittle, 1989). Results from the current study found statistically significant larger heel strike transients in the high heel (47.2% of body weight) and high heel with insert (50.0% of body weight) condition when compared to the barefoot (23.9% of body weight) condition ($p < 0.05$) (Figure 23). This does not agree with the hypothesis that the barefoot condition would exhibit the least amount of force during the transient followed by high heel and high heel with insert because there was no statistically significant difference between the high heel and high heel with insert conditions. The lack of statistical differences between the high heel and high heel with insert conditions indicate that the Insolia® insert did not bring heel strike transients closer to the barefoot condition. Results from this the current study agree with the findings presented by Snow and Williams (1994) which postulated an increase in heel strike transient was a result of a perception of instability while walking in shoes with a high heel. Jefferson and associates (1990), however, attributed an increase in heel strike transient to the activation of hamstrings, quadriceps, and gastrocnemius at foot contact as a method to stabilize the knee at heel strike. Because there were no significant differences between the high heel and high heel with insert condition, the author believes the addition of the orthotic did not contribute to attenuation of the heel strike transient at the start of the gait cycle. In addition, based on the findings of Collins and Whittle (1989), the additional joint loads observed while walking in high heel shoes may be contraindicated for those who are afflicted with knee osteoarthritis. Heel strike transients were not measured for the toe walking condition because the foot is in constant plantar flexion during the gait cycle and the heel does not strike the ground.

Loading response occurs after heel strike transient during a period of double support between initial contact and opposite toe off. At this time the foot is lowered to the ground via plantar flexion as GRF increases rapidly (Whittle, 1996). Shock absorption, which began during initial contact, continues as body weight is transferred from the trailing to the leading limb (Perry & Burnfield, 2010). Loading response is indicated by a peak in vertical GRF (Figure 31). At normal walking speed, Perry and Burnfield (2010) report this peak is approximately 110% of the body weight of an individual (Perry & Burnfield, 2010). Results from the current investigation found that the loading response in the barefoot condition was smaller than the loading response in the high heel, high heel with insert, or toe walking condition ($p < 0.05$) (Figure 24). Comparisons amongst the other conditions revealed no statistical differences. These results do not agree with the hypothesis that the barefoot and toe walking conditions would exhibit the least amount of vertical GRF, followed by high heel with insert and high heel conditions as a continuum was not observed. Results from the current study agrees with the study by Ebbeling and colleagues (1994) which reported loading response increased as heel height increased. This increase in loading response, Ebbeling and associates (1994) opined, was a result of the lack of dorsiflexion detected in the ankle due to the foot being placed in a plantar flexed position by the high heel shoe and therefore the body could not attenuate the GRF to the degree observed while walking in flat heel shoes. Stefanyshyn and associates (2000) reported similar results indicating that as heel height increased, so did loading response, except for the highest heeled shoe (8.5cm) which displayed a lower loading response. Stefanyshyn and colleagues (2000) opined that the lower loading response discerned in the highest heel shoe was a compensatory act to decrease impact and loading, in an effort to protect the body from injury. A study by Snow and Williams (1994) found increased loading responses when heel height increased. Snow and Williams (1994) believed this increase in loading response resulted from the increase in plantar flexion and alteration in center of gravity associated with increasing heel height. A main difference between the above mentioned studies and the current study is that the previous investigators observed walking in different heel heights and not during a barefoot condition, thus this project advances the previous studies and adds to the body of knowledge by comparing high heel gait to barefoot gait. Additionally, barefoot gait was compared to a condition with a foot position similar to high heel gait. With regard to toe walking, Bovi and associates (2010) reported an increase in peak loading response in the toe walking condition when compared to normal walking. This concurs with the results of the current study. It appears that it is the position of the foot, and not necessarily the shod/unshod condition of the foot, that affects the loading response observed in this study.

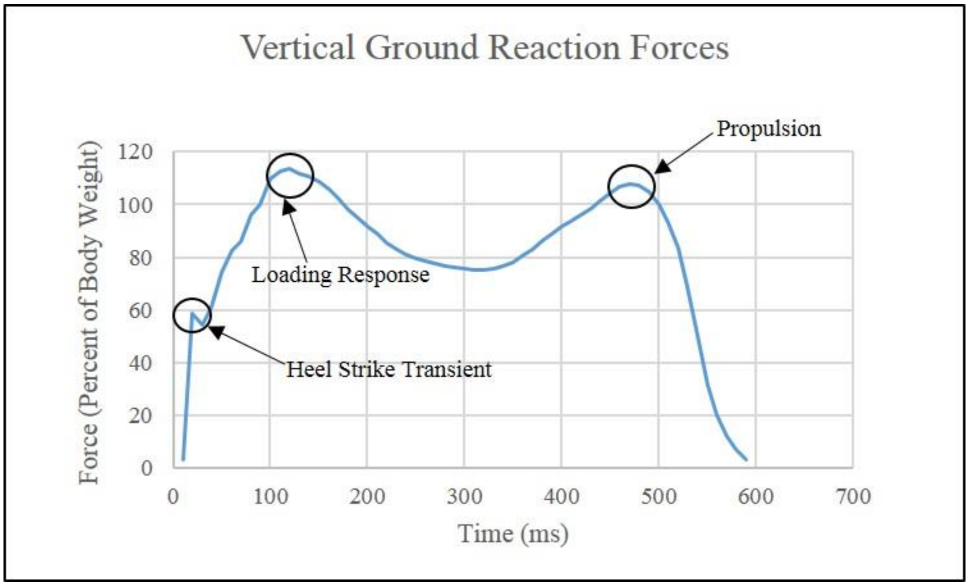

Figure 27. Vertical ground reaction forces over stance phase of gait.

Vertical force at propulsion occurs near the end of stance phase from terminal stance through preswing (Figure 27). Peak propulsive vertical GRF under the barefoot condition was found to be significantly smaller than the toe walking condition ($p < 0.05$). No significant statistical differences were found between barefoot and high heel, or high heel with insert condition. In addition, no significant differences were found between high heel and high heel with insert, or toe walking and high heel or high heel with insert conditions (Figure 21). These results do not agree with the hypothesis that the barefoot and toe walking conditions would exhibit the least amount of propulsive vertical GRF followed by high heel with insert and high heel conditions. Additionally, these results differ from previous data by Ebbeling and colleagues (1994), who described a trend towards larger propulsive forces with higher heels but with only a significant difference between 1.25 cm (1/2”) and 7.62 cm (3”) heels and opined that this was due to the more forward position of the COM brought about by wearing high heel shoes. Snow and Williams (1994) found significant increase in propulsive force with increase in heel height (1.91 cm (3/4”), 3.81 cm (1 1/2”), and 7.62 cm (3”)) also do not agree with the aforementioned hypothesis. Similarly, Stefanyshyn and associates (2000) reported an increase in maximal vertical propulsive force when comparing walking in flat heel shoes to 3.7 cm and 8.5 cm heeled shoes. This increase was attributed to the increased braking force necessitating larger propulsive forces to accelerate the COM during toe off. While these previous studies are similar, they did not compare high heel shoes to barefoot condition. Data from the current study do not agree with Bovi and associates (2010), who reported no significant differences between normal walking and toe walking when comparing propulsive vertical GRF. Thus, it is reasonable to conclude that the high heel with insert condition did not affect propulsive GRF when compared to high heel or even barefoot or toe walking conditions. Additionally, the high heel with insert condition did not affect any of the GRF measures.

Electromyographic Variable Effects

The purpose of this section of the study was to explore the effect of four footwear conditions on the muscle activity (EMG) of six muscles during the gait cycle. Dependent variables of interest were mean and peak muscle activation of the following muscles: gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, erector spinae, and rectus abdominis. Table 10 illustrates the sEMG (normalized mean and peak) differences for footwear conditions as compared to the barefoot condition.

Table 10. Effect of footwear condition on muscle activation as compared to barefoot.

Muscle	**Footwear Condition (compared to Barefoot)**		
	High Heel	High Heel Insert	Toe Walking
Gastrocnemius $_{mean}$	↑*	↑*	↑
Tibialis Anterior $_{mean}$	↑	↑	↑
Rectus Femoris $_{mean}$	↑	↑	↑
Biceps Femoris $_{mean}$	↑*	↑	↑
Rectus Abdominis $_{mean}$	↑	↑	↑
Erector Spinae $_{mean}$	↑	↑	↑
Gastrocnemius $_{peak}$	↑*	↑*	↑
Tibialis Anterior $_{peak}$	↑	↑	↑
Rectus Femoris $_{peak}$	↑	↑	↑
Biceps Femoris $_{peak}$	↑*	↑	↑
Rectus Abdominis $_{peak}$	↑	↑	↑
Erector Spinae $_{peak}$	↑	↑	↑

Note: ↔ = no increase, ↑ = increase, * = significant difference between toe walking. All indicated differences were significantly different ($p < 0 .05$).

The results from the current study compare EMG as a percentage of barefoot EMG and agree with the hypothesis that there would be significant differences in mean and peak activity in the gastrocnemiuis and biceps femoris, with the toe walking condition exhibiting an increase in mean and peak gastrocnemius activity when compared to high heel ($p < 0.05$) and high heel with insert ($p < 0.05$). In addition, a statistically significant increase in mean and peak biceps femoris was observed when comparing high heel and toe walking conditions ($p < 0.05$). It should be noted that high heel and high heel with insert did not display significant differences in mean or peak activity in any of the muscles measured, thus questioning the ability of the Insolia® insert to alter neuromuscular demands. The results from the current study, therefore, do not support the claim by the designers of Insolia® that this insert brings walking in high heel with insert closer to the barefoot condition.

During normal gait, the role of the gastrocnemius is to slow the rate of tibial advancement during mid-stance and the first half of terminal stance (Perry & Burnfield, 2010), thus contributing to the stability of the body during this action. Simon and colleagues (1978) reported that the gastrocnemius, in conjunction with the soleus, slows down the forward motion of the body which, in turn, allows the body to lean further forward beyond its base of support during gait. During the present study it was found that toe walking gait exhibited significantly higher peak and mean EMG activity than high heel and high heel with insert conditions (Figures 22 and 24). However, high heel and high heel with insert conditions were 127.3% and 129.2%, respectively, of the barefoot condition, indicating that the shod condition does increase the gastrocnemius muscle activity while walking. Additionally, the high heel with insert condition did not decrease the demand of the gastrocnemius which indicates that the addition of the Insolia® insert did not bring the neuromuscular demand closer to the barefoot condition. In addition, under toe walking condition, gastrocnemius activation was 198.2% of the barefoot condition, indicating that the position of the foot also plays a role in the increase of gastrocnemius muscle activity. Increased gastrocnemius activity while in the high heel condition is understandable because there was an increase in knee flexion in the high heel and high heel with insert condition. Thus agreeing with Mika and associates (2011a) and Nascimento (2014) but disagreeing with findings from Stefanyshyn and colleagues (2000), and Simonsen and associates (2012) who reported no significant increase in gastrocnemius activity. Perry and colleagues (2003) reported increased medial gastrocnemius muscle activation during toe walking and speculated that this was necessary in order to stabilize the plantar flexed ankle. In addition, Perry and colleagues (2003) further noted that when the foot was in plantarflexion, the muscles were shortened due to the position of the foot which reduced the strength of the plantar flexors and the contributions of the passive structures utilized while walking. In other words, positioning the foot in plantarflexion resulted muscle fascicles placed in a position not optimal for force production. Additionally, positioning the foot in plantarflexion, as observed with the high heel, high heel with insert, and toe walking conditions utilized in the current study, placed the actin-myosin cross-bridge formation in possibly a less than efficient position. As a result, more muscle activity was necessary in order to complete the toe walking task (Perry, Burnfield, Gronley, & Mulroy, 2003). However, while these comparisons are in agreement with Mika and associates (2011a), Nascimento (2014) and

Perry and colleagues (2003), they should be read with caution as the technique used to normalize the EMG data in this study was different to that used in the other studies.

Significant differences in mean and peak biceps femoris muscle activity when comparing toe walking to high heel condition ($p < 0.05$) were also detected (Figures 23 and 25), but no significant differences between high heel with insert and high heel or high heel with insert and toe walking conditions were present. It should be noted that the high heel, high heel with insert, and toe walking conditions exhibited biceps femoris activity of 113.95%, 115.19% and 139.25%, respectively, of the barefoot condition. This indicates that the shod condition and position of the foot affects biceps femoris activity while walking. These results agree with findings from Simonsen and colleagues (2012) which reported an increase in biceps femoris activity when comparing barefoot walking and walking in 9 cm high heel shoes, but do not agree with Stefanyshyn and associates (2000) which did not show a graded response to heel height and biceps femoris activity, nor Mika and associates (2011(a)), which reported no significant increase in biceps femoris activity when comparing barefoot to high heel walking. As a muscle that crosses two joints, the biceps femoris slows down hip flexion during terminal swing in order to prepare the limb for weight acceptance during the gait cycle (Perry & Burnfield, 2010). The biceps femoris also assists with knee flexion during loading response (Perry & Burnfield, 2010). An increase in biceps femoris activity while walking in high heel shoes is feasible, as the current study found, that during high heel and high heel with insert gait, increased knee and hip flexion was observed at IC. However, with regard to the knee, the biceps femoris should be understood as a weak knee flexor at IC. Lack of significant difference between high heel and high heel with insert conditions indicate to the author that the high heel with insert condition did not appreciably alter biceps femoris activation when compared to walking under the high heel condition. The increase in biceps femoris activity during the toe walking condition noted in the current study, was also reported by Bovi and associates (2011). While Kerrigan and colleagues (2000) did not use EMG to record muscle activity, they did report increased strength, power, and work by hip extensors was necessary to extend the hip during toe walking. This could be inferred as a mechanism to reduce the work performed by plantarflexors during the gait cycle as well as the decrease of work by knee extensors at loading response during toe walking (Kerrigan, Riley, Rogan, & Burke, 2000), making the increase in biceps femoris activity likely. While these comparisons are similar to Simonsen and associates (2012) and Bovi and colleagues (2011), they should be read with caution as the technique used to normalize EMG data in this study differed from that used in the previously mentioned studies.

During gait, the tibialis anterior controls the descent of the foot to the ground during initial contact (Gage et al., 1995; Perry and Burnfield, 2010) and mitigates vertical forces produced during this time period (Ebbeling, Hamill & Crussemeyer, 1994). Studies involving walking in high heel shoes reported mixed results with some finding no significant differences (Simonsen et al., 2012; Stefanysyshyn et al., 2000; and Lee, Jeong, & Freivalds, 2001) and others reporting increased tibialis anterior activation (Nascimento et al., 2014; Mika et al., 2011a; and Cronin, Barrett, & Carty, 2012). On the other hand, Bovi and associates (2010) reported decreased tibialis anterior activity when toe walking. While the tibialis anterior failed to reach statistical significance between high heel, high heel with insert, and toe walking, the author believes it should be noted that tibialis anterior muscle activity was 133.28%, 138.37%, and 147.94% of barefoot condition, respectively. This increase in muscle activity, although not significantly different, indicates to the author that while both the shod and plantarflexed position of the foot detected in high heel, high heel with insert, and toe walking condition appears to affect tibialis anterior activation when compared to the baseline, which was the barefoot condition, the shod condition may afford some level of outside support for the foot, whereas the toe walking condition experienced no outside support. Although these comparisons agree with Nascimento and colleagues (2014), Mika and associates (2011a), and Cronin and colleagues (2012), they should be read with caution as the EMG data normalization technique used in this study was different from that used in the previously mentioned studies. In addition, the current study found an increase in normalized peak vertical ground reaction force at foot loading when comparing barefoot to the other footwear conditions. This indicates an alteration in gait kinetics and possibly decreased ability to mitigate vertical forces while walking in a plantar flexed position. Due to no significant findings between high heel and high heel with insert conditions, the author believes the addition of the Insolia® insert did not sufficiently ameliorate the effects of walking in the high heel condition.

The rectus femoris acts with the quadriceps to thwart excessive knee flexion during late pre-swing and early initial swing phase of gait (Whittle, 1996; Perry & Burnfield, 2010). Simonsen and colleagues (2012) reported an increase in rectus femoris activity and opined that this was in response to increased knee flexion during stance. Stefanyshyn and associates (2000) attributed the increase in rectus femoris activity to the anterior movement of center of mass and subsequent increase in knee flexion when walking in the high heel condition. Nascimento and colleagues (2014) found increased rectus femoris activtion during midstance for barefoot condition and terminal stance for high heel condition as well as swing phases of gait and attributed this increase as an attempt to stabilize the body and minimize the risk of falling while walking in the high heel condition. Mika and associates (2011a) also reported increased rectus femoris activity at initial contact and toe off. They speculated this as a compensatory measure, in conjunction with increased knee flexion, to mitigate the increase in ground reaction forces while walking in the high heel condition (Mika, Oleksy, Mika, Marchewka, & Clark, 2011a). Bovi and associates (2010) also reported increased rectus femoris activity while in the toe walking condition when compared to heel-toe walking. However, no explanation was given by Bovi and colleagues as to why this occurred. It was concluded that the combination of increased plantar flexion and decreased knee flexion contributed to an increase in rectus femoris activity as a mechanism to maintain stability in the distal segments while engaging in the toe walking condition. Results from the current study inidicate no significant differences in rectus femoris activity between high heel, high heel with insert and toe walking conditions. Rectus femoris activation for high heel, high heel with insert, and toe walking was 209.95%, 221.38%, and 237.81% of barefoot activation, respectively. Although not statistically significant, this indicates to the author that the plantar flexed position of the foot greatly increases rectus femoris activation while the body attempts to mitigate this position by restricting knee flexion and increase shock absorption (Perry & Burnfield, 2010; Ebbeling, Hamill & Crussemeyer, 1994; Whittle, 2007). While these comparisons agree with Simonsen and associates (2012), Stefanyshyn and colleagues (2000), Nascimento and associates (2014), Mika and colleagues (2011a), and Bovi and associates (2011), they should be read with caution as the technique used to normalize the EMG data in this study differed from that used in the previously mentioned studies..Lack of significant differences between high heel and high heel with insert conditions indicate the addition of the insert did not appreciably affect walking in high heel shoes.

Often considered postural musculature, the rectus abdominis and erector spinae work together during gait to stabilize the pelvis during stance and swing phases. A main trunk flexor (Nordin & Frankel, 2012), the rectus abdominis is also involved in posterior pelvic rotation and weakly associated with lateral trunk flexion (Floyd, 2012). Conversely, the erector spinae works to inhibit trunk flexion brought about by inertia following heel contact (Watanabe, 1996). Results from the current study indicate no significant differences in rectus abdominis or erector spinae activity between high heel, high heel with insert and toe walking conditions.

Pennation Angle Variable Effects

The purpose of this section of the study was to explore the effect of four footwear conditions on pennation angle of the gastrocnemius while standing. Significant main effects were found for footwear condition. This indicates that the position of the ankle affects the pennation angle of the gastrocnemius. Table 11 outlines the pennation angle differences when comparing toe walking, high heel and high heel with insert to the barefoot footwear condition. What is notable is that when participants actively put their foot into plantarflexion, the gastrocnemius pennation angle increased.

Table 11. Effect of footwear condition on pennation angle as compared to barefoot.

Position	**Footwear Condition (compared to Barefoot)**		
	High Heel	High Heel Insert	Toe Walking
Standing	↑* #	↑*#	↑*

Note: ↔ = no significant difference, ↑ = increase, * = significant difference between barefoot. # = significant difference between toe walking. All indicated differences were significantly different ($p < 0.05$).

While in the standing position, there appears to be a continuum of pennation angle with barefoot exhibiting the smallest pennation angle (19.12°), followed by high heel (27.95°), high heel with insert (27.96°), and finally toe walking (34.67°). Significant differences were found between barefoot and high heel ($p < 0.05$), barefoot and high heel with insert ($p < 0.05$), and barefoot and toe walking ($p < 0.05$). Additionally, significant differences were found between toe walking and high heel ($p < 0.05$), and toe walking and high heel with insert ($p < 0.05$). This partially agrees with the hypothesis of this study which put forth that the gastrocnemius pennation angle would be largest while standing in the toe walking condition and the smallest pennation angle would be observed in the barefoot condition. There were no significant differences between the high heel and high heel with insert condition with regard to pennation angle. However, the pennation angles in the high heel and high heel with insert conditions were significantly smaller than toe walking ($p < 0.05$) and significantly larger than barefoot ($p < 0.05$). These results indicate to the author that although the orthotic used in this study is purported to shift the weight of the body from the forefoot to the heel, this did not affect the pennation angle of the gastrocnemius. This is understandable because while the body weight may have moved posteriorly, it did not affect the position of the ankle appreciably. However, knee angle was not measured during pennation angle measurements. It may be possible that the stances taken under each condition while pennation angles were measured were different. Muscle activity also changes pennation angle. Interestingly, gastrocnemius activity was significantly lower in high heel when compared to toe walking ($p < 0.05$) and in high heel with insert when compared to toe walking ($p < 0.05$). Comparisons of gastrocnemius activity, however, was not significantly different between high heel to high heel with insert and neither was gastrocnemius pennation angle. Narici and associates (1996) reported that the pennation angle for the medial gastrocnemius can be affected by the position of the ankle and the intensity of an isometric contraction. It is interesting to note that the pennation angle increased as the standing participants moved from barefoot, where the foot was completely on the ground, to high heel and high heel with insert, which involved wearing shoes with a 3” heel, and then toe walking, which involved standing on the forefoot only. The change in pennation angle indicates that the gastrocnemius adapted its length as a result of change in foot and ankle position. It appears that the shod condition, even with a 3” heel, affords some support to the foot while walking in a plantarflexed position. This increase in pennation angle further confirms the findings by Narici and colleagues (1996). It should also be noted that these results are well within the range of gastrocnemius pennation angle change which occur during low level voluntary contractions as reported by Lieber and Fridén (2000) suggesting that standing in high heel, high heel with insert or toe walking require only a low level of activation.

Section 5: Summary of Findings

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

- Part 1: Summarize the discussion comparing the effects of footwear condition on kinematic variables.
- Part 2: Summarize the discussion comparing the effects of footwear condition on kinetic variables.
- Part 3: Summarize the discussion comparing the effects of footwear condition on muscle activation.
- Part 4: Summarize the discussion comparing the effects of footwear condition on the pennation angle of the gastrocnemius while standing.

Summary: Part 1

Kinematic Variables

Knee and hip flexion were observed while walking in four footwear conditions. At initial contact, it was found that toe walking resulted in the least amount of knee flexion, followed by barefoot and then high heel and high heel with insert. There were significant increases in knee flexion at initial contact and increased maximum knee flexion observed in the high heel and high heel with insert condition when compared to barefoot and toe walking condition. It should be noted that no significant differences were found between high heel and high heel with insert condition with regard to kinematic variables. Increased knee flexion in the high heel condition was expected as it has been speculated as a method to compensate for the ankle being forced into a plantar flexed position (Esenyel, et al., 2003; Simonsen et al., 2012; Mika et al., 2012).

Previously it has been reported that increased knee flexion is an attempt by the body to absorb in impact of the heel strike at initial contact (Ebbeling et al., 1994). In addition, increased rectus femoris activity was observed during the stance phase of high heel and high heel with insert gait. Rectus femoris activity during stance phase is a mechanism used by the body to slow the rate of knee flexion (Whittle, 2007). Previous studies reported increased rectus femoris activity in high heel gait (Stefanyshyn et al., 2000). It was not surprising that the toe walking condition exhibited smaller knee flexion than high heel and high heel with insert. Perry and associates (2003) reported no significant differences in knee flexion during stance phase when comparing toe walking to barefoot gait. Brunner and Rutz (2013) also postulated that the decrease in knee flexion while toe walking as an effort to increase leg stability at initial contact.

The least amount of hip flexion at initial contact was found in the barefoot condition. When compared to barefoot condition, high heel and high heel with insert exhibited significantly more hip flexion. In addition, high heel and high heel with insert displayed significantly larger hip flexion than toe walking at initial contact and at midstance. This does not agree with the hypothesis that the barefoot and toe walking condition would exhibit the least and comparable amount of hip flexion followed by high heel and high heel with insert conditions as the high heel and high heel with insert conditions did not differ significantly. Research reports hip flexion at initial contact can vary from between 20° (Perry and Burnfield, 2010) to ~35° (Gage, Deluca, & Renshaw, 1995). Hip flexion data from this study revealed hip flexion at initial contact within parameters of previously published data. At midstance, walking under the toe walking and barefoot conditions yielded significantly less hip flexion than when walking in the high heel or high heel with insert condition. Decreased hip flexion is understandable because at midstance the hip should be moving into extension. Similar to hip flexion at initial contact, no significant differences were found between high heel and high heel with insert. Increased hip flexion in the high heel condition concurs with the findings from Esenyel and colleagues (2003), yet is dissimilar to Ebbeling, Hamill & Crussemeyer (1994) as no significant increase in hip flexion with high heel walking was found in these projects. The orthotic used in the current study was designed to position the foot so that it gradually raises the area of the calcaneus proximal to the tuberosity of the calcaneus as well as gradually raise the second and third metatarsals and thereby shift some of the pressure borne by the forefoot while wearing high heel to the heel (USA Patent No. 7962986 B2, 2011). Dananberg and Trachtenberg further explained that the Insolia® insert would position the foot so that the talus movement would be restricted as compared to the movement observed while walking in high heel alone, thus altering how an individual walks in that footwear condition (Dananberg & Trachtenberg, 2000). In other words, the depression in the heel of the orthotic should move the foot so that the high point of contact is further away from the forefoot and now at a point closer to the posterior midfoot. However, the lack of significance between high heel and high heel with insert indicates this orthotic did not affect this population as intended.

To the author's knowledge, this is only the second study to explore the COGFC while wearing high heel shoes. The first study reported signficant differences in distances from center of gravity to point of initial contact between between barefoot and high heel, barefoot and toe walking, and high heel and toe walking with toe walking exhibiting the shortest distance followed by barefoot and high heel (Smallwood, et al., 2016).

Smallwood, et al., (2016) found that measuring the horizontal distance from the COG to the point of intial contact of the foot would provided a more sensitive measurement of how far an individual will reach outside of their base of support during gait than by just measuring stride length. Results from the current study ran contrary to the Smallwood, et al., (2016) study with toe walking exhibiting the largest distance followed by high heel, high heel with insert, and finally barefoot ($p < 0.05$). However, it should be noted that the previous study only observed seven individuals, whereas the present study observed 33 individuals. Additionally, no significant differences were found between high heel and high heel with insert condition, indicating that the addition of the insert in the high heel with insert condition did not appreciably alter the COGFC distance observed in the high heel condition. Increased vertical propulsion may be a factor in the longer COGFC detected in the toe walking condition. Larger propulsive forces are indicitive of an increase in acceleration at toe off. Additionally, larger propulsive forces in the toe walking condition may also lead to the amplified braking at loading response. As each participant was allowed to walk at a self selected speed, it may be prudent to repeat the study with a predetermined velocity. Additional investigation should be done to further clarify this variable.

Results from stride length measurements found that under the barefoot condition, stride length was significantly longer than the other conditions ($p < 0.05$). However, no significant differences in stride length were found between high heel, high heel with insert, and toe walking. The current study defined stride length by measuring the distance between initial contact of the dominant foot to the subsequent initial contact of the same foot. There may be instances, however, when individuals walking in high heel make initial contact with their forefoot instead of their heel. This study did not mandate which part of the foot had to strike the ground at initial contact. Therefore, for some individuals, it may be likely that the toe walking and high heel conditions were, in fact, similar. Interestingly, a slight trend could be detected with toe walking exhibiting the smallest stride length, following in ascending order, high heel with insert, and high heel, leaving a partial agreement with the hypothesis of this study that the longest stride length would be observed under the barefoot condition, followed by, in descending order high heel with insert, high heel, and toe walking. These results agree with other studies which explored the effects of walking in high heel shoes (Esenyel, Walsh, Walden, & Gitter, 2003; Opila-Correia, 1990; Snow & Williams, 1994) and toe walking (Bovi, Rabuffetti, Mazzoleni & Ferrarin, 2011; Davids, Foti, Dabelstein & Bagley, 1999). However, Simonsen and colleagues (2012) reported no signficiant differences in stride length when walking at a fixed-speed over ground in either the barefoot or high heel condition. Results from this study implies that walking with the foot in the plantar flexed position, whether it be while walking in high heel shoes and toe walking, shortens the stride length an individual is willing to take during gait.

Change in participant height was examined with and without normalization. Significant differences were observed between barefoot and the high heel, high heel with insert, and toe walking conditions ($p < 0.05$), which precludes this study from declaring a continuum amongst the variables for this condition. These results do not agree with the hypothesis that the barefoot condition would exhibit the least change in participant height followed by toe walking, high heel with insert and high heel. Saunders and colleagues (1953) hypothesized that more energy was needed to accomplish a gait cycle if the sinusoidal curves produced during gait are larger. An increase in change of participant height would indicate a larger sinusoidal curve during the gait cycle and that more energy would be needed to complete this task (Saunders, Inman, & Eberhart, 1953). Previous studies reported changes in height from 1-3 cm (Tesio, Lanzi, & Detrembleur, 1998), to 2.5 to 8.5 cm (Gard & Childress, 2001), to 4.6 cm (Saunders, Inman, & Eberhart, 1953). Data from this study is above results reported by Tesio, Lanzi, and Detrembleur (1998) and within the results reported by Gard and Childress (2001). A limitation of comparing the data to Gard and Childress is that the gender of their participants was not revealed. This makes it difficult to definitively compare the two studies. Change in absolute height observed with this study may be an indicator to which footwear condition is more efficient for walking. Data were normalized in order to reduce inter-subject variability for future studies. This variable can help indicate less energy consumption in walking when smaller changes in height are observed during gait.

Summary: Part 2

Kinetic Variables

The effects of footwear condition on GRF in the anterior/posterior and vertical directions were examined. Specifically, braking and propulsion GRF were examined in the anterior/posterior direction and heel strike transient, foot loading, and propulsion GRF was also examined in the vertical direction. With regard to anterior GRF, significant differences were found between barefoot and high heel with insert ($p < 0.05$) and toe walking ($p < 0.05$) with the least amount of braking detected while barefoot, followed by high heel with insert, and toe walking. No significant differences were found between barefoot and high heel, high heel and high heel with insert, toe walking and high heel, or toe walking and high heel with insert. This indicates that the position of the foot in plantarflexion and the inability of it to engage in the amount of dorsiflexion observed while walking barefoot affects the magnitude of ground reaction forces detected during braking. These do not agree with the hypothesis that the least amount of braking would be observed with barefoot, followed by high heel, high heel with insert, and toe walking. In addition, these results concur with Snow and Williams (1994) who hypothesized that increased braking occurred in the high heel condition because the foot was unable to fully pronate because the shape of the shoe forced the foot to remain in plantarflexion during the stance phase of gait. Ebbeling, Hamill, and Crussemeyer (1994) also reported similar braking results and attributed this to the inability of the foot to engage in the amount of dorsiflexion observed while walking in low heeled shoes which hindered attenuating ground reaction forces via eccentric contraction of the dorsiflexors. Stefanyshyn and colleagues (2000), did see larger braking forces with shoes outfitted with heels up to 5.4 cm and smaller braking forces with shoes that had an 8.5 cm heel. Since Stefanyshyn and colleagues did not employ a shoe with a 7.62 cm heel, it may be inferred that the point at which defense mechanisms of the body engage involve heel heights that are larger than the heel height used in this study. Bovi and associates (2010) also found increased peak braking in the toe walking condition when compared to normal walking among adults. Thus suggesting that placing the foot in plantarflexion, through either the high heel or toe walking conditions, puts the body in a situation where it must accommodate the lack of dorsiflexion through larger braking forces in the anterior/posterior direction.

Footwear effects on the magnitude of propulsion while walking resulted in no statistically significant differences in the anterior/posterior direction. This is counter to the hypothesis that barefoot gait would produce the least amount of propulsive GRF, followed by toe walking, high heel with insert and high heel condition. Ebbeling and colleagues (1994) also reported no significant differences between heel height and propulsive force. Data from Bovi and associates (2010) concurs, finding no significant differences in propulsive force when comparing normal walking to toe walking. Lack of significant differences between high heel and high heel with insert indicates that the Insolia® insert had no effect on propulsive forces.

With regard to vertical GRF, Simon and colleagues (1981) reported a heel strike transient can occur in either a shod or unshod condition. So it is not surprising to find a heel strike transient in all but the toe walking condition. Statistically significant larger heel strike transients were found in the high heel and high heel with insert condition when compared to the barefoot condition ($p < 0.05$). This does not agree with the hypothesis of this study that barefoot condition would exhibit the least amount of transient force followed by high heel and high heel with insert conditions. However, no statistical differences were found between the high heel and high heel with insert conditions. Simon and colleagues (1981) reported increased heel strike transients when participants walked barefoot than when they walked while shod which are contrary to the results from this study. It should be noted, however, that the shod condition used in the study by Simon and associates (1981) did not involve high heel shoes. Results from the current study agree with the outcomes presented by Snow and Williams (1994) and Jefferson and associates (1990). Snow and Williams (1994) theorized that an increase in heel strike transient resulted from a perception of instability while walking in high heel shoes. Jefferson and associates (1990) believed larger heel strike transients occurred because the individual used the ground instead of muscle activity to slow down the limb just before initial contact. Thus it is speculated that walking in the high heel or high heel with insert condition did not attenuate heel strike transient and that the act of placing the foot in plantarflexion through that particular shod condition may have exacerbated heel strike transient forces detected in this study. Additionally, it is apparent that the use of the orthotic insert did not reduce heel strike transient.

The current study investigated the loading response during gait that occurs after heel strike transient and indicated by a peak in vertical GRF. This investigation found the barefoot loading response was smaller than the loading response in the high heel, high heel with insert, or toe walking conditions ($p < 0.05$) which did not agree with the hypothosis that the barefoot condition would exhibit the least amount of vertical GRF at foot loading followed by toe walking, high heel with insert and high heel conditions. No statistical differences were found amongst the other conditions. Data from the current study concur with findings reported by Ebbeling and colleagues (1994) and Stefanyshyn and associates (2000). However, it should be noted that there was a main difference in methodology between Ebbeling and colleagues (1994) and Stefanyshyn and associates (2000) as both observed different heel heights but not a barefoot condition. With regard to toe walking, Bovi and associates (2010) reported an increase in peak loading response in the toe walking when compared to normal walking. The current study differs from previous investigations of GRF at loading response in that barefoot and toe walking variables were explored along with a shod condition observed with and without an orthotic (high heel and high heel with insert). Results from the current study indicate that placing the foot in the plantar flexed position and the subsequent lack of attenuation through dorsiflexion increased GRF at loading response. The body appears to compensate for the high heel and high heel with insert condition by shortening stride length and increasing knee flexion at IC in the high heel and high heel with insert condition, leading the author to believe that the body was unwilling to place the lead foot further out in front of the trailing foot. In other words, the body was trying to keep the leading foot within the base of support while in the high heel, high heel with insert and toe walking conditions. Thus, the author expected to see the COGFC exhibiting the least horizontal distance in the toe walking condition, followed by high heel, high heel with insert, and barefoot. However, this study resulted with toe walking exhibiting the largest distance followed by high heel, high heel with insert, and finally barefoot ($p < 0.05$). The propulsion component of vertical GRF occurs near the end of stance phase, from terminal stance through pre-swing. Significantly smaller peak propulsive vertical GRF was found under the barefoot condition when compared to the toe walking condition, indicating that more force was needed to propel the individual forward while in the toe walking condition. No significant statistical differences were found between barefoot and high heel, or high heel with insert conditions. Also, no significant differences were detected between high heel and high heel with insert, or toe walking and high heel or toe walking and high heel with insert conditions. These results do not agree with hypothesis in that the barefoot condition would exhibit the least amount of propulsive vertical GRF followed by toe walking, high heel with insert, and high heel. This finding also partially agrees with studies performed by Ebbeling and colleagues (1994) and Stefanyshyn and associates (2000). Their findings reported similar results except Ebbeling and colleagues (1994) and Stefanyshyn and associates (2000) observed different heel heights and not a barefoot condition. Bovi and associates (2010) reported an outcome different to this study with no significant differences between normal walking and toe walking with regard to propulsive vertical GRF. Walking efficiently should involve a smooth trajectory with little displacement of the center of gravity in the vertical and lateral directions (Saunders, Inman, & Eberhart, 1953; Waters & Mulroy, 1999). The increase in vertical propulsive forces displayed with toe walking condition could result in an increased change in height, which was reported earlier in this study, and may lead to additional energy consumption. Energy consumption has been reported to rise with the increase in heel height. To that end, Ebbeling and associates (1994) reported increased energy consumption, in the form of elevated heart rate and increased consumption of oxygen and expiration of carbon dioxide, occurred when individuals walking in shoes with a 7.62 cm (3”) heel. Curran and colleagues (2010) performed another study measuring energy consumption while wearing a variety of heel heights (1.5 cm, 4.5 cm and 7.0 cm) with and without two different shoe orthotics and found that while walking in higher heeled shoes alone, heart rate and oxygen consumption also increased. Interestingly, Curran and associates (2010) reported that wearing the highest heel (7.0 cm) with an orthotic insert resulted in heart rate and oxygen consumption significantly lower than while walking in the high heel alone. If utilizing the above proposal that an increase in propulsive forces should result in a larger change in height, then the lack of differences between the barefoot, high heel, and high heel with insert conditions result in no height changes while walking in the aforementioned conditions; however, this did not occur. Instead, a larger change in height was reported in the high heel and high heel with insert condition when compared to barefoot. Therefore, the author believes that in the present study, walking in high heel, high heel with insert, or toe walking would require more energy than when walking barefoot.

Summary: Part 3

Muscle Activation

Footwear effects on mean and peak muscle activation of the gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, erector spinae, and rectus abdominis during the gait cycle were explored. Results from this examination did not agree with the hypotheses that mean and peak activity for gastrocnemius would be the least in high heel with insert, followed by high heel and toe walking conditions. However, both the high heel and high heel with insert conditions resulted in less mean and peak gastrocnemius activity than the toe walking condition ($p < 0.05$). Additionally, there were significant differences between toe walking and high heel conditions with regard to biceps femoris activity ($p < 0.05$). No significant differences were found between muscle activity in the high heel and high heel with insert conditions, indicating that the addition of the orthotic insert did not affect muscle activation with regard to walking. The increase in gastrocnemius activity in the high heel and high heel with insert conditions ($p < 0.05$) may be attributed to increased knee flexion and heel strike transient observed while walking in the aforementioned conditions. Mika and associates (2011) and Nascimento (2014) also found increased gastrocnemius activity while walking in the high heel condition. Increased mean and peak gastrocnemius activation reported in the toe walking condition was expected. Previous studies by Perry and colleagues (2003) reported this outcome and postulated the increase in EMG activity while toe walking was a mechanism employed to stabilize the ankle while in plantarflexion. Additionally, because the high heel, high heel with insert, and toe walking conditions caused the foot to be forced into plantar flexion, muscle fascicles may not in their optimal position for force production as would be observed in the barefoot condition. This action would may the actin-myosin cross-bridge formations to be in a less efficient configuration thus leading to the necessity for more muscle activity in order to complete the task (Cronin, Barrett, & Carty, 2012). Biceps femoris activity exhibited significant differences only between toe walking and high heel conditions ($p < 0.05$). No significant differences were displayed between high heel and high heel with insert, or toe walking and high heel with insert, indicating that the addition of the orthotic insert did not appreciably alter biceps femoris activity. In order to prepare the forward limb for weight acceptance, the biceps femoris may slow down hip flexion (Perry & Burnfield, 2010). An increase in biceps femoris activation as well as greater hip flexion at IC in the high heel condition, may be the result of the body attempting to compensate for the footwear condition. In regard to the toe walking condition, increased biceps femoris activity can be interpreted as compensation for increased work performed by the plantarflexors during gait (Kerrigan, Riley, Rogan, & Burke, 2000) as hip and knee flexion were less than that observed in the other conditions.

While mean and peak activation of the tibialis anterior, rectus femoris, erector spinae, and rectus abdominis did not reach statistical significance, the author believes that these results should be addressed. The tibialis anterior is responsible for controlling the descent of the foot to the ground (Gage et al., 1995; Perry & Burnfield, 2010) as well as mitigating vertical forces during gait (Ebbeling, Hamill & Crussemeyer, 1994). While no signficant differences were found between high heel, high heel with insert, and toe walking, it should be noted that tibialis anterior activity was larger than the baseline. It appears to the author that although the shod condition affords some support to the foot where toe walking does not, the high heel, high heel with insert and toe walking conditions did not mitigate vertical forces at IC. During gait, the rectus femoris works to avert excessive knee flexion. Although rectus femoris activity did not have significant differences between the high heel, high heel with insert, and toe walking conditions, the activity recorded was over 200% of that observed in the barefoot condition. With regard to the high heel and high heel with insert condition, the rise in rectus femoris activity could be explained by the increased knee flexion at IC and larger heel strike transient. For high heel, high heel with insert, and toe walking conditions, increased rectus femoris activity is most likely a mechanism to maintain stability while walking in plantar flexion. The rectus abdominis and erector spinae are considered the postural muscles of the body, where the rectus abdominis is responsible for trunk flexion and the erector spinae is responsible for keeping the trunk upright. Muscle activation for these two muscles were not as dramatic as the rectus femoris. However, it appeared that there was a trend where the shod conditions exhibited less rectus abdominis activation. This leads the author the speculate that the shod condition may ameliorate some of the increased muscle activity observed while toe walking. Very little difference was detected in erector spinae activity when comparing high heel, high heel with insert, and toe walking conditions which leads the author to believe that although the body is walking with a smaller base of support due to the foot being in plantar flexion, the shod condition does not appear to afford any support to keeping the trunk upright. A limitation to this study was that instead of normalizing muscle activation to maximum voluntary isometric contractions, muscles of interest were normalized to mean and peak muscle activity during barefoot gait (Yang & Winter, 1984; Burden, Trew, & Baltzopoulos, 2003).

Summary: Part 4

Pennation angle of the gastrocnemius

Significantly larger pennation angles were detected in the high heel, high heel with insert, and toe walking conditions when compared to barefoot while standing ($p < 0.05$). The toe walking pennation angle was also found to be significantly larger than high heel and high heel with insert while standing ($p < 0.05$). This does not agree with the hypothesis that after walking five minutes, the pennation angle of the gastrocnemius would be largest while standing in plantar flexion, as observed with toe walking, and decrease in the following order: standing in high heel, standing in high heel with insert, and standing barefoot. No significant differences were found between high heel and high heel with insert, indicating that the addition of the orthotic did not alter the position of the foot nor alter the neuromuscular demand placed on the gastrocnemius to the degree that the pennation angle of the gastrocnemius would be affected. It is interesting to note that while in the standing position, a continuum of pennation angle with barefoot exhibiting the smallest pennation angle, followed by high heel and high heel with insert, and then toe walking exhibiting the greatest pennation angle appears. An increased pennation angle was expected as the standing participants moved from barefoot, where the foot was completely on the ground, to high heel and high heel with insert, which involved wearing shoes with a 3” heel, and then toe walking, which involved standing on the forefoot only. In addition, these results are well within the range of gastrocnemius pennation angle change of ~20 to 45° which occur during low level voluntary contractions as reported by

Lieber and Fridén (2000). This indicates that a low level of gastrocnemius activity was necessary for standing in the high heel, high heel with insert and toe walking conditions. Further studies utilizing ultrasound during gait trials would help further clarify the relationship between foot loading, ankle position, and gastrocnemius pennation angle.

Overall, the Insolia® insert did not significantly alter kinematic, kinetic, electromyographic, or pennation angle variables observed while walking in high heel as no significant differences were detected between the high heel and high heel with insert conditions. However, a continuum was revealed in knee flexion where the toe walking condition displayed the least amount of knee flexion, followed by barefoot condition, and finally high heel and high heel with insert conditions. Hip flexion at midstance displayed a continuum with the barefoot condition demonstrating the least amount of hip flexion followed by toe walking, and finally high heel and high heel with insert conditions. Observation of COGFC indicated a continuum where the barefoot condition exhibited the smallest distance followed by high heel and high heel with insert conditions with similar results, and finally toe walking condition. Lastly, a continuum was revealed with the barefoot condition exhibiting the smallest gastrocnemius pennation angle followed by high heel and high heel with insert conditions, and finally toe walking condition displaying the largest pennation angle. The observance of the aforementioned continuums indicate that altering the position of the foot while walking through either high heel or toe walking, some kinematic variables and pennation angle of the gastrocnemius change when compared to barefoot gait. Toe walking appears to illicit more alterations in gait than when walking in high heel, thus, indicating that the toe walking condition may be more taxing to the body than the high heel condition. Additionally, the results from the current study can add to the body of knowledge for gait studies.

Future Research

Future Research for Footwear Effects

The present study was one of few that have explored the effect of high heel footwear (which included an orthotic claiming to change the kinematics of gait), and toe walking on gait. Significant differences were not found between the high heel and high heel with insert condition suggesting that the insert did not significantly alter the kinematics, kinetics, muscle activity and pennation angle. However, only one heel height was examined. The designer of the high heel with insert claims that the higher the heel the better the results from this insert (Dananberg & Trachtenberg, 2000). Another study found that wearing a shoe with a 7.0 cm heel with the insert used in this study did improve energy efficiency by observing heart rate, volume of oxygen consumed while, respiration exchange ratio, physiological cost index, and number of steps, as well as perceived comfort while walking when compared to walking in flat shoes (Curran, Holliday, & Watkeys, 2010). Future studies should investigate a variety of heel heights, starting at 3” and up to ascertain if the higher heel condition can bring kinematic and electromyographic variables, including timing of muscle activity, closer to that of those detected in the barefoot condition. Comparing pennation angles found significant differences between standing in the barefoot, high heel and high heel with insert ($p < 0.05$), barefoot and toe walking conditions ($p < 0.05$), as well as signficant differences between toe walking and high heel conditions ($p < 0.05$), and toe walking and high heel with insert conditions ($p < 0.05$). Future studies utilizing a device to place the foot in the same position as that observed with standing barefoot or toe walking may give a clearer comparison of what occurs with the gastrocnemius pennation angle when it is in an identical loaded and unloaded positions.

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Appendix A

The Influence of High Heel Shoes and Toe Walking on Gait and Kinematics and Kinetics

Participant Health Questionnaire

Participant code: ________________________ Date: __________

Date of Birth: _______________ Age: __________

	Yes	No
Have you suffered any injury to your lower limbs during the past 6 months?	___	___
Have you undergone any surgery to your lower limbs during the past 6 months?	___	___
Are you in good general health?	___	___
Are you pregnant?	___	___
Do you have a neuromuscular disease affecting mobility or gait?	___	___
Are you allergic to adhesives?	___	___

Appendix B

****DO NOT AGREE TO PARTICIPATE UNLESS AN IRB APPROVAL STAMP HAS BEEN APPLIED TO THIS DOCUMENT****

INFORMED CONSENT FOR

The Influence of High Heel Shoes on Gait

You are invited to participate in a research study of "The Influence of High Heel Shoes and Toe Walking on Gait and Kinematics and Kinetics." With your help it is hoped that we can explore the differences in muscle activity, pennation angle, and gait kinematics and kinetics between walking in high heel fashion shoes and high heel fashion shoes with an insert and toe walking. You were selected as a potential participant because you are between the ages of 19 and 45 years, and your current health status permit you to perform the test safely and successfully. General inclusion criteria include, but are not limited to: (a) no current injury or illnesses that will prevent you from walking in high heel shoes (b) no known allergy to adhesive (c) a level of comfort and/or familiarity with walking in high heel shoes and (d) the ability to wear the women's size shoe that will be provided for this study. Participants will be excluded if they affirm any of the following: (a) experience back or lower limb injury or surgery during the previous six months, (b) experience any injury that would jeopardize successfully performing required tasks, (c) allergies to adhesive or adhesive type products, (d) cannot fit into the shoes provided for the study (Ladies' US 7-9), 5) any known balance or inner ear maladies, (e) pregnancy (f) fear of wearing or walking in high heel shoes and/or (g) any neuromuscular disease affecting mobility or gait. The results of the study will be used to compare the muscle activity, pennation angle, and kinematic and kinetic effects of walking while barefoot, barefoot while toe walking, high heel fashion shoes, and high heel fashion shoes with an insert.

Purpose: The purpose of this study is to explore the differences in muscle activity, pennation angle, and gait kinematics and kinetics between barefoot walking, walking in high heel fashion shoes, high heel fashion shoes with an insert, and toe walking. Combined this will allow us to answer the following questions: (1) Does the muscle activity observed while walking barefoot, wearing high heel fashion shoes, wearing high heel fashion shoes with an insert, and toe walking, differ? (2) Does the kinematics of walking change when walking barefoot, wearing high heel fashion shoes, wearing high heel fashion shoes with an insert, and toe walking? (3) Does the calf exhibit different muscle morphology when standing barefoot, wearing high heel fashion shoes, wearing high heel fashion shoes with an insert , or standing barefoot on their toes? And (4) Are the kinetics of walking barefoot, walking in high heel fashion shoes, high heel fashion shoes with an insert, and toe walking different?

Methodology: A meeting will be arranged at the Auburn University Sports Biomechanics Laboratory. At the meeting anthropometric data will be collected, and participants will be familiarized with the protocol by practicing each condition. The protocol involves walking across a 30 foot walkwaywith an embedded force plate through the capture volume of the motion capture system that includes a 10 Vicon T-Series T40S cameras (Denver, Colorado, USA). Foot dominance will be determined by asking the participant to stand. While standing, the participant will be asked to point their toe towards a box and return their foot to the original position. The foot that the participant extends will be deemed their dominant foot. Retroreflective markers will be placed at various anatomical landmarks. These markers will be used to provide position data for the kinematic portion of the study. To prepare for the experiment the participant will first undergo careful skin preparation (shaving, abrasion, and cleaning with alcohol) and electrode placement over the muscle belly of various muscles commonly known as the thigh, calf, and lower back (rectus femoris, biceps femoris, tibialis anterior, medial gastrocnemius, rectus abdominus, and erector spinae) on the dominant side. Electromyography will show the timing and magnitude of muscle activity in the four different experimental conditions. After electrode placement is complete participants will perform a maximal voluntary contraction for the muscles which will be measured by electromyography. This allows the muscle activity during performance of the experimental conditions to be normalized so as to allow for comparison across people who might have different impedances. Next, the participant will be allowed to walk in the shoes for 5 minutes to become accustomed to the shoes. Ultrasonography will also be performed. An ultrasound probe will be positioned on the calf of the dominant leg to determine pennation angle while resting, during maximum voluntary contraction of the medial gastrocnemius, while standing in the footwear condition, and while standing immediately after completing the footwear condition. Retroreflective markers will also be attached over the pelvis, knees and feet so as to be detected by the Vicon-T Series T40S motion capture system.

Next the experimental protocol begins. Using a counter balanced design, the participants will be randomly assigned to an order of footwear conditions group. Three successful

Participant's Initials

trials of each condition will be performed. A successful trial will include striking the force platform with the dominant foot. Condition group A, will start by walking barefoot across the walkway. Condition group B involves walking barefoot on your toes across the walkway first. Condition group C requires you to walk in high heel fashion shoes across the walkway first. Condition group D begins by walking in high heel fashion shoes with an insert across the walkway. This session will last approximately 90 minutes. You should not participate on the day of testing if any illness or medication you are taking would make you less able to successfully walk in the shoes.

Risk: While participating in this study it is possible that you might experience irritation due to shaving the areas under the electrodes. There is also a possibility that you may incur a joint sprain, muscle strain, or injury if you should fall. Furthermore, these factors may lead to serious injury or death. However, injuries are unlikely due to care taken to warm-up and this task is one encountered during activities of daily living. In the unlikely event you experience an injury as a result of participation in this study, the investigators will summon help. However, the investigators have no plans to provide funds for any medical expenses or other costs you may incur. It should be noted that you will be responsible for any and all medical cost resulting from injury during or related to this study.

Benefit: There is no direct benefit to you other than the opportunity to gain a better understanding of human movement.

Participant's Initials

Confidentiality: Any information obtained in connection with this study that can be identified with you will remain confidential. Your decision whether or not to participate will not jeopardize your relation with Auburn University or the School of Kinesiology. If you decide later to withdraw from the study you may also withdraw any identifiable information, which has been collected about you in this study.

Contact/Questions: If you have any questions now or later about this study, please feel free to contact Lorraine Smallwood (lls0017@tigermail.auburn.edu) at 844-1468. Additionally, if you have questions about your rights as a research participant you may contact the IRB Chair at hsubjec@auburn.edu or IRBchair@auburn.edu or you may call them at 844-5966. You will be given a copy of this form to keep.

YOU ARE MAKING A DECISION WHETHER OR NOT TO PARTICIPATE. YOUR SIGNATURE INDICATES HAT YOU HAVE DECIDED TO PARTICIPATE HAVING READ THE INFORMATION PROVIDED ABOVE.

Participant’s Printed Name	Participant’s Signature	Date
Investigator Conducting Consent (Printed)	Investigator Conducting Consent (Signature)	Date