Influence of Various Thong Style Flip-flops on Gait Kinematics and Lower Leg Electromyography

by

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A dissertation submitted to the Graduate Faculty of
Auburn University
in partial fulfillment of the
requirements for the Degree of
Doctor of Philosophy

Auburn, Alabama
December 18, 2009

Keywords: flip-flops, gait, kinematics, footwear,
biomechanics, electromyography

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Abstract

Flip-flops are a common footwear choice for the masses. This influx of flip-flop usage is in despite of the plethora of anecdotal evidence and consensus among the health field that flip-flops are not conducive to the health of individuals' lower extremities. The influence of footwear research on gait measures can be seen in the ever changing design of running shoes and there is abundant research of the effects of orthotics on gait, but still a lack of research on the effects of flip-flops on gait exists. The purposes of this investigation were; (1) to examine the effects that different components of the thong style flip-flop have on gait kinematics in individuals classified with normal arched (NA) feet; (2) to investigate the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified with either low (LA), normal (NA), or high arched (HA) feet; and (3) to determine if there is an increase in muscular activity of the tibialis anterior (TA) at the ankle during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA.

The results show that flip-flops decrease stride length and peak eversion when compared to barefoot. In addition, a flip-flop with components such as an arch support, midtarsal support, toe ridge, and wider straps result in a gait which resembles a "normal" gait in college aged females with NA. The results also show that for LA, NA and HA individuals, a flip-flop with arch support resulted in a gait that resembled a "normal" gait

compared to a flip-flop without an arch support. Finally, the current study found that increased activity of the TA was observed without a subsequent increase in dorsiflexion of the ankle in two out of three flip-flop conditions. In conclusion, no flip-flop investigated was exactly like walking barefoot; however, certain structural components of flip-flops do result in a gait similar to walking barefoot. Future research is still needed to investigate and design a flip-flop that results in a gait identical to walking barefoot.

Acknowledgements

The author would like to thank his advisor, Dr. Wendi Weimar for the opportunity to study Biomechanics under her guidance. Over the years she has been an inspiration, a mentor, and is now a dear friend. Dr. Weimar's knowledge, guidance, understanding and wisdom have allowed me to mold and shape myself into a true professional in the field of Biomechanics.

The author would like to thank the members of his committee, Dr. Mary Rudisill, Dr. JoEllen Sefton, and Dr. Lloyd Wade, for their help and support through this process. The author would also like to thank Dr. James Witte for serving as the outside reader.

The author would like to thank the Department of Mechanical Engineering, specifically Dr. Nels Madsen and Dr. Ruel Overfelt, and Industrial and Systems Engineering, specifically Dr. Lloyd Wade, for the use of equipment which made this project possible.

The author would also like to thank his parents, Mr. and Mrs. Scott Shroyer and Mr. and Mrs. Richard Booker, as well as his grandparents, Mr. and Mrs. Joeddie Adkins, for the countless sacrifices and endless support they have graciously given over the years.

Most importantly, the author would like to thank his personal Lord and Savior, Jesus Christ, who makes all things possible.

Finally, the author would like to thank the love of his life, Joanna, for her support, encouragement, patience, and love through this wearisome journey.

Table of Contents

List of Tables

List of Figures

List of Abbreviations

OFM	Oxford Foot Model
BF	Barefoot
FF1	Flip-flop 1
FF2	Flip-flop 2
FF3	Flip-flop 3
SL	Stride length
EV	Eversion at midstance
EV_{PEAK}	Peak eversion during stance
PRO	Pronation at midstance
PRO_{PEAK}	Peak pronation during stance
$DORSI_{PEAK}$	Peak dorsiflexion during swing phase
HX_{PEAK}	Peak hallux extension during swing phase
sEMG	Surface electromyography
TA	Tibialis anterior

CHAPTER I. INTRODUCTION

There is much anecdotal evidence that flip-flops are not conducive to the health of individuals' lower legs and especially feet. Yet, when asked why people wear flip-flops comfort is the usual response. Several quotes by podiatrists support this anecdotal evidence. Dr. Rock Positano, a podiatrist at New York's Hospital for Special Surgery states that, "Flip-flops have singlehandedly caused more problems with people's feet in the last couple years than probably any other type of shoe" (abcNEWS, 2007). Podiatrist Dr. Greg Cohen, from Long Island College Hospital in Brooklyn, New York, states, "Flip-flops don't really hold on the foot like most shoes do, so we use the tendons and muscles to hold them on" (Yara, 2006). It seems to be that there is a consensus among podiatrist that flip-flops are not a healthy type of footwear. Podiatrist also seem to agree that by wearing flip-flops, people have to recruit more muscle fibers in the lower leg and foot to prevent the flip-flop from coming off the foot. This increased muscular activity, in an attempt to keep the flip-flop on, may cause or exacerbate foot muscular and skeletal problems. Although these are just anecdotal claims and there is no empirical evidence to support the podiatrist claims. Podiatrist are the experts on the foot, so is there any legitimacy to what they report? One would argue that just because they are experts on the foot, that not everything they say is correct. So where is the empirical evidence to support these claims by the experts?

Even though there is a health stigma associated with wearing flip-flops via the podiatrist, flip-flops are becoming a common footwear option for individuals, and according to the NPD Group in Port Washington, a provider of consumer and retail market research information, men's sports sandal sales in 2003 were up five percent from the previous year while overall footwear sales were down six percent (Wilson, 2004). In addition, the Surf Industry Manufactures Association (SIMA) in 2007 reported that one of the top surf industry trends was sandal sales. The SIMA stated that overall footwear sales were down, but sandal sales were up over $300 million, which was an increase of $50 million since 2004 (SIMA, 2007). Although thong flip-flops are a type of sandal and increase sales of sandals noted by SIMA does not necessarily mean an increase in the thong style, it is interesting to note that men's thong flip-flop sales in department stores had a fourfold increase from 2002 to 2006 as reported by the NPD Group in Port Washington (Dash, 2006).

Casual observation of individuals wearing thong flip-flops has indicated that: (1) individuals wear flip-flops beyond the structural limit of the flip-flop (i.e. the foot bed is worn out and there is no cushioning properties left in the EVA foam of the flip-flop), (2) flip-flops are designed and sold with a one size fits all mentality, and (3) individuals have a different gait while wearing flip-flops versus shoes. This observed altered gait may lead to compensation or unusual stresses that flip-flop wearers do not encounter while wearing a more traditional shoe such as an athletic sneaker. Though flip-flops are often worn for comfort, the excessive wearing of flip-flops has been linked to discomfort. In fact, according to the American College of Foot and Ankle Surgeons (ACFAS) in 2006, an increase in usage of flip-flop sandals by teens and young adults has led to an increase in heel pain; however, there are few studies to confirm these statements. These are again just anecdotal evidence that labels flip-flops as "bad" for the feet. ACFAS spokesperson Marybeth Crane, DPM, FACFAS stated, "We're seeing more heel pain more than ever in patients 15 to 25 years old" and heel pain is a marker of plantar fasciitis which accounts for 15 percent of all adult foot complaints. Furthermore, the ACFAS recommends that patients with heel pain should avoid flat shoes with paper-thin soles, and should also avoid walking barefoot since wearing flat shoes and walking barefoot provides little to no arch support. As a result, this lack of arch support and cushioning of the heel while wearing flip-flops seems to exacerbate any abnormalities in the biomechanics of foot motion, and may perpetuate heel pain and inflammation. These statements suggest that flip-flops are a contributor to heel pain and should not to be worn if heel pain is present (Surgeons, 2006). While the causal relationship between flip-flops and heel pain seems to be accepted clinically, the manner by which this is achieved is not. So, is there a basis upon which to evaluate and discern with empirical scientific evidence as to why the scientific community has already (anecdotally) labeled flip-flops as medically harmful?

If one looks to the motor learning literature as a place to start, then Karl Newell would encourage us to investigate the influence of the environment on the task and the individual. In 1984, Karl Newell introduced constraints to the motor development community. Newell noted three distinct constraints that include the individual, the task, and the environment. These constraints are factors that are put together in an interacting model to create a picture of what affects the development and specifically the movement of a person. Newell's interacting model shows that a researcher must consider not only the individual of interest, but also the environment (physical and socio-cultural) the individual is in, and the task (goals, rules, and equipment) that the individual is performing (Tomas, 1984). If the role of flip-flops is to be considered then the interactive nature of Newell's Model would suggest that the environment and task are inextricably linked, and as such would imply that footwear (specifically in this case thong style flip-flops) will have a direct impact on the movement of an individual. As a result, applying Newell's Model, the task of walking is inherently influenced by the footwear that the individual wears.

If one thinks about the task of walking in the context of Newell's Model, footwear is important in human movement. Footwear can be considered the equipment for walking, and it is the sole interface between the foot and the ground in shod activities. Because the foot is the first and sometimes only interaction with the ground during normal gait, it must therefore, play a key role in the regulation of normal walking gait patterns (Nurse & Nigg, 2001).

The influence of footwear research on gait measures can be seen in the ever changing design of running shoes. There are numerous brands and types of running shoes available; however there are three main categories: motion control shoes, stability shoes, and cushion shoes. The driving force for these three categories is the thought that when a person is matched with the appropriate shoe, injuries are reduced (Butler, Hamill, & Davis, 2007). The importance of matching the correct type of shoe to a runner's needs has been claimed to be even more important to individuals with pes cavus and pes planus. These individuals are typically more susceptible and experience more overuse injuries than those with "normal" arches (Kaufman, Brodine, Shaffer, Johnson, & Cullison, 1999). For example, one study assigned footwear based on an initial arch type screening at a military base. The proper assignment of footwear resulted in a 50% decrease in all lower extremity injuries reported (Knapik, Feltwell, Canham-Chervak, Arnold, & Hauret, 1999).

While the research supports that it is beneficial to wear the appropriate type of footwear based on an individual's foot type, there is also research that suggests footwear may be a contributor to the improper development of the foot during the early childhood years. In this case, it would seem that footwear is both the cause and effect, for footwear brings about changes in the foot and changes in the foot are accommodated by other footwear. Wolf and colleagues (2008) state that the primary function of shoes for adults and children is to protect the foot from injuries and from the environment. However, they suggest that optimum foot development can only occur in barefoot conditions as footwear has been labeled as the culprit for various foot deformities and symptoms. For example, valgus deviation has been shown to develop in toddlers soon after beginning to wear shoes. The detriment to foot health is not only seen in the earlier years, but it also persists into the adulthood. Due to excessive high-heel wearing, women tend to have a high occurrence of a variety of destructive foot and lower leg pathologies. Other possible health problems that result from footwear include plantar ulcerations, stress fractures, plantar fasciitis, heel spurs, and metatarsalgia (Burnfield, Few, Mohamed, & Perry, 2004; Morag & Cavanagh, 1999).

With the plethora of research on numerous types of footwear (athletic shoes, shoes, children's shoes, and high heels) showing that indeed, footwear does have a direct effect on the movement of the human body as Newell's Model suggests, there seems to be a dearth of research on other types of footwear; and in light of the prevalence of use of flip-flops, a serious lack of attention to the effects of thong style flip-flops. Sparse research that has been done to investigate certain types of the broader category of sandals (Hillstrom, Song, Kim, & Heilman, 2005; Kim, Hillstrom, Song, & Heilman, 2005; Song, Hillstrom, Kim, & Heilman, 2005), but the results of those studies applied to thong flip-flops is only speculative. The only article to date that has directly addressed the influence of flip-flops on gait investigated the effects of flip-flops on gait kinetic and kinematics compared to sneakers (J. F. Shroyer & Weimar, *In Press*).

In conclusion, there is substantial anecdotal evidence that wearing flip-flops is harmful to the foot; however, there has been an observable increase in the number of sandals and flip-flop sales which would indicate that more people are wearing them. According to Newell's Model, the individual, task, and environment are intertwined and ultimately affect each other. In the task of walking, footwear can be considered part of the environment and therefore must be considered if the movement is to be fully understood. Scientific empirical evidence has been collected on various types of footwear to show that footwear does affect the biomechanics of tasks such as walking and running. However, this research has almost exclusively been in the lucrative athletic footwear arena, with only limited research done on sandals. There have been two studies that investigated the effects of thong style flip-flops on gait kinetics, but none to our knowledge on the effects of thong flip-flops on gait kinematics and lower leg EMG during gait.

Purpose of the Study

Therefore, the purpose of this investigation was three-fold: (1) to examine the effects that different components of the thong style flip-flop have on gait kinematics in individuals classified with normal arched (NA) feet; (2) to investigate the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified with either low (LA), normal (NA), or high arched (HA) feet; and (3) to determine if there is an increase in muscular activity of the tibialis anterior (TA) of the ankle during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA.

Hypotheses

The null hypotheses for the present study are as listed:

H_{o1}: Walking in thong style flip-flops with or without features such as an arch support and a cupped heel will result in no significant difference in gait kinematics in individuals with a normal medial arch.

H_{o2}: The medial arch height of a person will result in no significant difference in gait kinematics when walking in thong style flip-flops with or without an arch support.

H_{o3}: There will be no significant difference in the muscular activity of the tibialis anterior in individuals with a normal medial arch when walking in thong style flip-flops compared to walking barefoot.

Limitations

The limitations for the present study are as listed:

1. Self reported health status will be used.
2. Not every type of thong style flip-flop will be evaluated.
3. The amount of flip-flop/sandal usage experience that each participant possesses will not be taken into consideration.

Delimitations

The delimitations for the present study are as listed:

1. Participants will be required to wear retroreflective markers and surface electrodes on the right side of body.
2. The flip-flops that will be used for this study will be new and will have never been worn before.
3. The participants will be asked to walk in the Auburn Sport Biomechanics Laboratory within the capture volume at a self selected pace.

Definition of Terms

Surface Electromyography

Surface electromyography (sEMG) is a technique for evaluating and recording physiologic properties of muscles at rest and while contracting. EMG is performed using an instrument called an electromyograph that produces a record called an electromyogram. An electromyograph represents the spatial and temporal summation of all motor unit action potentials in the proximity of the recording electrode. The summation of the motor unit potentials is indicative of the level of muscle activity.

Kinematics

A branch of classical mechanics that focuses on describing the motions of an object without considering the factors that cause or affect the motion.

Pes cavus

Pes cavus is a characterized by a high and rigid medial arch of the foot as well as hyperextension of the toes. Pes cavus foot does not flatten when load is applied.

Pes planus

Pes planus is characterized by a low arch medial arch of the foot causing the foot to abnormally flattened and spread out. Pes planus is often referred to as flat foot.

Sandals

Sandals are an open type of footwear like flip-flops; however, sandals consist of a sole held to the foot by straps passing over the medial and lateral sides of the foot and/or around the anterior and posterior of the ankle (Figure 1).

Figure 1. Sandal (Doran, 2004)

Thong flip-flops

Thong flip-flops are a flat, backless, usually rubber sandal consisting of a flat sole held loosely on the foot by a Y-shaped strap, like a thin thong, that passes between the hallux (big toe) and the second phalange and continues around both the medial and lateral sides of the foot. For the rest of this project, when flip-flops are mentioned, unless otherwise stated, it will be referring to thong style flip-flops (Figure 2).

Figure 2. Thong flip-flop (Seper, 2008)

CHAPTER II. REVIEW OF LITERATURE

Flips flops are a popular footwear option, but have been linked to altered gait kinematics (J. F. Shroyer & Weimar, *In Press*) and anecdotally to lower extremity discomfort. This study will build upon the preliminary study by Shroyer & Weimar (*In Press*), and look to the kinematic changes that may lead to the lower extremity discomfort. Therefore, the purpose of this investigation was three-fold: (1) to examine the effects that different components of the thong style flip-flop have on gait kinematics in individuals classified with normal arched (NA) feet; (2) to investigate the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified with either low (LA), normal (NA), or high arched (HA) feet; and (3) to determine if there is an increase in muscular activity of the tibialis anterior (TA) of the ankle during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA.

This chapter will be divided into five sections. Section one will describe human locomotion and the phases of gait. Section two will describe the anatomy of the foot, ankle, and lower leg. Section three will describe the arches of the foot and the roles of the arch in human locomotion. Section four will describe the influence that footwear has on human locomotion as well as the motion of the foot. Finally, section five will summarize the pertinent findings of previous literature as it pertains to the present project.

Section 1: Human Locomotion and the Phases of Gait

For the purposes of this paper, when the word “gait” is used, it refers to normal forward bipedal locomotion of a human unless otherwise specified. As a person walks, one of the lower limbs serves as a mobile support while the other lower limb moves towards a new support site that is in front of the current support site. As the gait cycle continues the roles of each of the lower limbs alternates. A single sequence of one lower limb going from being the support leg to the free leg, back to the support leg is called a gait cycle. To fully investigate the human gait cycle, one can evaluate gait from three different approaches. The first approach is to divide the cyclic motion into segments based on ground contact between the two feet. A second approach evaluates gait by using time and distance measurements of a stride. The third approach is to separate the different phases of gait by the function or purpose of the segment (J. Perry, 1992). This section will provide an overview of the three approaches to investigating gait, as well as provide the phases and functions of each phase of gait.

The basis of the first approach to investigating gait is to break the gait cycle down based on the contact point of the reciprocal foot motion. Because walking is a cyclic activity and each part transitions to the next part, there is no true beginning or end to the gait cycle, so gait evaluation can begin at any point of the motion. Traditionally, however, gait analyses are begun at initial ground contact of one of the limbs. In normal gait, the heel is the first part of the foot to strike the ground; therefore, heel contact is often marked as the start of the gait cycle; however during some abnormal walking patterns the heel may not be the first part of the foot to strike the ground. As a result a more general term, initial contact, is used instead of heel contact (J. Perry, 1992). Each gait cycle can be divided into two basic time periods, the stance phase and the swing phase. The stance phase for a particular limb is the time period in which the foot is in contact with the ground, beginning with initial contact and ending just before toe off. The swing phase for a particular limb is the time period in which the foot is not in contact with the ground. Swing phase begins with toe off and terminates at the instance just before initial contact (Figure 3).

In the second approach to investigating gait, the gait cycle's stance and swing phases are further broken down into three distinct intervals with regard to the sequencing of the contralateral feet. The three intervals of stance phase include initial double stance phase, single limb support, and terminal double stance phase. The gait cycle is said to begin during initial double stance phase when both feet are in contact with the ground and specifically at initial contact of the new lead foot. When the opposite foot (current rear foot) is lifted off the ground at toe off, single limb support (of the current lead foot) begins. At this point, one leg is in contact with the ground (the current lead foot) while the other leg (rear foot) swings through to be place back on the ground (and becomes the new lead foot). When the swing leg makes initial contact, the single limb (for the current lead foot) support phase is over and the terminal double stance (for the current lead foot) phase begins. The terminal double stance phase lasts until the original stance limb is lifted off the ground at toe off (J. Perry, 1992) (Figure 3). It is interesting to note that terminal double stance phase for one leg is initial double stance phase for the opposite leg.

As a crude estimate, the gait cycle is comprised of 60% stance time and 40% swing time for each leg. One sixth of the stance time is spent in initial double stance which accounts for 10% of the total gait cycle. Two thirds of the stance time is spent in single limb support, which accounts for 40% of the total gait cycle. The final one sixth of the stance time is spent in the terminal double stance phase, which accounts for the remaining 10% of the total gait cycle (Murray, Drought, & Kory, 1964) (Table 1).

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

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

The 60/40 split of stance versus swing is a crude estimate and the more precise measurements vary with individuals and walking velocity (Mann, 1982). Mann (1982) states that at a velocity of 80 m/min the stance phase was 62% of the gait cycle and the swing phase was 38% of the gait cycle. The author further noted that the time of the stance and swing phases of the gait cycle had an inverse relationship to the walking velocity in that both the stance and swing phases decreased in duration as velocity increased. Also, as walking velocity increases, the percentage in single stance increases as the percentage in the two double stance phases decreases. The reciprocal is true as velocity decreases. Both responses when graphed are curvilinear (Mann, 1982). Therefore as velocity increases, one trades double stance for single stance and then ultimately double stance is eliminated. At the point that double stance is eliminated, the person has progressed into a run (J. Perry, 1992).

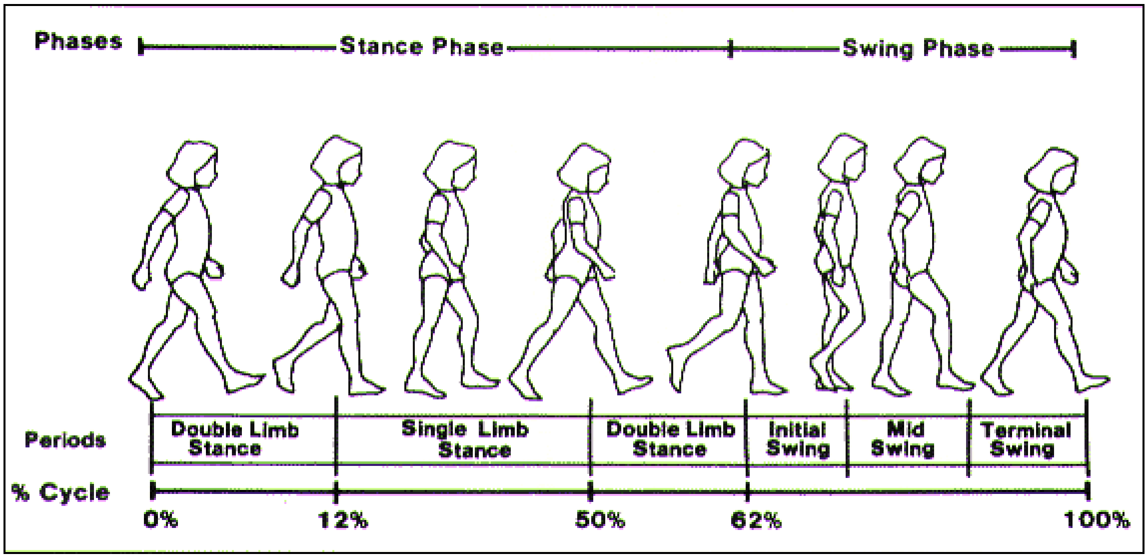


Figure 3. Phases of gait (Kirtley, 2008)

In locomotion there is another descriptive term called a stride. The term step is often used in place of stride; however, a stride and step are two separate things. A stride is analogous to a gait cycle. A stride is the time period from initial contact of one foot to initial contact of the ipsilateral or same foot. For example a stride would be the time period from initial contact of the right foot until initial contact of the right foot again (Kirtley, 2006; J. Perry, 1992). A step however refers to the time period from initial contact of one foot until the initial contact of the contralateral foot. A step is half a gait cycle; therefore two consecutive steps equals one gait cycle (or one stride) (J. Perry, 1992).

Finally, the third approach to investigating gait is the identification of the functional purposes of each phase of gait. This approach incorporates the first two approaches but ultimately gives the functional role of each component as the subdivision of the gait cycle. The first approach can describe gait as a sequence of two basic phases, swing and stance. Furthermore, using the second approach the swing and stance phases can be further divided into initial/terminal double limb stance, single limb stance, and swing phases. The gait cycle can also be broken down into functional phases based on the objective and pattern of the movement. For example, walking has a functional purpose and therefore needs to be analyzed and identified based on these specific and critical functions. A gait cycle can be divided into three basic tasks and further divided into eight gait phases (Figure 4). The three basic tasks include weight acceptance, single limb support, and limb advancement. The first task, weight acceptance, is the beginning of the gait cycle and includes initial contact and loading response. Single limb support is next, and includes mid-stance, terminal stance and pre-swing. The gait cycle ends with limb advancement which includes pre-swing, initial swing, mid swing, and terminal swing. The reason the pre-swing is part of both the single limb support task and the limb advancement task is that during the stance phase the positioning the foot is important for the initial limb advancement task; therefore, there is a crossover from the single limb support task to the limb advancement task. The pre-swing is an important part of the limb advancement, occurring during the stance phase, the pre-swing phase is included in both the single limb support and limb advancement tasks (J. Perry, 1992) (Figure 5)

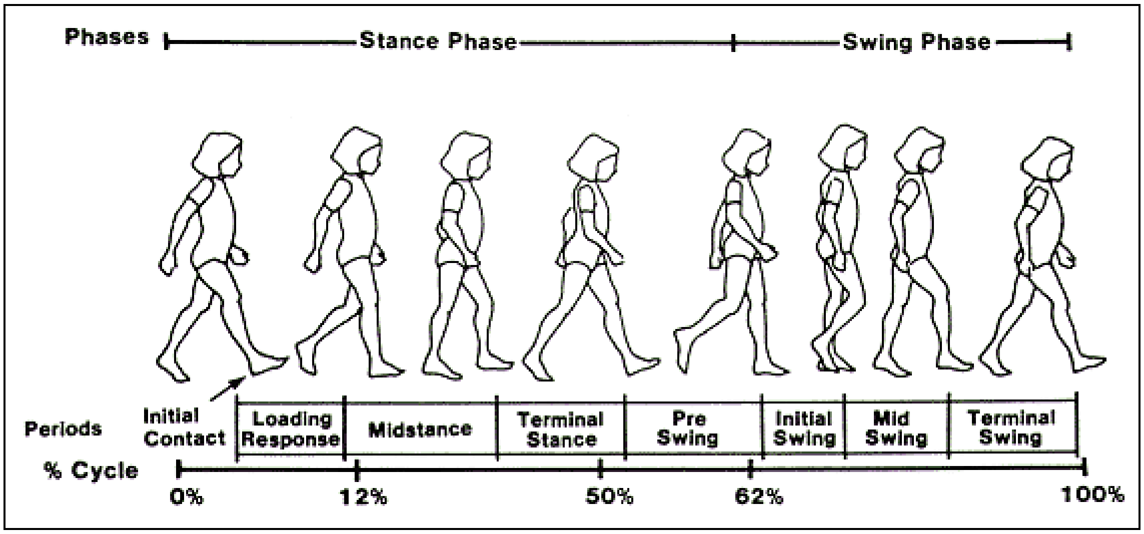


Figure 4. Functional phases of gait (Kirtley, 2008)

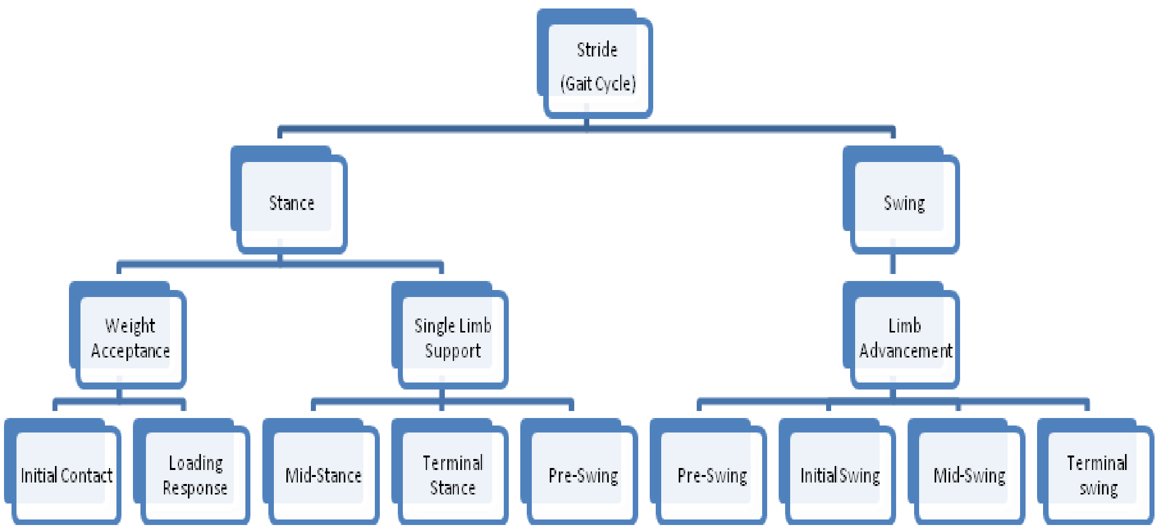


Figure 5. Gait cycle divisions (J. Perry, 1992)

The weight acceptance task consists of the first two functional phases of the gait cycle (initial contact and loading response). The purpose of the weight acceptance task is to transfer the body weight to a limb that was in swing phase. The two phases in the weight acceptance task are the initial contact and loading response, and these phases often overlap. The first phase, initial contact, accounts for 0-2% of the interval of the gait cycle and represents the instant the swing foot makes contact with the ground. The main objective of the initial contact is to position the loading limb for the start the stance phase with a functional pivot point that allows the progression of the body over the supporting foot, called a heel rocker. The second phase, the loading response, accounts for the 0-10% interval of the gait cycle and is the initial double stance period. The loading response phase's main objectives are shock absorption, weight-bearing stability, and progression of the body. This phase begins with the initial contact and ends when the contralateral limb enters swing phase (J. Perry, 1992).

The single limb support task consists of the third, fourth, and fifth functional phases of the gait cycle (mid-stance, terminal stance & pre-swing). The purpose of the single limb support task is to support the weight of the translating mass of the body onto a single limb. The third phase of the gait cycle is the mid-stance phase which accounts for the 10-30% interval of the gait cycle. The mid-stance phase is also the first half of the single limb support and ends when the center of mass of the person is directly over the forefoot. The main objectives of mid-stance are the translation of the center of mass over a fixed foot as well as limb and trunk stability. The fourth phase of the gait cycle is terminal stance. Terminal stance accounts for the 30-50% interval of the gait cycle and the main objective is the translation of the center of mass outside the body's base of support. Terminal stance is the end of single limb support (Figure 4) (J. Perry, 1992).

The limb advancement task consists of the last four functional phases. The fifth phase, the pre-swing, is the terminal double stance phase and is the final part of the stance phase as well as the initial part of the limb advancement task. The pre-swing phase accounts for the 50-60% interval of the gait cycle and is the time period from initial contact of the opposite limb to the toe-off of the ipsilateral foot. The main objective of pre-swing is to position the limb for the swing phase. The sixth phase of the gait cycle is the initial swing phase. This phase accounts for the 60-73% interval of the gait cycle and is one-third of the entire swing period. The initial swing phase is the time from when the foot leaves the ground until it is across from the stance foot. The main objectives of the initial swing phase are foot clearance of the ground and advancement of the foot forward. The seventh phase of the gait cycle is the mid-swing. The mid-swing accounts for the 73%-87% interval of the gait cycle and begins with the swing limb opposite the stance limb, terminating when the swinging limb is forward of the stance leg, and the tibia of the stance leg is perpendicular to the ground. The objective of the mid-swing phase is advancement of the limb and foot clearance. The eighth and final phase of the gait cycle is the terminal swing. The terminal swing begins with the end of the mid-swing and continues until the foot makes contact with the ground, which is where the weight acceptance phase begins again. The main objectives of the terminal swing include complete limb advancement and preparation of the limb for the stance phase (J. Perry, 1992) (Figure 4).

If locomotion is considered to include more than just the lower body, then the human body can be divided functionally into two sections in regards to locomotion: the passenger unit and the locomotor unit. The passenger unit can be considered the head, neck, trunk, and arms and is sometimes referred to with the acronym HAT (Elftman, 1954). The HAT comprises a large proportion of the total mass of the human body, specifically it makes up approximately 67.8% of the total body mass (Miller & Nelson, 1973). The passenger unit is called the passenger unit because some researchers believe that during normal locomotion, the upper body is somewhat independent of the locomotor unit and as such, does not aid in walking locomotion and basically just rides on top of the locomotor unit (Ford, Wagenaar, & Newell, 2007; J. Perry, 1992; Umberger, 2008). Any muscle action of the passenger unit is to maintain neutral vertebral alignment and to minimize any postural changes. The arms are ambulatory during normal gait; however, they serve no principle function in the locomotion (J. Perry, 1992). In a study that looked at the effects of constraining one arm during walking found that there were decreases in transverse pelvic, thoracic, and trunk rotation with an increase in the non-constrained contralateral arm movement amplitude; however, there were no differences in frequency and phase relations between the arm and leg (Ford et al., 2007). A separate study by Umberger (2008) also looked at the effect of suppressing arm swing on walking. Umberger found that gait kinematics, kinetics and energetics were not very different between people that walked with or without arm swing. Energetics were 10% higher for people that did have arm swing as well as some joint torques; however, joint angles, angular velocities, and ground reaction forces were not different (Umberger, 2008). These two studies by Ford (2007) and Umberger (2008) suggest that indeed the head, trunk, neck, and arms are passengers in locomotion. The position and balance of the passenger unit or HAT is highly dependent on the locomotor unit to move the body's base of support underneath the HAT's center of mass (J. Perry, 1992).

The locomotor unit consists of the pelvis and the right and left lower limbs. It acts as the transporter for the passenger unit. The locomotor unit is a highly complex system with numerous muscles, joints, and bones, and it is the coordination of all these muscles, joints, and bones that make normal locomotion possible. In normal walking, the forward fall of the body's center of mass is the primary propelling force (J. Perry, 1992). The mobility of the distal portion of the limb at the foot is imperative to control this fall and is manipulated to ensure that momentum is preserved. There are three functional rockers at the foot that aid in locomotion: the heel, ankle, and forefoot rocker. These rockers allow the center of mass to pass over the base of support while allowing the knee to remain in a relatively extended position (J. Perry, 1992). The contralateral limb during the swing phase provides another pulling force to maintain forward progression of the body's center of mass; moreover, the contralateral limb provides a segment to catch the falling center of mass (J. Perry, 1992). This repetitive cycle is what makes horizontal translation possible. The joints of the locomotor unit rotate to control the fall of the center of mass in order to translate the body forward.

As indicated there are three basic approaches to interpreting and analyzing gait, the present project will be concerned with the limb advancement task and the single limb support task portions of the gait cycle introduced in the third approach to gait analysis. It is these points that will provide the clearest evidence regarding the questions of this research project. Specifically, this project will focus on: (1) stride length from initial contact of the right foot to initial contact of the right foot again, (2) mid-/hindfoot supination and hindfoot/tibia eversion during the single leg support task, and (3) sEMG activity of the tibialis anterior muscle in conjunction with the joint actions of hallux flexion and tibio-talar flexion during limb advancement. Previous research has indicated that the HAT does not contribute directly to locomotion (Elftman, 1954; Ford et al., 2007; Umberger, 2008). While this concept is open for debate, there is no research to countermand this assertion and as a result, the present project will not include the HAT or any component of the HAT in the analysis of gait.

Section 2: Anatomy of the Foot and Lower Leg

Bones of the Lower Leg and Foot

While the author concedes that the pelvis and upper legs contribute to locomotion, the focus of the present project will focus on the behavior of the lower leg and foot; as such, this section will include the anatomical description of only these components. The lower leg is comprised of two bones, the tibia and the fibula (Figure 6). The foot is a multifaceted system of the body that is comprised of 28 bones and 24 joints. The bones of the foot include the tarsus, metatarsus, and the phalanges (Figure 7). The foot can further be divided into three parts, the hind or rear foot, the midfoot, and the front or forefoot. The hind foot and midfoot are made up of the tarsus and contain seven bones called the tarsals. The tarsals include the talus, calcaneus, navicular, cuboid, medial cuneiform, intermediate cuneiform, and lateral cuneiform. The forefoot is comprised of the metatarsals and phalanges of the foot and these are analogous to the metacarpals and phalanges of the hand. There are five metatarsals and they are numbered one through five with number one being the most medial metatarsal and number five being the most lateral metatarsal. Also often found in the forefoot are two sesamoid bones on the distal plantar surface of the 1^{st} metatarsal. These sesamoids are called the tibial and fibular sesamoids and are positioned to behave as a class one pulley, improving the line of action of the hallux flexors. There are 14 phalanges in total. Each digit has three phalanges, except the hallux which only has two phalanges. The digits of the foot are numbered the same as the metatarsals in that the most medial digit is number one and the most lateral digit is number five. The phalanges of each digit are named proximal, middle, and distal for their anatomical position (Marieb & Mallatt, 2003) (Figure 7).

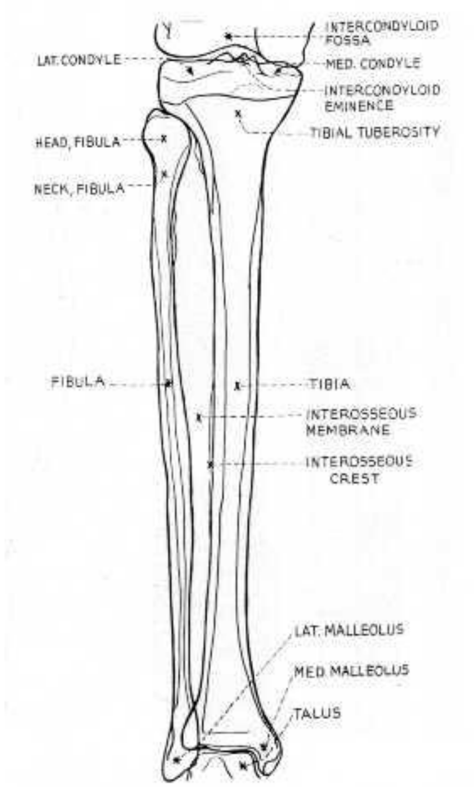


Figure 6. Bones of the lower leg anterior view (Oldnall, 2009).

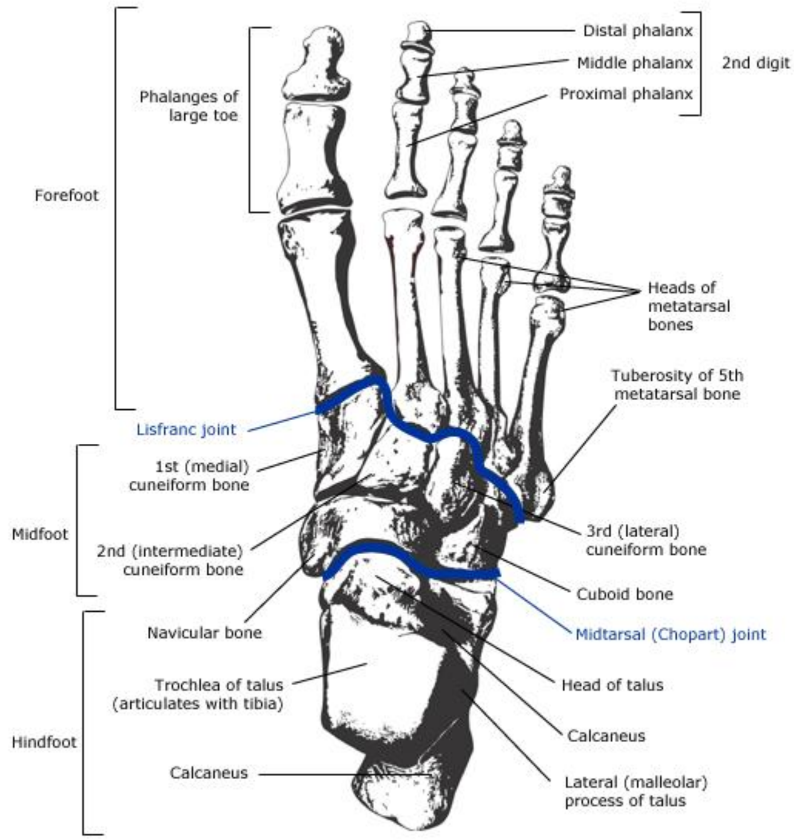


Figure 7. Bones of the foot superior view and the transverse tarsal joint shown as midtarsal joint in figure ("Bones of the foot," 2009).

Joints of the Lower Leg and Foot

The distal articulation between the tibia and fibula makes up the distal tibiofibular joint; in conjunction with the talus of the foot create the talocrural joint, or the mortis for the ankle joint (Figure 7). The talocrural joint is just one of 24 joints of the ankle and foot complex. The other joints included the proximal tibiofibular joint, distal tibiofibular joint, talocalcaneal joint, talonavicular joint, calcaneocuboid joint, five tarsometatarsal joints, five metatarsophalangeal joints, and nine interphalangeal joints. Dorsi and plantar flexion occur at the talocrural joint, while inversion, and eversion (and a component of pronation, supination) of the foot occur at the subtalar joint, also known as the talocalcaneal joint (Figure 8). The transverse tarsal or midtarsal joint is a compound joint formed by the talonavicular joint and the calcaneocuboid joint; this is also where supination and pronation of the foot occurs. Separation of the hind foot from the midfoot occurs at the transverse tarsal joint, the articulation between the calcaneus and cuboid on the lateral side of the foot and the articulation between the talus and navicular on the medial side of the foot. The metatarsophalangeal and interphalangeal joints are where flexion and extension of the digits occur.

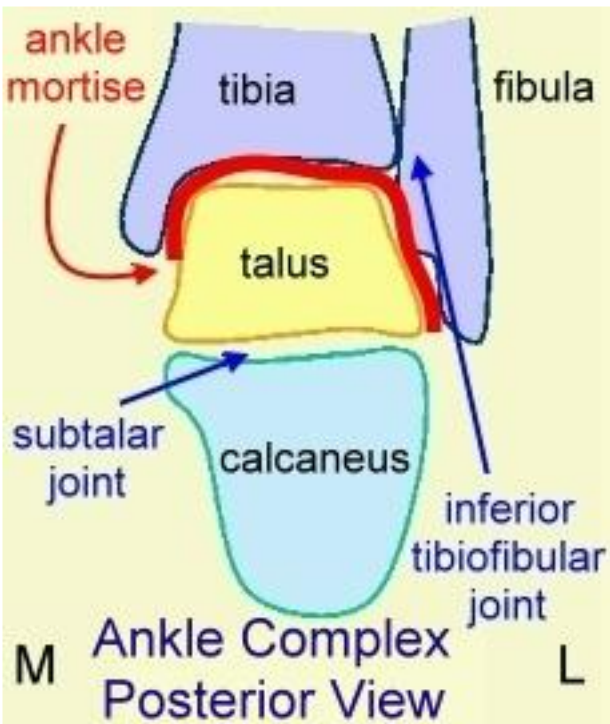


Figure 8. Talocrural and subtalar joint posterior view (Huei-Ming, 2004).

Musculature of the Lower Leg and Foot

The lower leg is divided into three distinct compartments: the anterior, lateral, and posterior compartments. The musculature of the lower leg is responsible for motions at the ankle, intertarsal joints, and the phalanges. At the ankle, the muscles of the lower leg contribute to plantar and dorsi flexion. At the intertarsal joints, the muscles of the lower leg contribute to inversion and eversion; and at the toes, the muscles of the lower leg cause flexion and extension. There are similarities between the musculature of the forearm and the lower leg. The muscles that are housed in the anterior compartment are analogous to the extensor muscle group of the forearm (also located on the anterior component of the forearm and contribute to wrist and finger extension) in that the muscles of the anterior compartment dorsi flex the foot and extend the toes. The muscles of the anterior compartment include the tibialis anterior, extensor digitorum longus, peroneus tertius, and extensor hallucis longus. The lateral compartment houses the fibularis or peroneal muscles. The peroneal muscles are responsible for eversion and contribute to plantar flexion. The muscles of the lateral compartment of the lower leg include the peroneus longus and peroneus brevis. The muscles of the posterior compartment are analogous to the flexor muscle group of the forearm (also located on the posterior component of the forearm and contribute to wrist and finger flexion). The posterior compartment muscles are responsible for plantar flexion of the ankle and/or flexion of the phalanges. The muscles of the posterior compartment include the gastrocnemius, soleus, plataris, popliteus, flexor digitorum longus, flexor halluci longus and tibialis posterior (Marieb & Mallatt, 2003).

Table 2
Muscle actions at the ankle and phalanges PM = prime mover (Marieb & Mallatt, 2003)

	Actions at the Ankle Joint				Actions at the Toes	
	Plantar Flexion	Dorsiflexion	Inversion	Eversion	Flexion	Extension
Tibialis Anterior		X (PM)	X			
Extensor Digitorum Longus		X				X (PM)
Peroneus Tertius		X		X		
Extensor Hallicus Longus		X	X (weak)			X (hallux)
Peroneus Longus and Brevis	X			X		
Gastocnemius	X (PM)					
Soleus	X (PM)					
Pantaris	X					
Flexor Digitorum Longis	X		X		X (PM)	
Flexor Hallicus Longus	X		X		X (hallux)	
Tibialis Posterior	X		X (PM)			

The intrinsic muscles of the foot are responsible for flexion, extension, abduction, and adduction of the phalanges as well as support for the arches in the foot. There is one muscle on the dorsal aspect of the foot and nine muscles on the plantar aspect of the foot. The single muscle on the dorsal side is the extensor digitorum brevis. The nine muscles of the plantar aspect are divided into four layers. The first layer being the most superficial and the fourth layer the deepest. The first layer houses the flexor digitorum brevis, abductor hallicus, and abductor digiti minimi. The second layer consists of the flexor accessories and the lumbricals. The third layer consists of the flexor hallicus brevis, adductor hallicus, and flexor digiti minimi brevis. The fourth layer consists of the plantar and interossei. The actions of the muscles of the lower leg and foot are summarized in Table 2 (Marieb & Mallatt, 2003).

Investigating all the joints and the muscles of the lower leg and the foot is beyond the scope of this project, the focus of this project will be specifically the talocrural joint, the subtalar joint, the transverse tarsal joint. The talocrural joint is of importance to this study because this is where plantar and dorsiflexion occur. The subtalar and transverse tarsal joints is where supination and pronation occur. Studying the motions of these joints will help to provide an understanding of the motion of the foot during the gait cycle while wearing various flip-flops. As with the joints, this project will not investigate all muscles of the leg. The muscle of interest for this particular study will be the tibialis anterior (TA) because of its primary role as a dorsiflexor. This will provide insight as to possible increased muscular activity in the TA due to the principle of reciprocal inhibition of the toe extensors during the swing phase of gait while wearing flip-flops because of the potential increase in toe flexor activity to maintain foot contact with the flip-flop.

Section 3: The Foot Arches

The foot is a segmented structure comprised of numerous bones and joints. In mechanics, the only way a segmented structure can support weight is if it is arched, and the foot is no exception with three arches. The arches of the foot form a half dome that distributes forces experienced at the foot to the heel and metatarsal bones. The three arches of the foot are the medial and lateral longitudinal arches and the transverse arch (Figure 9). The medial longitudinal arch is generally the highest of the three arches and is on the medial side of the foot. The medial longitudinal arch originates at the calcaneus, goes to the talus and then terminates at the three medial metatarsals. The lateral longitudinal arch is generally lower than the other arches and originates at the calcaneus, rises to the cuboid and then descends to the fifth metatarsal. The medial and longitudinal arches serve as anchors for the transverse arch. The transverse arch runs from the medial longitudinal arch to the lateral longitudinal arch along the joints formed by the tarsals and metatarsals (Marieb & Mallatt, 2003). As already mentioned, the role of the arches is to support the weight of the body. The arches of the foot have two extremes of anatomical structural position, pes cavus and pes planus. Although there are three separate arches within the foot, the medial longitudinal arch (MLA) has been found to have the greatest clinical significance in both pes cavus and pes planus. An individual with pes cavus has a high MLA as show in Figure 10. An individual with pes planus has a low MLA as show in Figure 10 (Franco, 1987). The role and function of the arches in gait will be discussed later. The sole purpose of this section is to describe the anatomy of the three foot arches. Even though all three arches play an important role in the locomotion of the human, for the purposes of this study only the medial longitudinal arch will be considered as it will be the arch most affected by the presence or lack of presence of a medial arch support in the footwear.

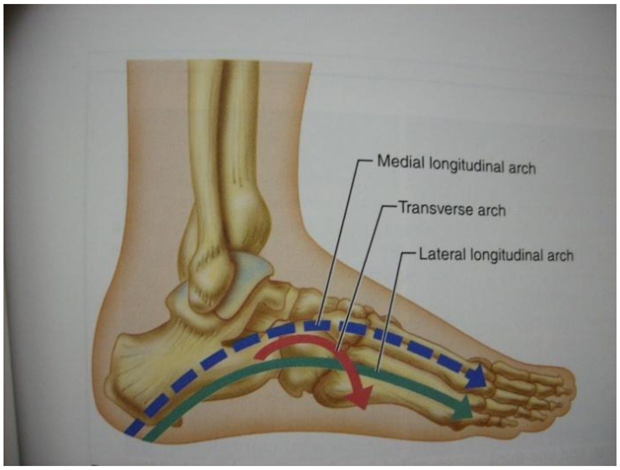


Figure 9. Arches of the foot (Marieb & Mallatt, 2003).

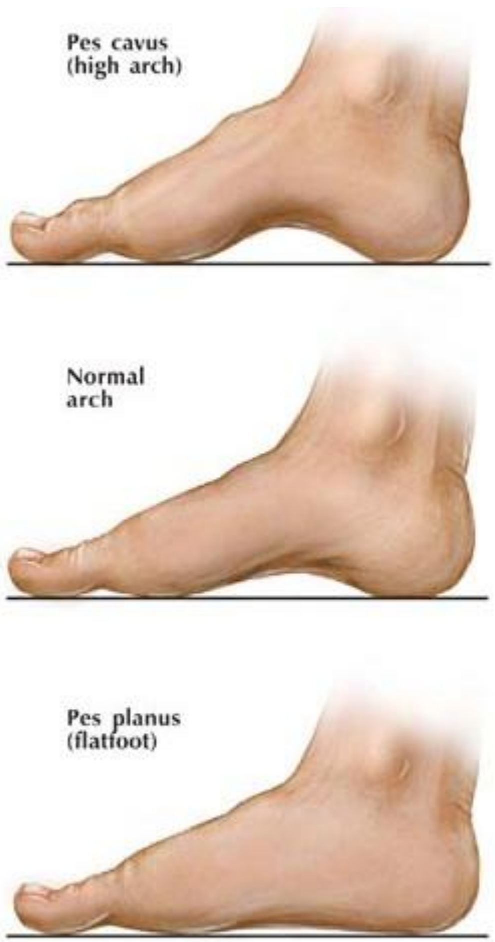


Figure 10. Foot arch types (Horwitz, 1999).

Section 4: Influence of Footwear on Gait

Footwear design

Because footwear is a main point of interaction of the human body with the ground when humans walk or run upright, footwear plays a major role in the protection of the foot itself. One type of activity in which protection of the foot is imperative is running. Some advantages of athletic footwear include protection of the plantar surface, improved traction, motion control during activity, and attenuation of impact forces (McPoil, 2000). Impact forces on the foot can be quite substantial, with research noting that these forces can be as high as one and a half times a person's body weight while walking, and even as high as eight three times a person's body weight while running (Nilsson & Thorstensson, 1989). This increase in forces during movement is the reason for the cushioning of footwear. To combat these forces, athletic shoes are equipped with a cushioned midsole and a cupped heel that is comprised of various foam materials such as Ethylene Vinyl Acetate (EVA) or polyurethane (PU). Athletic shoes may also have a gas, fluid, or gel, component to provide increased cushioning. The midsole of the shoe acts as a protective layer between the foot and ground and attenuates the forces of impact as well as reduces the magnitude of peak plantar pressures by distributing the forces acting on the foot over a larger surface area (McPoil, 2000; Orendurff et al., 2008). A separate outer sole layer of abrasion-resistant rubber compound is used to combat excessive wear of the foam midsole (McPoil, 2000), and the stiffness of the outsole is the largest factor in determining lateral forefoot cushioning (or protection of the fifth toe) (Orendurff et al., 2008).

In addition to providing cushion, a shoe may act to control foot motion such as pronation. Shoes accomplish motion control by providing heel stabilization and midfoot support (in the form of arch support). The degree of motion control provided by the shoe is dependent on the density of the midsole and most importantly the fit of the foot within the shoe and its snugness around the foot (McPoil, 2000). The properties of the shoe that relate to motion control are most crucial for field and court sports where athletic moves such as cutting and turning are prevalent. As the amount of side-to-side movement increases, instability of the forefoot increases (in the upper portion of the shoe). Due to this increase in movement, the material in the upper portion of the shoe can become over-stretched, leading to excessive shear and compressive forces specifically to the medial, lateral, and plantar surfaces of the forefoot (McPoil, 2000). Another part of the shoe that is of importance is the heel area. The heel experiences more than double the peak pressure during athletic moves such as cutting, jumping, and landing compared with running straight (Orendurff et al., 2008). Therefore, for improved performance and reduced risk of injury, shoes should be designed to provide both cushioning and support during athletic movements such as cutting (Orendurff et al., 2008). This is why the heel cup is critical to shoe design. With the increased pressures associated with athletic activities at the heel, the concave design of the heel cup allows for increased surface area to be in contact with the foot and ultimately aids in the dissipation of the forces experienced at the heel over a larger surface area.

The pertinent components of shoes that this project is going to investigate are the varus wedge and the heel cup. By investigating whether or not the same is true for the thong-style flip-flop, this project will attempt to build on the studies that have already shown decreased foot motion in footwear that has a varus wedge. The heel cup is also of interest because in the flip-flop there is no rear foot contact on the medial, posterior, or lateral side of the heel, as there is within shoes that completely surround and encompass the foot

Running and Training Shoes

Much of the research available on the influence of footwear on foot mechanics has been done with activities such as running. With the repetitive nature of the activity and the increased loading of the foot, the potential for overuse injuries of the lower extremities is increased and as a result has drawn attention in the research world. One component of interest is the stiffness of the midsole, since the midsole has been found to be important in shoe design as a method of limiting foot motion.   Specifically, running shoes that have stiffer midsoles have been shown to reduce foot eversion (De Wit, De Clercq, & Lenoir, 1995). De Wit and colleagues studied seven trained male long-distance runners, running with two different harness shoes, the softer Asker C40 and the harder Asker C65. The harder soled shoe resulted in smaller initial peak vertical impact and, and more rearfoot eversion at initial contact. On the other hand, the soft midsole shoes demonstrated an increase in eversion and pronation during mid-stance (De Wit et al., 1995). Although this study did not take foot type into consideration, these results imply that harder midsole shoes are better for runners with foot types that have increased motion, such as individuals with pes planus, while the softer midsoled shoes would be more suited for runners with foot types that have decreased shock absorption capacities, such as individuals with pes cavus.

The importance of matching the correct type of shoe to specific types of runners has been claimed to be even more important to individuals with pes cavus and pes planus. These individuals are typically more susceptible and experience more overuse injuries than those with “normal” arches (Kaufman et al., 1999). Kaufman and colleagues (1999) evaluated the foot structure of 449 trainees at the Naval Special Warfare Training center in Coronado, California. Prior to the training, Kaufman’s team took measurements of the trainees which included arch height, plantar/dorsiflexion range of motion (ROM), subtalar inversion, and subtalar eversion. Post training, they found that trainees with pes cavus or pes planus were twice as likely to get a stress fracture as individuals with an average arch height. Kaufman et al. (1999) also identified restricted ankle dorsiflexion, and increased hindfoot inversion as potential risk factors for lower leg and foot overuse injuries. Although this study did not look at preventing injuries by implementing an intervention, this study it important in showing that foot types predispose individuals to injuries. If it can be shown that certain gait mechanics can be altered by flip-flops, then it may be possible to reduce injuries that result from flip-flop usage.

Another study focusing on the arches of the foot and the importance of proper footwear was performed at Fort Drum, New York (Knapik et al., 1999). At Fort Drum, when incoming recruits came for training, arches were assessed by a physical therapist. Recruits with pes cavus were assigned a cushioned trainer shoe with a more compliant and softer midsole. Recruits with pes planus were assigned a motion control shoe with a harder midsole and varus wedge. Recruits with a normal arch were assigned a stability shoe. A stability shoe (which is a hybrid of the motion control shoe and the cushioning shoe) offers moderate motion control and moderate cushioning. The proper assignment of footwear based on arch type resulted in a 50% decrease in all lower extremity injuries reported (Knapik et al., 1999). This research was a critical study showing that lower extremities resulting from an overall physical conditioning program can be reduced just by proper footwear usage based solely on arch height.

Two more recent studies have looked at the influence of foot arch type and footwear on running mechanics (Butler, Davis, & Hamill, 2006; Butler et al., 2007). In the first study, Butler and colleagues studied 40 recreational runners. Twenty of which had pes planus and the other 20 had pes planus. Running kinematics and kinetics were collected as the participants ran at 3.5 ms^{-1} along a 25 m runway under two shoe conditions. The two shoe conditions were a motion control shoe (New Balance 1122 – which incorporates a stiffer midsole and varus wedge) and a cushion shoe (New Balance 1022 – which incorporates a softer and more compliant midsole). The results of the study indicated that motion control shoes resulted in decreased rear foot motion for both groups. Also, the cushioning shoes resulted in increased shock attenuation for both groups; however, there were no interactions between arch heights on any variable except instantaneous loading rate (Butler et al., 2006). In conclusion, this research shows that a person's foot, whether it be pes cavus or pes planus, does not behave differently in running shoes designed for either motion control or cushioning. The study did show that arch height does have an effect on loading rate across the two shoe types, which would suggest that a person should chose an appropriate running shoe (motion control or cushioning) based on their arch type in order to attenuate forces and the rate at which those forces are applied. Individuals with pes planus would benefit from a cushioning shoe due to the rigidity and poor shock absorption in the pes cavus foot, the cushioning shoe provides a shock absorbing material to protect the structures of the lower extremity from increased ground reaction forces. One major limitation to this study was that the running protocol was a self-selected pace for only a distance of 25 meters. This is not comparable to what runners experience during prolonged runs. It is anticipated that had these researchers used a more appropriate running distance that muscle fatigue would result in increase foot motion in the cushioning shoe and less foot motion in the motion control shoe for people with high arches. If this supposition holds true, then this would lend more credence to the premise that people with low arches would be better served by a shoe designed with a harder midsole and varus wedge.

In 2007, Butler, et al. did a follow up study that evaluated the influence of foot arch height on a prolonged run under the same two footwear condition as his 2006 study. The researchers hypothesized that increased motion would be experience during the prolonged run instead of the previous protocol of only 25 meters. Kinematic data were collected on 24 recreational runners, 12 of which had pes planus and 12 that had pes cavus. The participants ran on a treadmill for 30-45 min at a self selected pace. The running shoe conditions were a motion control shoe (New Balance 1122 – which incorporates a stiffer midsole and varus wedge) and a cushion shoe (New Balance 1022 – which incorporates a softer and more compliant midsole).   The results of the study indicated that the cushion shoe resulted in decrease tibial shock for the pes cavus group. Peak tibial internal rotation increased over the time of the run in the cushioning shoes whereas, it was decreased in the motion control shoe for the pes planus group. Also, there was no change in peak tibial rotation for the pes cavus group (Butler et al., 2007). Tibial rotation is the “twisting” of the tibia along its long axis in the transverse plane.

Increase tibial rotation is thought to be detrimental to runners in that it may increase susceptibility to overuse injuries at the location of where the tibia articulates with other bones, such as the femur at the knee and the talus and fibula at the ankle. This follow up study sheds more light on the importance of footwear selection based on foot type. This study shows that motion control shoes are beneficial to individuals with pes planus in that the motion control shoe decreases excessive foot motion; thus people with pes planus should look to purchase all shoes, not just running shoes, with a stiffer midsole and varus wedge. This study also suggests that people with pes planus experience less impact forces on the lower extremities in cushioning shoes; therefore, if individuals with pes cavus wore shoes designed for cushioning they could decrease potential overuse injuries. Another beneficial outcome of this study is that in the clinical setting, if mechanical analysis of a person's gait is not feasible or attainable, then clinicians can make footwear recommendations based on foot type alone.

Orthotics

The following studies have investigated the effects of altering and changing the architecture of footwear on locomotion. Generally, the main goal of altering footwear is to limit eversion and ground reaction forces (Mundermann, Nigg, Humble, & Stefanyshyn, 2003). Specific shoe design can limit eversion and attenuate ground reaction forces by changing the plantar pressures on specific areas of the foot. There are three ways in which the footwear can be altered: by an orthotic, posting, or a combination of both. A foot orthotic is typically a neutral shell that is fabricated by molding a polypropylene shell to a positive mold of an individual's foot. Posting is achieved by adding material to the medial or lateral aspect of foot area to the shoe directly or in the form of an orthotic (Mundermann et al., 2003).

Researchers suggest that the body adapts to changing plantar pressures on the foot by changing the motion of the foot, brought about by altering muscle activity in the lower leg. Therefore, if one could fabricate foot orthoses that modified the footwear in a way to reduce muscle activity, muscle fatigue and overuse injuries could be negated (Nigg, 2001). The midsole and the heel are the two primary locations for orthotics or orthotic like adaptations to be applied, moreover differences in the makeup of the midsole has been found to influence running mechanics in running shoes as previously mentioned (Butler et al., 2006; Butler et al., 2007; De Wit et al., 1995; McPoil, 2000).

In addition, midsole wedging has also been found to change gait kinetics (S. D. Perry & Lafortune, 1995). Perry and Lafortune (1995) investigated the effect of pronation on impact forces using a 10° varus wedge and a 10° varus wedge while ten individuals walked at a self-selected pace along an 18 m runway. The wedges were uniform thickness for the posterior one-third of the shoe, and then tapered to the metatarsal. The wedges were placed on the top of the footwear midsole. The valgus wedge was designed to exacerbate the eversion, and the varus wedge was designed to oppose eversion. This study also employed a control condition which utilized a midsole with no wedge. The result of the study showed that the varus wedge decrease rear foot eversion; however, it also increased tibial shock, impact peak loading rate, and vertical loading rate when compared to the valgus wedge and the midsole with no wedge. The results indicated decreased motion with midsole wedging, but with decreased motion there is an increase in the forces experienced by the foot (S. D. Perry & Lafortune, 1995).

Unfortunately, arch height was not considered in this project and as a result it is difficult to apply these results to individuals with foot arch pathologies such as pes planus and pes cavus. However, since the varus wedge did limit foot motion it is hypothesized that the varus wedge would be best for individuals with pes cavus and the valgus wedge would be better for individuals with pes planus.

Other research on the effects of orthotic and posting of shoes has found that building up or posting of foot orthoses reduces foot eversion, and the application of molded foot orthoses lowers impact and maximum vertical loading by as much as twenty percent (Mundermann et al., 2003).   Twenty one recreational runners (9 male and 12 female) ran on a treadmill at a velocity of 4 ms^{-1} in Bryce canyon running sandals by Rockport. This sandal is characterized by a rubber sole with adjustable Velcro straps that go over the dorsal part and the heel of the foot to secure the sandal to the foot. Inserts of the running sandals were removed and 4 conditions created: a control, a medial post, a neutral molding, and a custom molding with posting. The researchers noted that isolated, posting and molding have different effects on gait kinematic and kinetics. When molding and posting were combined, the effects of the molding were more influential than posting. The study concluded with the statement that if maximal foot eversion is a primary risk factor in running injuries, then runners should use posting; however, if increase vertical loading rate is a larger contributor to running injuries, then runners should use molded foot orthotics based on the individual’s foot (Mundermann et al., 2003).

Extreme shoe types

Modification of footwear has also been used in special populations such as with people with diabetes. Rocker bottom shoes (Figure 11) are commonly used in patients with diabetic neuropathy. Specifically, a study investigated the effects of rocker bottom shoes when compared to a normal commercially available shoe on foot plantar pressures. In-shoe pressure distribution was recorded in two conditions: a conventional extra-depth shoe and the same shoe which was modified into a rocker bottom configuration with a 24 degree rocker. The results of the study were a reduction in peak pressures in the medial forefoot, central forefoot, and toe region by 30% in the rocker shoe. However, heel, midfoot, and lateral forefoot pressures were elevated overall. This research shows that shoe modification can alter plantar pressures (Schaff & Cavanagh, 1990). The problem is that by decreasing pressures in one area, they are subsequently increased in another area. The good news is that under careful and individual design, shoe modifications may be beneficial to individuals with diabetic foot ulcers, or any other condition in which pressure redistribution is needed.

Figure 11. Rocker bottom shoe (Nigg, Hintzen, & Ferber, 2006).

Another study investigated the effect of rocker shoes, rocker height and rocker axis location on plantar pressures. A study by Van Schie and colleagues (2000) investigated the effects of nine different rigid rocker shoe designs to a control condition, which was a flexible, non-rockered extra-depth shoe with the same flat insole as the rocker shoes. All nine rocker shoes and the control shoe were worn by 17 healthy male subjects. Overall, peak pressure was reduced at the forefoot and increased in the midfoot and heel in the rocker shoe conditions. In addition, the rocker axis location was shown to have an effect on hallux pressures. Evaluation of these pressures yielded the best location of the rocker axis for reducing metatarsal head (MTH) pressure in the region of 55-60% of shoe length. However, for reducing pressure in the toe region the optimal rocker axis position was 65% of the shoe length. There was no single rocker shoe that provided optimal plantar pressure distribution for all subjects; however, the study showed that any of the rocker shoes was better than the control shoe in reducing peak plantar pressures in the forefoot (van Schie et al., 2000).

These two studies (Schaff & Cavanagh, 1990; van Schie et al., 2000) are another illustration of how footwear can influence gait kinetics. Further research needs to be done to see the influence of rockers shoes on gait kinematics, but it is apparent that through alterations or varying footwear, variations in gait parameters can be observed and that footwear can be individualized to benefit certain populations with foot abnormalities or pathologies.

One specific feature of footwear that has been research heavily is the heel. Numerous studies have investigated the effects of heel height on gait kinematic and kinetics (Hong, Lee, Chen, Pei, & Wu, 2005; Nyska, McCabe, Linge, & Klenerman, 1996; Snow & Williams, 1994). One study on the effects of high heels saw increases in vertical and anteroposterior forces during walking and less ankle abduction with increase heel height. Other differences in gait kinematic included increased plantarflexion during the entire gait cycle and decreased maximum knee ankle during the swing phase. Also during the swing phase, knee extension velocity was seen to be less with increased heel height (Snow & Williams, 1994).

A study done in 1996 investigated the effects of high heel shoes on plantar pressures (Nyska et al., 1996). This study suggests that when women wear high heeled shoes, plantar pressures are increased on the forefoot, which caused a decrease of plantar pressures on the hindfoot. Plantar pressures were also increased on the medial forefoot and the hallux.

With the increased plantar pressures that is associated with wearing high heels, one study by looked at the effectiveness of inserts on the high heel comfort as well as gait kinetics, which may be affect from these increased pressures observed when people wear high heeled shoes. This research study showed that there was a correlation of discomfort and heel height with the participants noting more discomfort in the higher heel. The introduction of the heel moved peak plantar pressures from the heel and midfoot to the medial part of the foot. In addition, vertical and anteroposterior ground reaction forces increased in the high heel conditions (Hong et al., 2005).

The research on heel height indicates that heel height increases ground reaction forces and relocates peak plantar pressures to different areas of the foot, specifically the medial forefoot and hallux. The significance is that the concentration of forces at these locations may acerbate symptoms in patients with foot deformities such as hallux valgus, bunions, or other foot complications.

To summarize, there are many people searching for new ways to modify shoes either for comfort, weight loss, treatment of systemic issues, or for beauty. One type of shoe types that has been receiving much attention is the rocker shoes and the ways rocker shoes are able to redistribute where pressures are placed on the plantar surface of the show. Lastly, the high heel is a shoe type that is very popular among the females. Its popularity in the professional workplace has led to various research projects investigating their effects on gait. Overall with these extreme shoe types it is apparent that different shoe types have various effects on gait mechanics. Even though there has been a focus on other types of popular footwear, there is no substantial research on the common and popular flip-flop.

Barefoot versus Shod

The study of the influence of footwear on gait kinematics has received relatively large attention in the current literature; however, since a comprehensive look at all literature regarding kinematics and kinetics is beyond the scope of this project only those studies that address the comparison between shod and barefoot conditions will be presented. This area of research was chosen for review because one of the anecdotal claims is that walking in flip-flops is like walking barefoot.

In a study investigating the gait of children, it was found that shoes cause a change in loading patterns during the gait cycle of children when compared to barefoot walking. For the experimental protocol, three shoe conditions and a barefoot condition were used as the 30 children walked across two Kistler force platforms. The barefoot condition resulted in an increase in ground contact duration, a shifting of maximal load from the rear towards the midfoot area, and an increased maximal load. The researchers concluded that shoes do affect loading patterns when compared to barefoot and that there is also variations between shoe types (Kristen, Kastner, Holzreiter, Wagner, & Engel, 1998).

On the contrary, another study reported that shoes have only a small impact on gait kinetics and kinematics in able-bodied children when compared to barefoot walking. This study included 14 children (8 females and 6 males) that were 7 to 10 years of age. Kinematic and kinetic data as collected as the children walked at a self-selected place under two conditions, barefoot and shod. The shod condition was a pair of low cut athletic shoes with arch supports and rubber soles. The measures that differed in shoes when compared to barefoot included decreased external foot rotation, decreased knee flexion from initial contact to mid stance, decreased plantar flexion, and increased stride length. The researchers stated that while some of the gait kinematic and kinetic differences were statistically significant they were not clinically significant; and that walking barefoot could be used clinically and for research instead of using shoes because of the ease of getting data without having to worry about covering the foot with a shoe or having different types of shoes (Oeffinger et al., 1999). Overall, it seems the research indicates that there is a difference between shod and barefoot walking even in young walkers; however, the clinical significance of these differences is still in debate.

Another study that investigated the influence that footwear has on the kinematic behavior of children's feet was based on the assumption that barefoot walking represents the best condition for the development of a healthy foot. To test this, scientists collected gait kinematics of 18 children walking under multiple testing situations. The first situation had the children walking in two conditions: barefoot and a commercial shoe. In the second situation, the children walked in three conditions: barefoot, commercial shoe, and an experimental shoe. The experimental shoe was designed to represent walking barefoot and was slimmer and more flexible than the commercial shoe. The results of the study showed that shoes do influence the motion of the foot. Specifically, tibio-talar range of motion (ROM) was increased in the commercial shoe when compared to the barefoot condition, and ROM in foot torsion along the long axis of the foot was reduced. The results of the study also showed that the experimental shoe was more similar to walking barefoot than a commercially available shoe (Wolf et al., 2008). The implications of this study indicate that if barefoot walking is the optimal condition for the healthy development of the foot, then footwear can be modified to be more like walking barefoot and still retain the added protection that shoes provide.

Further studies have been done to evaluated shod and unshod running gait kinetics and kinematics. Nine trained long distance male runners that ran at three different velocities in two footwear conditions: barefoot and shod, were investigated. The running velocities were 3.5, 4.5, and 5.5 ms^{-1}. The results indicated that the barefoot conditions yielded higher loading rates and a more horizontal foot position during impact than in the shod condition. During the swing phase, the barefoot condition resulted in a larger plantar flexion and a larger knee angle, presumably to adopt the previously mention foot placement; specifically to position the foot to allow for flatter foot placement. According to the researchers, this flatter foot placement was preferable because it decreased the pressure on the heel area and presumably would mean smaller forces passed up along the kinetic chain. The researchers also observed a shorter step length and larger step frequency in the barefoot condition and attributed that to “touch down geometry,” or otherwise stated, how the individual positioned the foot for initial contact (De Wit, De Clercq, & Aerts, 2000). However, another study proposed that the decrease in stride length observed when people wear slippers when compared to a heavier shoe was the direct correlation of the decreased mass of the slipper causing an decreased inertia of the distal segment (the foot) during the swing phase of gait (Mundermann et al., 2003). This finer point will be investigated in the present study, by the comparison of stride lengths between the flip-flops and barefoot, as the flip-flops will provide additional mass to the distal segment.

Running barefoot compared to running shod has also been investigated in anoter study measuring the kinematics of 8 males who ran on a treadmill under three footwear conditions: hard shoe, soft shoe and barefoot. The kinematic measures of interest were the ankle and knee angle in the sagittal plane. The results showed no difference in the ankle or knee angle for the two shoe conditions; however, they did find significant differences between barefoot and both hard and soft shoe conditions with the barefoot condition resulting in larger angles at both the ankle and knee joints during the stance phase of running. It was concluded that there is a larger variability in ankle and knee joint patterns while running barefoot than while wearing hard or soft shoes. This variability in joint kinematics may be an attempt by the body to vary the mechanism of the stresses place on the body during a repetitive motion such as running, and by varying the loading on the body, overuse injuries may be minimized (Kurz & Stergiou, 2003).

Sandals

While there is a scarcity of research on flip-flops and sandals and the affect on gait mechanics, this is not the case on the effects on foot development. Most of the available literature on flip-flops and sandals pertains to the influence they have on the development of the foot and not the influence on gait mechanics. There are numerous studies that suggest footwear affects the development of the foot (Echarri & Forriol, 2003; Gould, Moreland, Alvarez, Trevino, & Fenwick, 1989; Kusumoto, 1990; Oeffinger et al., 1999; Rao & Joseph, 1992; Sachithanandam & Joseph, 1995; Staheli, 1991). Footwear is important for the protection of the foot from the environment, but during early development, shoes can be harmful to the normal development of the foot. For instance, decreased incidences of pes planus (flat feet) are found in unshod individuals when compared to individuals of a shod population, and this occurs in both adults and children (Echarri & Forriol, 2003; Kusumoto, 1990). In fact, one study has shown that when comparing close-toe shoes, sandals and barefoot, flat feet is most prevalent in children who wear closed-toed shoes and least prevalent in children who go barefoot (Rao & Joseph, 1992).

Studies have shown that it is not just barefoot conditions that provide for an optimal foot development environment, but open-toe footwear such as flip-flops or sandals may provide similar benefits. Open-toed shoes such as flip-flops and sandals are better for the development of the arch of the foot than close-toed shoes. One possible explanation could be the increased intrinsic muscle activity to prevent the flip-flops or sandals from falling off. Open-toe shoes are not typically tied down to the foot and there is less contact between the foot and the footwear. As a result during movements such as the swing phase, the sandal or flip-flop has a propensity to separate from the foot. In doing so people may compensate to keep the footwear on their foot by using the digits to attempt to grip onto the surface of the footwear. To cause the digits to function, muscles must be used; therefore, open-toed shoes that are not firmly affixed to the foot may result in increased muscular activity. It is this increased muscular activity that may aid in the development of higher arches (Rao & Joseph, 1992).

Cultures, such as those in India, where primary footwear consists of slippers and sandals, normally engage in play while barefoot because of the ease of taking off and putting on of flip-flops and slippers. Researchers suggest that this increased play while unshod may explain the lower prevalence of flat foot in these children (Rao & Joseph, 1992). To investigate their claim, these researchers performed a study in which they evaluated the footprints of 2300 children (1237 boys and 1063 girls) between the ages of 4 and 13 years of age. Footprints via a differential pressure footprint mat were used to classify the foot type of each child: high arch, normal arch, low arch. This study showed a high concentration of flat foot among six year old children who wore shoes as compared to those who did not, suggesting that before the age of 6 is the critical period for the proper development of the foot arch. The researchers also found that close-toed shoes inhibit the development of the foot arch more than sandals or slippers. They also stated that children should be encouraged to play unshod and that slippers and sandals are actually less harmful than close-toed shoes (Rao & Joseph, 1992)

The findings in the previous study (Rao & Joseph, 1992) are supported by a follow up study done in which the prevalence of flat feet based on when individuals began wearing shoes was investigated (Sachithanandam & Joseph, 1995). The study consisted of static foot prints using a differential pressure footprint map of 1846 individuals of at least 16 years of age. The results showed that flat feet was highest in individuals who wore footwear for over eight hours each day; however, there were other factors besides footwear that resulted in a higher prevalence in flat feet. Two of these factors include obesity and ligament laxity. The researchers then adjusted for obesity and ligament laxity, and still found that higher rates of flat feet were seen in adults who began wearing shoes before the age of six (Sachithanandam & Joseph, 1995). The findings of these two studies imply that the critical age for development of the arch is before the age of six (Rao & Joseph, 1992; Sachithanandam & Joseph, 1995), and that shoe choice, and shoe versus barefoot play is critical for the healthy development of foot architecture.

Recently, scientists have begun to investigate not just the developmental effects of flip-flops and sandals, but also the effects this type of footwear has on gait kinetic and kinematics. Several studies at the Gait Study Center at Temple University have investigated the effects of sandal arch height on kinetics and kinematics (Hillstrom et al., 2005; Kim et al., 2005; Song et al., 2005). One of the studies investigated five different models of sandals with different arch heights: Santa Cruz (4.0 cm), Iceland (4.2 cm), Arizona (soft footbed, 4.3 cm), Arizona (hard footbed, 4.4 cm), and Fulda (4.6 cm). The participants were 20 individuals with a mean age of 27 years with moderate pes planus. It was observed that sandal arch height did have an effect on kinetic and kinematic variables in individuals with pes planus. These variables included ankle dorsiflexion, ankle eversion, walking velocity, stance time, and step length. It was also found that sandals with an increased arch height resulted in faster walking speeds and larger peak ankle adductory moments (Hillstrom et al., 2005).

A separate study from the Gait Center at Temple University found that postural sway was influenced by sandal arch height (Kim et al., 2005). As with the study by Hillstrom et al. (2005), this study investigated the same five sandal conditions on 20 individuals with moderate pes planus. The five sandal conditions with different arch heights were the Santa Cruz (4.0 cm), Iceland (4.2 cm), Arizona (soft footbed, 4.3 cm), Arizona (hard footbed, 4.4 cm), and Fulda (4.6 cm). Center of pressure (COP) was collected for each participant under each footwear condition using a Kistler force platform. The results suggested that there were differences in the sway velocities between sandal conditions, the authors suggested that there could be an optimal sandal arch height for individuals with moderate pes planus. This study also included a 2 month follow up for the Arizona sandal and found that there was a trend towards a decrease in sway velocity; however it was not statistically significant (p = 0.1546) (Kim et al., 2005).

In addition, Song and colleagues (2005), at the Gait Center at Temple University performed a study on the effect of arch height on plantar pressures in sandals. The study included 20 individuals with moderate pes planus that wore five separate sandals; all of which had the same Birkenstock footbed technology with varying arch heights. Visual analog scale (VAS) assessments were done to each participant after performing activities such as, going up and down stairs, 50 ft timed walk at a self-selected pace, and a 50 ft timed walk at one's fastest pace. In addition to VAS, in-shoe plantar pressures were recorded as each participants walked at a self-selected pace. Both the VAS and in-shoe plantar pressures were repeated after a two month wear period. Results indicate that a two month wear time resulted in a decrease in plantar pressures and an increase in overall comfort rating given by the participants. This suggests that for the Birkenstock footbed, a wear time of less than 2 months is necessary to acclimate to the sandal. The results of the study also showed that arch height did have an influence on plantar pressure distribution. While these studies are informative and provide insight on the influence of sandal arch height, it does not address the influence of flip-flop arch height on gait kinetics and kinematics. One limitation to these studies is that all the participants had moderate pes planus (Hillstrom et al., 2005; Kim et al., 2005; Song et al., 2005). More research needs to be done to investigate the effects of flip-flop arch height on individuals with not only pes planus but also individuals with pes cavus and normal arches.

Flip-flops

While there have been a few studies on the effects of sandals on gait mechanics and balance, there is very limited research on the effect of flip-flops specifically. One study by Carl and Barrett (2008) evaluated the differences in peak plantar pressures between flip-flops, athletic shoes, and bare feet. Carl and Barrett (2008) tested the plantar pressures using an in-shoe pressure mapping system of 10 women in a flip-flop condition, an athletic shoe condition and a barefoot condition. The results showed an increase in peak plantar pressures in the flip-flop condition when compared to the athletic sneaker condition. The barefoot condition resulted in the highest plantar pressures when compared to both footwear conditions. The results indicated that flip-flops are not optimal for dissipating ground reaction forces and produced similar peak plantar pressures as barefoot walking (Carl & Barrett, 2008). This study could have been greatly enhanced if the researchers had evaluated different type of flip-flops and had included kinematic measures.

An initial study that investigated the influence of flip-flops and gait kinematics was presented at the American College of Sports Medicine in 2008. This project included 39 people (20 female and 19 male) and looked at the kinetics of individuals under two footwear conditions: athletic sneakers and flip-flops. The participants walked at self selected pace across an AMTI force platform; attack angle (sagittal plane resultant vector) and peak vertical forces were compared within subjects. The results indicated a difference in peak vertical forces between flip-flops and athletic sneakers with flip-flops producing a decreased peak vertical force at heel contact. The study also showed that sex had an influence on the attack angle between flip-flops and athletic sneakers. No differences were noted for the male; however, females had a more vertical attack angle in flip-flops when compared to athletic sneakers. The reason for these differences between sexes is unknown and needs future investigation. The implications of the study are that gait kinetics are altered when flip-flops are compared to athletic sneakers (J. Shroyer, Weimar, Garner III, Knight, & Sumner, 2008). More research is needed to ascertain the causes of these altered kinetics to evaluate if the changes could lead to lower leg injuries or complications. Also, research needs to be done to determine if there are any three-dimensional kinematics or other kinetic variable that may be altered by wearing flip-flops.

Wearing flip-flops has also been linked to abnormal neuromuscular activity in athletes with iliotibial (IT) band friction syndrome, and wearing flip-flops less frequently is recommended by Certified Athletic Trainers to reduce IT band friction syndrome in athletes (Pettitt & Dolski, 2000). Iliotibial band friction syndrome is caused by the iliotibial band rubbing over the prominence of the lateral femoral epicondyle from repetitive flexion and extension of the knee joint in activities such as running. This increased friction causes direct irritation of the IT band or inflammation of the bursa under the epicondyle (Renne, 1975) . In a case study, it was found that limiting the use of flip-flops as a casual footwear choice aided in the rehabilitation of a runner with IT band friction syndrome (Pettitt & Dolski, 2000). This suggests that flip-flops can contribute to pain in the lower extremity and when pain is present, flip-flops are counterproductive to alleviating this pain.

Section Five: Summary of the Literature

The current literature is inundated with various studies that have investigated the effects of footwear on foot development as well as gait mechanics. Much of the research on gait mechanics has been focused on footwear such as athletic shoes, high heels, and rocker shoes and how changes in the architecture of footwear can alter gait. The research on footwear has also taken steps to evaluate other types of more casual shoes such as sandals, but has yet to fully evaluate all of the casual footwear styles. It is apparent from the literature that footwear does alter gait mechanics and muscular activity, and that footwear can be modified to change these observed gait mechanics. The benefit of altering these gait parameters allows clinicians to treat and lower the incidences of lower extremity injuries.

The previous research on footwear contributes to the present project in the following ways. There is a relationship between a person's arch height and the wedge type of a shoe. Specifically, the previous literature shows that varus wedges limit excessive motion of the foot; therefore we expect that people with lower arches will have a greater reduction in foot motion with a flip-flop that has an arch support (varus wedge) than people with high or normal arches. What is important to note here is that while the current literature states these differences in foot motion with various wedges, the "snugness" or fit of the foot in the shoe is of primary importance. In the case of wearing thong flip-flops this "snugness" is not present; it is the lack of this component from the shoe research arena that makes the role of the arch height of the participant and the flip-flops intriguing. It is also why we think the presence of a heel cup takes on such an important role while wearing flip-flops. This project will investigate whether or not a deeper heel cup will provide enough "snugness" for the foot to take advantage of the wedged arches or does the foot simply "slide off" the wedged arch.

Even though there is a rising concern about the flip-flop among the medical community, there is little research that exists regarding thong-style flip-flops. To date only two studies have been done that have researched thong style flip-flops directly, and one case study that made anecdotal suggestions regarding flip-flop usage. Both empirical scientific studies only evaluated gait kinetics. There has been no research to investigate three dimensional gait kinematics or muscular EMG during the gait cycle; therefore, this project will be an attempt to answer basic questions regarding the wear of flip-flops such as what lower extremity kinematics and surface electromyography changes are seen when walking in different styles of thong flip-flops and barefoot. Furthermore, is there merit to the anecdotal claim that walking in flip-flops is analogous to walking barefoot or more like walking barefoot?

CHAPTER III. METHODS

Flip-flops are a popular footwear option, but have been linked to altered gait kinematics (J. F. Shroyer & Weimar, *In Press*) and, anecdotally, to lower extremity discomfort. The current study will build upon a preliminary study by Shroyer & Weimar (*In Press*), and look to explain the kinematic changes that may be leading to lower extremity discomfort. Therefore, the purpose of this investigation was three-fold: (1) to examine the effects that different components of the thong style flip-flop have on gait kinematics in individuals classified with normal arched (NA) feet; (2) to investigate the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified with either low (LA), normal (NA), or high arched (HA) feet; and (3) to determine if there is an increase in muscular activity of the tibialis anterior (TA) at the ankle during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA. The following chapter presents the methods that were used to address the three purposes of the present study and includes the following sections: (a) participants, (b) equipment, (c) procedure, and (d) statistical analysis. The research protocol was approved by the Auburn University Institutional Review Board for Research Involving Human Subjects.

Participants

Seventy-nine college aged female students with an age range of 19 to 25 years served as participants for the study. The overall purpose of this project was to investigate differences in gait mechanics between flip-flop and arch height conditions. A recent study (Shroyer and Weimar, *In Press*) showed that women demonstrate larger variability across footwear types, so to avoid problems with power, interaction effects of sex, and to include the most demonstrative participants, only women were included in the study. A power analysis was conducted (effect size = .25, alpha = .05, and power = .80) with G*Power v3.0.10 for Windows utilizing the O'Brien and Shieh Algorithm and determined that 65 participants would be required to demonstrate significance (Faul, Erdfelder, Lang, & Buchner, 2007). This study included 79 participants to ensure adequate power. For the purpose of this study, people with diagnosed pes cavus or pes planus were not sought, nor was either condition diagnosed during data collection; however, it is believed that people with HA will behave more like people with pes cavus and people with a low arch LA will behave more like people with pes planus. Participants were of various ethnic backgrounds and reportedly in good health. Each participant completed a health screening survey and was excluded if they answer to any question was "yes" and it was deemed by the primary investigator that the "yes" would influence the participant's performance in the study (Appendix M). Exclusion criteria included any current lower extremity injury and any previous low extremity injury that jeopardized the successful performance of tasks utilized in the study. Participants indicated their voluntary willingness to participate by signing an Institutional Review Board approved Inform Consent document before data collection began (Appendix K).

Setting

All testing and data collection took place in the Sport Biomechanics Laboratory at Auburn University. The Sport Biomechanics Laboratory is a large enclosed laboratory with the necessary equipment to carry out this research.

Instrumentation

Foot size and width were measured using a Woman's Brannock Device (BDW01, The Brannock Device Company, Liverpool, NY, USA) to determine appropriate flip-flop sizes (Figure 12). Participants ensured proper fit of flip-flops by reporting appropriate fit following the trying on of each flip-flop type included in the study. A custom Arch Height Index (AHI) Measurement System was used to measure the participants' AHI according to a procedure introduced by Williams and McClay (2000). From the measured AHI, each participant was classified as having a low arch (LA), normal arch (NA), or a high arch (HA). The previously mentioned measurement method to determine the arch height index has been used throughout literature (Butler et al., 2006; Butler et al., 2007; Butler, Hillstrom, Song, Richards, & Davis, 2008) and participants' AHI was measured through the use of a modified AHI Measurement Device that was constructed for measurement of the participants AHI (Figure 13).

Figure 12. Brannock device (Sanders, 2008).

Figure 13. Arch Height Measurement Index device.

Kinematics

A Vicon MX motion analysis system (Vicon, Los Angeles, CA, USA) was used to collected three-dimensional kinematic measures of the foot, ankle, and lower leg during each testing condition. Six cameras were positioned to record all three cardinal planes (frontal, sagittal, and transverse) of the participant during one full gait cycle. Kinematic data were captured at 100Hz (Figure 14).

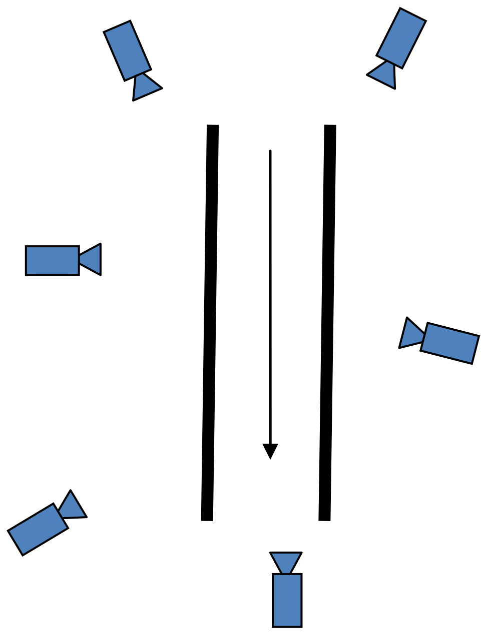


Figure 14. Vicon camera motion capture setup.

Spatial positions of spherical 14 mm retro-reflective markers (MKR-6.4, B & L Engineering, Tustin, California, USA) attached with double sided tape (CLEAR-1R36, Hair Direct Inc., Bainbridge, Pennsylvania, USA) to the right side of the participants' lower leg, were captured to determine movement strategies of the foot, ankle, and lower leg during gait. The Oxford Foot Model (OFM) was implemented as defined by Carlson et al. (2001) to identify and calculate kinematic variables of the foot, ankle and lower leg.

Seventeen markers were placed on the lower body on the right leg according to the Oxford Foot Model. These markers included 14 permanent markers and 3 temporary markers. Table 3 and Figure 15 describe marker descriptions and applied segments.

Table 3

Names and positions of markers used in Oxford Foot Model

Marker name	Position	Segment
KNE	Femoral condyle	Femur
HFB	Head of fibula	Tibia
TTB	Tibial tuberosity	Tibia
SHN	Anterior aspect of the shin	Tibia
ANK	Lateral malleolus	Tibia
MMA	*Medial malleolus*	*Tibia*
CPG	Wand marker on posterior calcaneus	Hindfoot
HEE	Posterior distal aspect of calcaneus	Hindfoot
PCA	*Posterior proximal aspect of calcaneus*	*Hindfoot*
LCA	Lateral calcaneus	Hindfoot
STL	Sustentaculum tali	Hindfoot
P1MT	Proximal and dorsal 1st metatarsal	Forefoot
D1MT	*Distal and medial 1st metatarsal*	*Forefoot*
P5MT	Proximal and lateral 5th metatarsal	Forefoot
D5MT	Distal and lateral 5th metatarsal	Forefoot
TOE	Between 2nd and 3rd distal metatarsals	Forefoot
HLX	Proximal end of 1st phalanx	Hallux

Note. Names in italics are for static trials only and were removed for dynamic trials

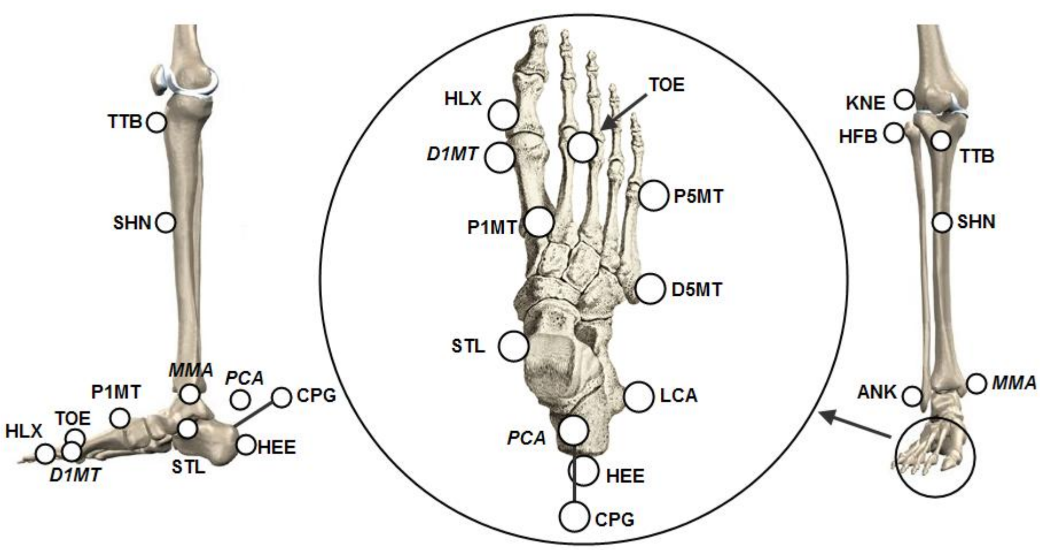


Figure 15. Anterior, medial and dorsal view of OFM marker locations.

The OFM is comprised of a rigid tibial segment (tibia and fibula), a rigid hindfoot (calcaneus and talus), a forefoot (metatarsals), and a hallux (proximal phalanx of the hallux) (Carson et al., 2001; Stebbins, Harrington, Thompson, Zavatsky, & Theologis, 2006) (Figure 16).

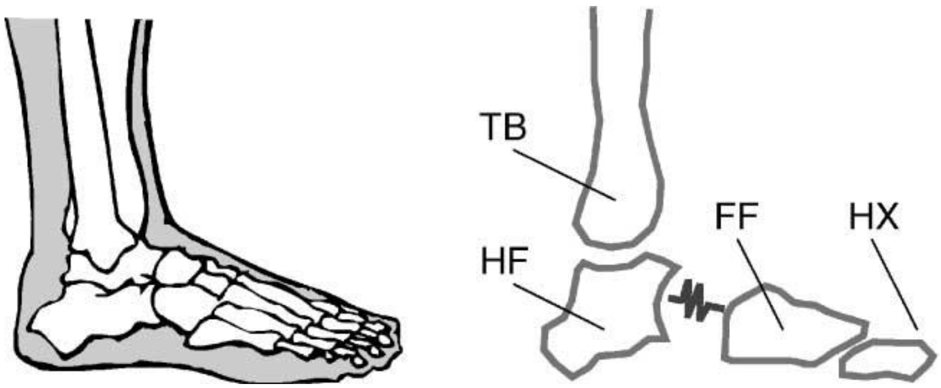


Figure 16. Drawing of the three rigid foot segments of the OFM (Carson et al., 2001).

The segment axes of the tibia were defined as the intersection of: (1) frontal plane: defined by the medial malleolus (MMA), the lateral malleolus (LMA), and the fibular head (HFB) and (2) sagittal plane: defined by the tibial tuberosity (TTB), the midpoint of the lateral and medial mallioli (LMA, MMA), and the shin (SHN) (Carson et al., 2001; Stebbins et al., 2006), (Figure 17).

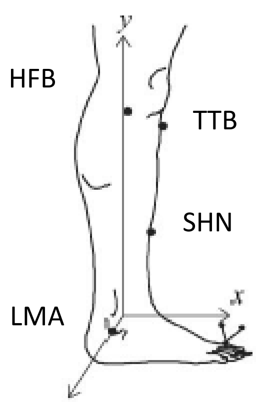


Figure 17. Tibia segment axes (Carson et al., 2001).
Note. positive y is upward, positive x is anterior and positive z is lateral.

The segment axes of the hindfoot were defined as the intersection of: (1) sagittal plane: defined by the distal posterior calcaneus (HEE), the proximal posterior calcaneus (PCA) and the midpoint of the lateral and medial mallioli (LMA, MMA), and (2) transverse plane: defined by a ray parallel to the floor through the distal posterior calcaneus (HEE) (Carson et al., 2001; Stebbins et al., 2006), (Figure 18).

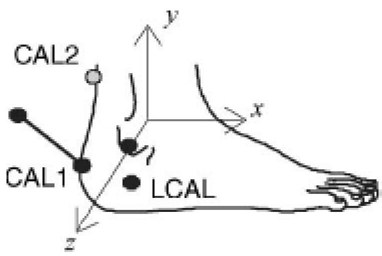


Figure 18. Hindfoot segment axes (Carson et al., 2001).
Note. positive y is upward, positive x is anterior and positive z is lateral.

The segment axes of the forefoot were defined as the intersection of: (1) transverse plane: defined by the distal medial 1st metatarsal (D1MT), the proximal lateral 5th metatarsal (P5MT), and the distal lateral 5th metatarsal (D5MT), and (2) sagittal plane: defined by the mid 2nd and 3rd distal metatarsal (TOE), one-third the distance from the proximal/dorsal 1st metatarsal (P1MT) and 5th metatarsal (P5MT) (Carson et al., 2001; Stebbins et al., 2006), (Figure 19).

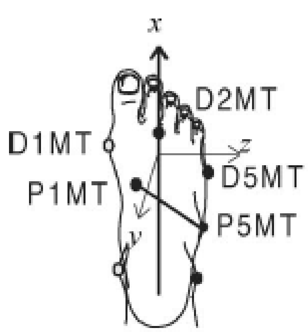


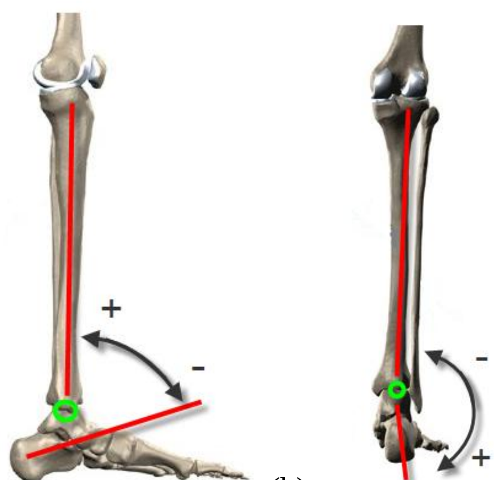
Figure 19. Forefoot segment axes (Carson et al., 2001).
Note. positive y is upward, positive x is anterior and positive z is lateral.

The segment axes of the hallux were defined as: (1) sagittal plane: defined by a ray perpendicular to the floor (from static calibration) and through the proximal end of 1st phalanx (HLX), and (2) transverse plane: defined by a ray parallel to the floor (from static calibration) (Carson et al., 2001).

Angles of rotation were calculated according to the joint coordinate systems proposed by Grood and Suntay (1983) and are defined as the following:

For the hindfoot relative to the tibia

- Plantar/dorsiflexion: about the transverse axis of the distal tibia (Figure 20a).
- Inversion/eversion: about the long axis of the hindfoot (Figure 20b).



(a) (b)

Figure 20. (a) Plantar/dorsiflexion with dorsiflexion positive and plantar flexion negative (b) Inversion/eversion with inversion positive and eversion negative.

For the forefoot relative to the hindfoot

- Supination/pronation: about the long axis of the forefoot (Figure 21).

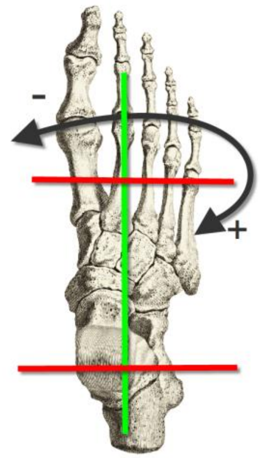


Figure 21. Supination/pronation with pronation negative and supination positive.

For the hallux relative to the forefoot

- Extension/flexion: about the transverse axis of the distal forefoot (Figure 22).

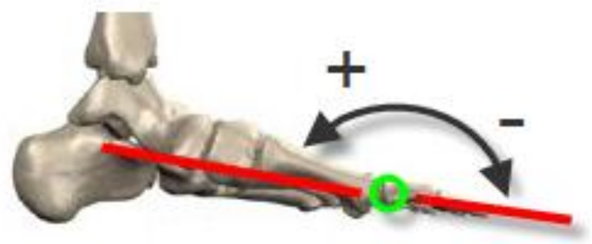


Figure 22. Hallux extension/flexion with extension positive and flexion negative.
Note. Line representing the hallux has been extended for ease of understanding.

The gait cycle used for the current study included right-foot/initial contact, right-foot/midstance, right-foot/toe-off, left-foot/initial contact, left-foot/midstance, left-foot/toe-off, and a second right-foot/initial contact (Figure 4). This cycle was used to ensure that the right foot and lower leg were captured in the capture volume for a full stride length, and to evaluate specific points during the gait cycle as well as periods between those specific points. In addition, these events correlate to the events in the OFM.

Surface Electromyography (sEMG)

Lower leg muscular activity of the tibialis anterior (TA) was collected using Ambu® Blue Sensor M-00-S self-adhesive (Ag/AgCl) snap (1 cm^2) dual electrodes (Ambu®, Ballerup, Denmark) (Figure 23). The sEMG preamplifier leads were connected to a Noraxon Telemyo 2400T V2 wireless transmitter (Noraxon U.S.A. Inc., Scottsdale, Arizona, USA) (Figure 24) which was connected to the Noraxon Telemyo 2400R – World Wide Telemetry receiver (Noraxon U.S.A. Inc., Scottsdale, Arizona, USA) (Figure 25). The signal was sampled at 1000HZ, and post processing included full wave rectification and a finite impulse response (FIR) filter (gain set at 1000, a bandpass of 10-350 Hz, and a Lancosh window of 79 points) using MyoResearch XP Master Edition 1.02.25 Software (Noraxon U.S.A. Inc., Scottsdale, Arizona, USA). The sEMG activity of the TA was recorded to measure the contribution of the TA to the dorsiflexion moment at the ankle during the swing phase of the gait cycle. The peak sEMG of the non-support leg during the single stance period was also included in the analysis to gain a better understanding of the activity of the TA at the end of the swing phase.

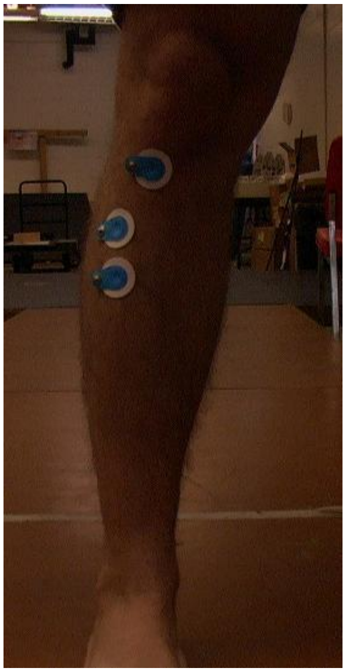


Figure 23. Tibialis anterior sEMG electrode placement.

Figure 24. Noraxon Telemyo 2400T V2 wireless transmitter

Figure 25. Noraxon Telemyo 2400R – World Wide Telemetry receiver

The sEMG and kinematic data were synchronized with a Vicon MX Control 64-channel A/D board (Vicon, Los Angeles, CA, USA) and all trials were videotaped with a Canon 3CCD Digital Video Camcorder GL2 NTSC (Canon U.S.A., Inc., Lake Success, NY, USA) for motion verification. During data collection, all data was stored on a computer and analyzed later offline. Kinematic data were analyzed with Nexus Polygon Software (Vicon, Los Angeles, CA, USA).

Footwear

The three flip-flop conditions utilized included: (a) flip-flop 1 (FF1), the "modified" Reef Ginger (Reef, San Diego, CA, USA) flip-flops had no heel cup or arch support (Figure 26), (b) flip-flop 2 (FF2), the Reef Ginger (Reef, San Diego, CA, USA) had a averaged 9.4 mm arch support but no heel cup (Figure 27), and (c) flip-flop 3 (FF3), the SOLE Platinum Sport Flips (Edge Marketing Inc., Great Falls, MT, USA) with both a heel cup and arch support as well as midtarsal support, toe ridge, and ridged rubber outersole (Figure 28). The Sole Platinum Sport Flips also had wider and longer straps than did either of the Reef Ginger flip-flops. The modified Reef Ginger (FF1) was identical in material and sized to the Reef Ginger (FF2), with the exception of an added arch support present in the Reef Ginger. Table 4 depicts measurement of the three flip-flop conditions. Figure 29 is a visual comparison of the arch supports of the three flip-flop conditions.

Figure 26. Flip-flop 1: "Modified" Reef Ginger without arch support or heel cup.

Figure 27. Flip-flop 2: Reef Ginger with arch support but no heel cup.

Figure 28. Flip-flop 3: SOLE Platinum Sport Flip with both a heel cup and arch support as well as midtarsal support, toe ridge, and ridged rubber outersole.

Table 4

Average measurements of flip-flop features (sizes 6 -10)

	FF1	FF2	FF3
Lateral heel cup height (mm)	–	–	26.4
Medial heel cup height (mm)	–	–	24.8
Arch height (mm)	–	9.4	22.4
Heel bed height (mm)	19.6	19.6	13.6
Forefoot bed height (mm)	14.8	14.8	14.4

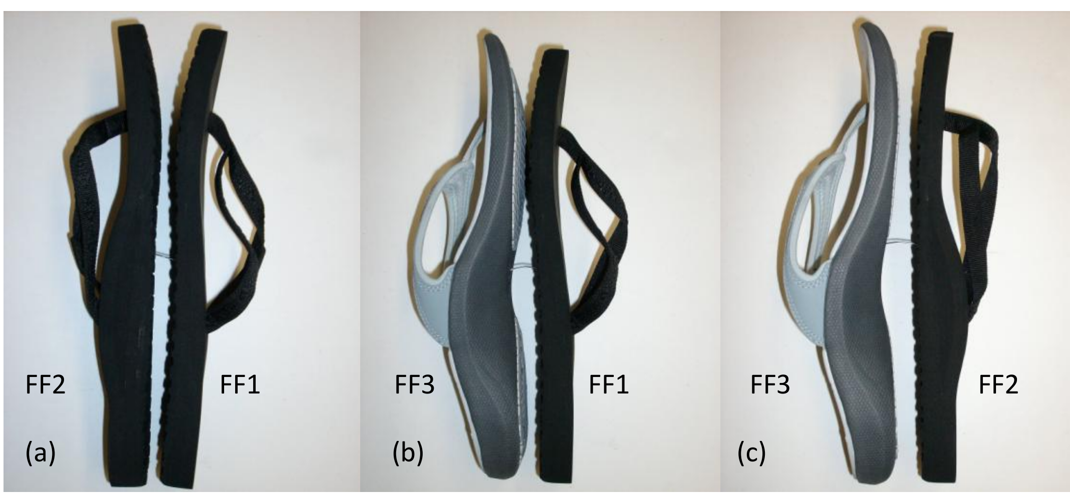


Figure 29. Visual comparison of arches for flip-flops: (a) FF2 and FF1, (b) FF3 and FF1, and (c) FF3 and FF2.

Design and Procedures

After meeting the criteria for participation, the feet of participants were measured for appropriate footwear size using a Brannock Device (Figure 12) to ensure proper fit the flip-flops. Flip-flops were then tried on to ensure proper shoe size selection. Participants were then prepared for retroreflective marker placement. Participants were asked to wear spandex shorts with shirt tucked in to minimize clothing artifact. The participant's legs were then wiped with a 91% alcohol solution. Next, each marker placement site was marked with permanent marker. Participants then were prepared for surface electromyography (sEMG) following the recommendations of SENIAM (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). The TA on the right leg was identified by palpation as the participant dorsiflexed the right ankle. The area was then swabbed with a 70% alcohol solution and abraded to reduce the electrical impedance of the skin. Two sEMG electrodes were placed 2 cm apart over the junction of the proximal and middle thirds of the tibia where the largest palpable portion of the muscle belly is located and parallel to the alignment of the muscle fibers. Proper placement was checked by manual muscle testing. A ground electrode was placed on the boney aspect of the tibial tuberosity. A stretch "pre-wrap" was placed over the electrodes to prevent electrode movement. Next, 17 markers (3 temporary and 14 trial markers) were affixed to the skin with double sided tape and placed according to the requirements of the OFM for the lower leg, ankle, and foot (Carson et al., 2001) (Figure 15).

A static capture was performed while the participants stood in the center of the capture volume in the anatomical position with shoulders abducted. Once the static capture was performed, the markers were identified and labeled to comply with the OFM. Following the static capture and application of the model, temporary markers were removed. Participants then performed dynamic trials which required the participant to walk at a self-selected pace through the motion capture system's capture volume under four separate conditions: barefoot (BF), flip-flop 1 (FF1), flip-flop 2 (FF2), and flip-flop 3 (FF3). Each flip-flop condition was performed three times in a row and the order of flip-flop conditions was randomized for each participant using a random number generator. Each participant was allowed to rest as needed between conditions and between each trial of every condition. Each flip-flop condition was separated by a barefoot trial where the participants walked through the capture volume barefoot. Incorporating the barefoot condition between footwear conditions allowed participants to have the same condition prior to each footwear condition and was in effort to decrease any affect that the type of footwear participants wore to data collection had on the first flip-flop trial, as well as reduce the influence of previous trials.

Following kinematic and sEMG data collection, the participant's arch height index was measured using the Arch height Measurement Index Device (Figure 13). The right foot was measured while the participants were in bilateral stance. Measurements taken included foot length, truncated foot length (the distance from the heel to the first metatarsophalangeal joint), and the height of the dorsum of the foot at half the total length of the foot (Figure 30) (Butler et al., 2006). The average AHI was calculated and participants with an AHI that was greater than one standard deviation less than the sample mean were considered LA. Participants with an AHI that was greater than one standard deviation more than the sample mean were considered HA. Participants with an AHI that were within one standard deviation of the sample mean were considered NA.

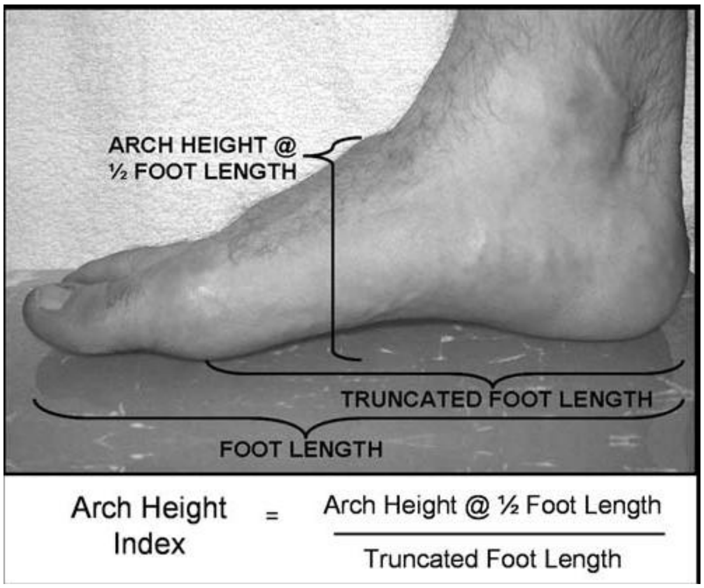


Figure 30. Arch Height Index (Butler et al., 2007).

Experimental Design and Data Analysis

All statistical analyses were conducted using SPSS software (version 16.0; SPSS Inc., Chicago, Illinois, USA) and an alpha level of statistical significance was set aprior at $p \leq 0.05$ ("SPSS for Windows," 2007); all data were stored in an Excel database and then imported into SPSS for analysis. To investigate the effects of footwear on gait kinematics in individuals classified with NA, the kinematic data were analyzed by conducting a one-way multivariate repeated measure ANOVA. The independent variable was footwear with 4 levels: BF, FF1, FF2, and FF3. The dependent variables were stride length (SL), ankle eversion at midstance (EV), peak ankle eversion (EV_{PEAK}), pronation at midstance (PRO), peak pronation (PRO_{PEAK}), peak dorsiflexion during swing phase ($DORSI_{PEAK}$), and peak hallux flexion during swing phase (HX_{PEAK}).

To investigate the effects of footwear arch support and foot arch type on gait kinematics, the kinematic data were analyzed by conducting a two-way multivariate repeated measure ANOVA. The independent variables were footwear and foot arch type. The independent variable of footwear had three levels: BF, FF1, and FF2. The independent variable of foot arch type had three levels: NA, HA, and LA. The dependent variables were SL, EV, EV_{PEAK}, PRO, PRO_{PEAK}, $DORSI_{PEAK}$, and HX_{PEAK}.

To investigate the effects of footwear on sEMG in individuals classified with NA, the sEMG data for the TA were analyzed by conducting a one-way multivariate repeated measures ANOVA. The independent variable of footwear had four levels: BF, FF1, FF2, and FF3. The dependent variables were the averaged sEMG activity (EMG_{AVG}) and the peak sEMG (EMG_{PEAK}) of the TA muscle during the swing phase for individuals classified with NA.

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

CHAPTER IV. RESULTS

Flip-flops are a popular footwear option, but have been linked to altered gait kinematics (Shroyer & Weimar, *In Press*) and anecdotally to lower extremity discomfort. This study will build upon the preliminary study by Shroyer & Weimar (*In Press*), and look to the kinematic changes that may lead to lower extremity discomfort. Therefore, the purpose of this investigation was three-fold: (1) to examine the effects that different components (i.e. arch support, midtarsal support, toe ridge, and longer/thicker straps) of the thong style flip-flop have on gait kinematics in individuals classified with normal arched (NA) feet; (2) to investigate the effects of a thong style flip-flop arch support on gait kinematics of individuals classified with either low (LA), normal (NA), or high arched (HA) feet; and (3) to determine if there is an increase in muscular activity of the tibialis anterior (TA) during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA. The following chapter presents results of the current study and includes sections; Section 1: Participant demographics, Section 2: Effect of footwear on gait kinematics in individuals classified with NA, Section 3: Effect of footwear arch type on gait kinematics in individuals classified with LA, NA, and HA, and Section 4: Effect of footwear on TA surface electromyography (sEMG) in individuals classified with NA.

Section 1: Participant Demographics

Seventy-nine females who volunteered for participation in the current study met the initial qualifications. Upon answering the medical questionnaire, all volunteers were included in the study and each participant signed an Informed Consent document approved by the University Institutional Review Board for Human Subjects (Appendix K). The averages of the participants' demographics are summarized in Table 5. For arch height classification, participants with an AHI greater than .3749 were classified HA. Participants with an AHI between .3749 and .3092 were classified NA. Participants with an AHI lower than .3092 were classified LA. The sample population used for this study included 14 HA, 53 NA, and 11 LA.

Table 5
Subject demographics

	Mean	Standard Deviation
Age (years)	21.54	1.526
Height (m)	1.646	.5833
Mass (kg)	63.53	10.61
Foot Size	8.10	1.165
Arch Height Index (AHI)	.3420	.03282

Section 2: Normal Arch and Gait Kinematics

The first research question proposed by the current study was to examine the effects that different components of the thong style flip-flop have on gait kinematics in individuals classified with NA feet. This section includes the results for that research question. For all kinematic variables, one participant's data had to be omitted from the present study due to erroneous data. This omission decreased the sample population from $N = 79$ to $N = 78$. Fifty-three individuals of the $N = 78$ sample population were classified as having a NA, which made the sample size for NA and gait kinematic results $N = 53$.

The multivariate repeated measures ANOVA with a *P* value set *apriori* at $p < .05$ yielded a significant effect of footwear on the kinematic variables stride length (SL), eversion at midstance (EV), peak eversion during stance phase (EV_{PEAK}), pronation at midstance (PRO), peak pronation during stance phase (PRO_{PEAK}), peak dorsiflexion during swing phase ($DORSI_{PEAK}$), and peak hallux extension in the beginning swing phase (HX_{PEAK}), (Wilks's $\Lambda = .088$, $F(21,32) = 15.792$, $p < .001$, $\eta^2 = .912$, $Power = 1.000$). Descriptive statistics for all dependent variables are presented in Appendix A.

Stride Length

A follow-up repeated measures ANOVA indicated a significant effect of footwear on SL ($F(3,52) = 55.555$, $p < .001$, $\eta^2 = .517$, $Power = 1.000$). A post hoc test using the Least Significant Difference (LSD) yielded a significant difference between BF and all flip-flop conditions: BF and FF1 ($p < .001$), BF and FF2 ($p < .001$), BF and FF3 ($p < .001$). For the flip-flop conditions, there was only a significant difference between: FF1 and FF2 ($p = .042$). There was no significant difference between any of the other flip-flop conditions: FF1 and FF3 ($p = .843$), and FF2 and FF3 ($p = .056$) (Figure 31). All flip-flop conditions yielded longer stride lengths when compared to walking barefoot; and FF1 yielded a longer stride length than FF2.

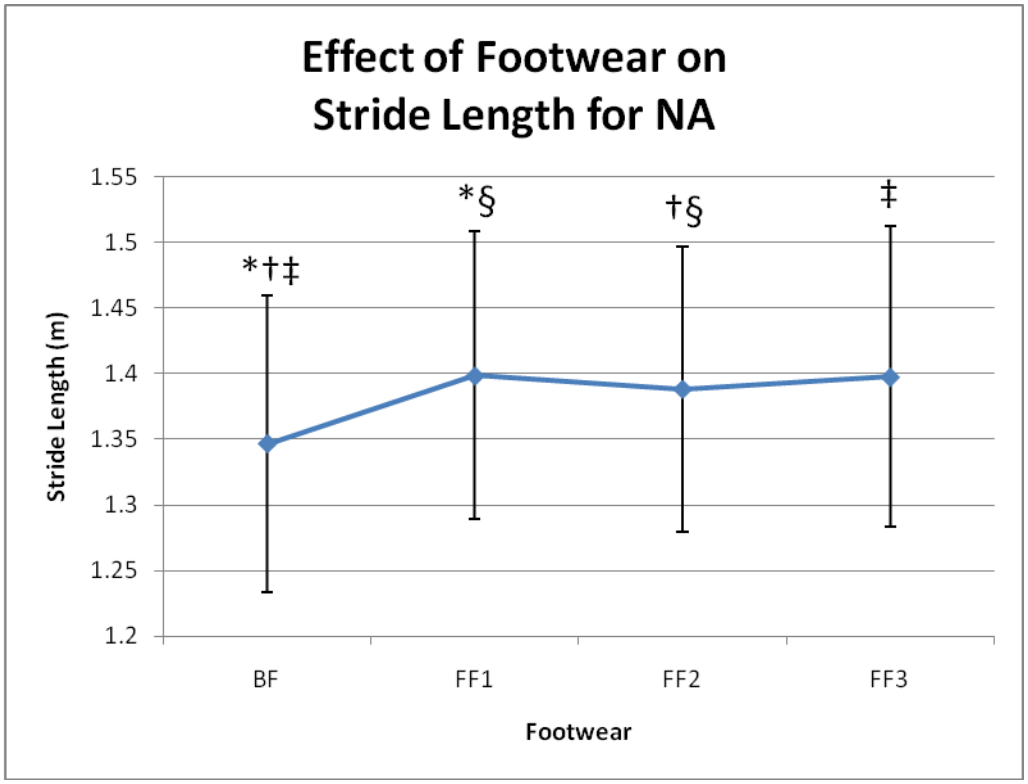


Figure 31. Effect of footwear on stride length for NA participants.
$^{*\dagger\ddagger}p < .001$. $^{\S}p = .042$.

Eversion

The eversion angle (EV) is defined as the relationship between the tibia and the hindfoot at the ankle in the frontal plane about the anterior/posterior axis (Figure 20b). A follow-up repeated measures ANOVA indicated a significant effect of footwear on EV ($F(2.438,52) = 9.624$, $p < .001$, $\eta^2 = .156$, *Power* = .991). Post hoc test using the LSD criterion yielded a significant difference between BF and FF1 ($p < .001$) and BF and FF2 ($p = .016$). There was no significant difference between BF and FF3 ($p = .502$). A significant difference was found between flip-flop FF1 and FF3 ($p < .001$) and FF2 and FF3 ($p = .003$). There was no significant difference between FF1 and FF2 ($p = .127$) (Figure 32). Based on Oxford Foot Model (OFM) output, a smaller number indicates greater eversion than a larger number (the angle between the calcaneus and the tibia was smaller); therefore, these data suggest that ankle eversion is greatest in the BF (M = -4.00 deg) and FF3 (M = -4.11 deg) conditions and smallest in the FF1 (M = -3.39 deg) and FF2 (M = -3.55 deg) conditions. This implies that during the midstance phase, the ankle everts the same degree in BF as it does in FF3. Also, the ankle everts the same degree in FF1 as it does in FF2.

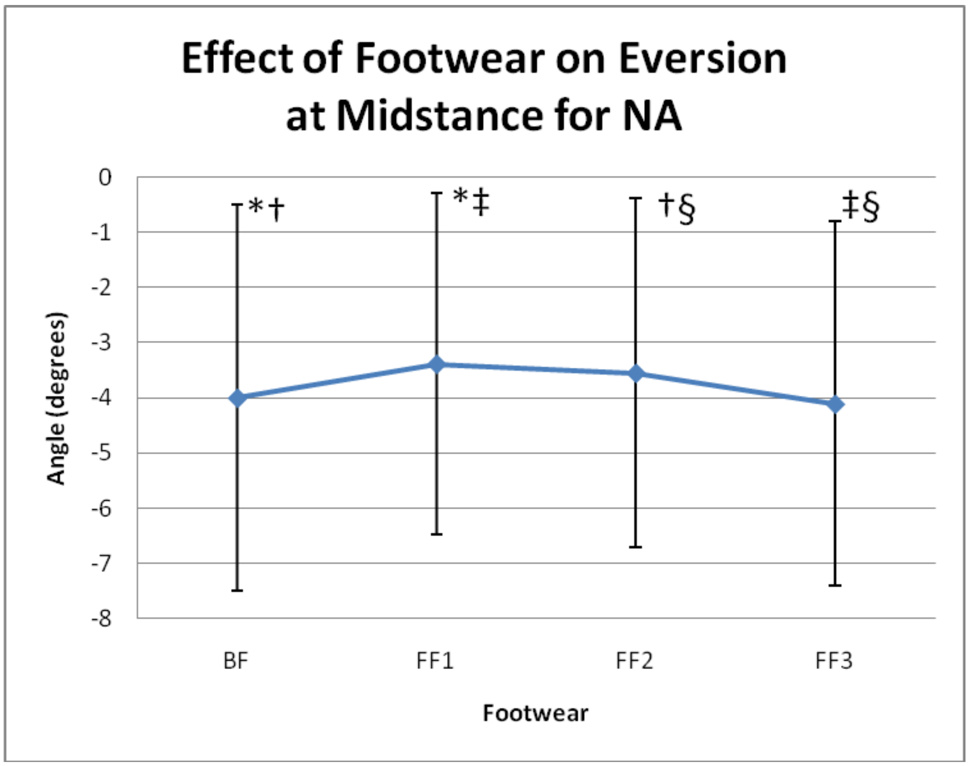


Figure 32. Effect of footwear on eversion during midstance for NA participants.
$^{*\,\ddagger}p < .001$. $^{\dagger}p = .016$. $^{\S}p = .003$.

Peak Eversion

Peak eversion (EV_{PEAK}) occurred between heel off and toe off during the stance phase. Eversion is defined as the relationship between the tibia and the hindfoot at the ankle in the frontal plane about the anterior/posterior axis (Figure 20b). A follow-up repeated measures ANOVA indicated a significant main effect for footwear on EV_{PEAK}

($F(3,52) = 20.889$, $p < .001$, $\eta^2 = .287$, *Power* = 1.000). Post hoc pairwise comparisons using the LSD criterion yielded a significant difference in EV_{PEAK} between BF and each footwear condition: BF and FF1 ($p < .001$), BF and FF2 ($p < .001$), BF and FF3 ($p = .001$). Within flip-flop conditions, there was a significant difference between FF1 and FF3 ($p = .001$), FF2 and FF3 ($p < .001$); however, there was no significant difference between FF1 and FF2 ($p = .802$) (Figure 33). These results imply that FF3 resulted in an EV_{PEAK} closer to BF than FF1 and FF2.

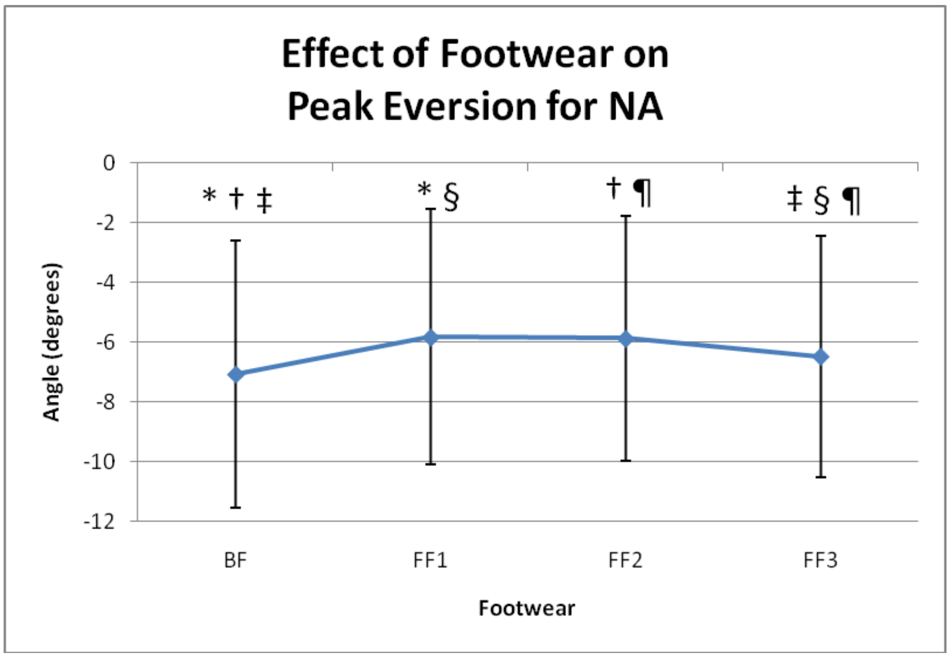


Figure 33. Effect of footwear on peak eversion for NA participants.
$^{*\,\dagger\,\P}p < .001$. $^{\ddagger\,\S}p = .001$.

Pronation

A follow-up repeated measures ANOVA indicated no significant effect of footwear on PRO ($F(3,52) = .578$, $p = .630$, $\eta^2 = .011$, *Power* = .168), indicating no statistical difference in pronation angles between BF, FF1, FF2, or FF3 (Figure 34). This result suggests that each flip-flop condition permitted pronation akin to barefoot.

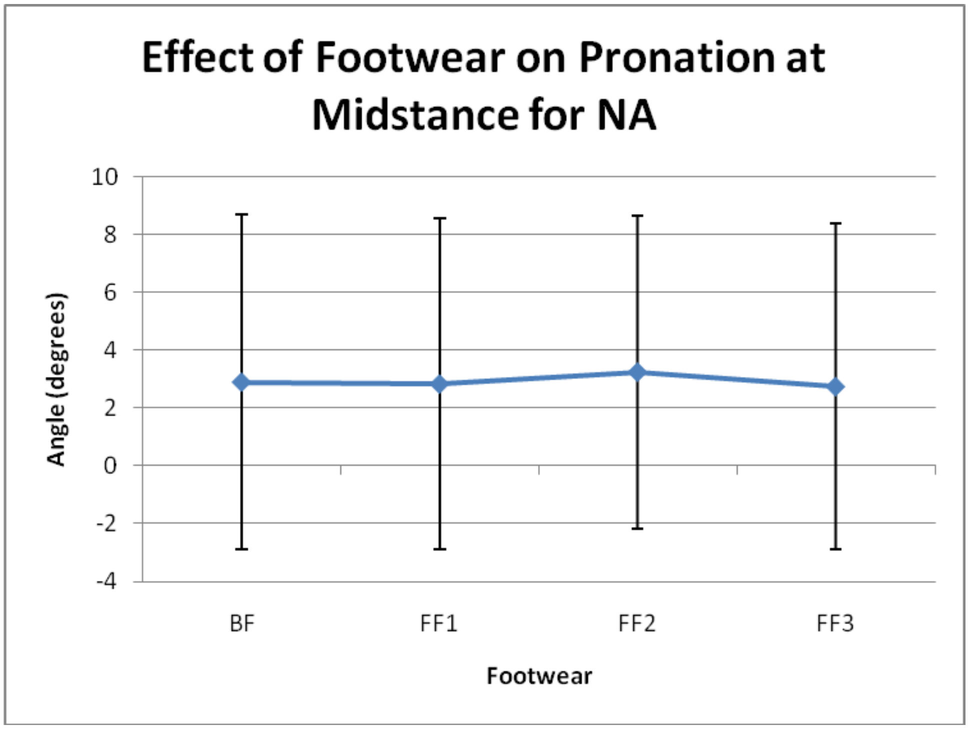


Figure 34. Effect of footwear on pronation during midstance for NA participants, p = .630.

Peak Pronation

A follow-up repeated measures ANOVA indicated no significant effect of footwear on PRO_{PEAK} ($F(3,52) = .963$, $p = .412$, $\eta^2 = .018$, *Power* = .259), indicating no difference in peak pronation angles between BF, FF1, FF2, or FF3 (Figure 35). This result suggests that each flip-flop condition permitted peak pronation analogous to barefoot.

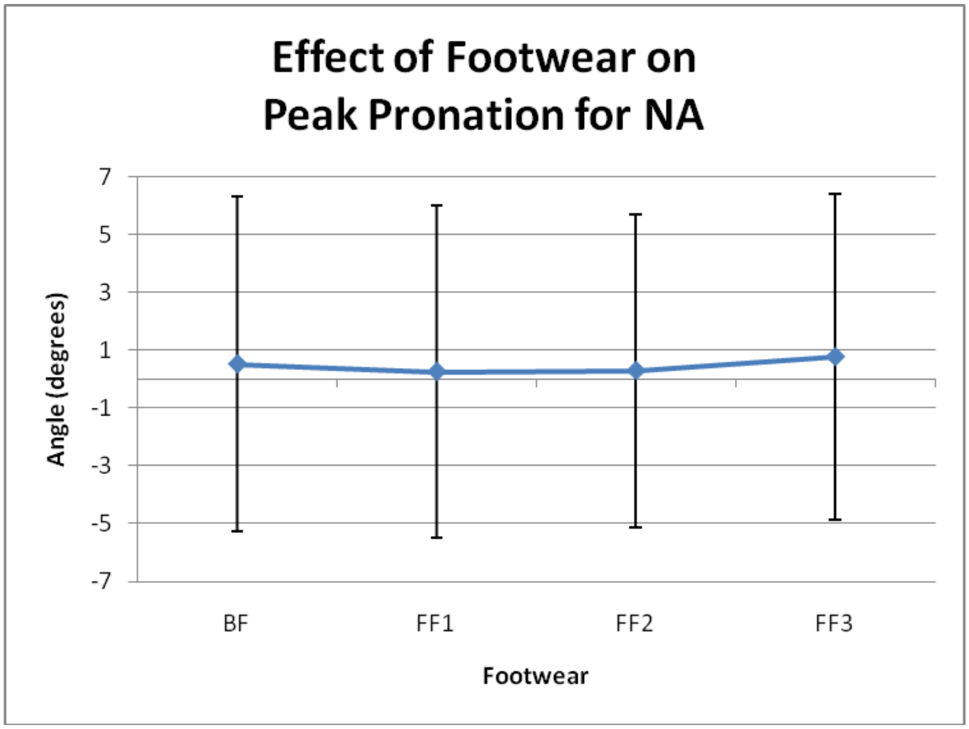


Figure 35. Effect of footwear on peak pronation for NA participants, $p = .412$.

Peak Dorsiflexion

Dorsiflexion, for the purposes of this study, was defined as the relationship of the hindfoot relative to the tibia about the transverse axis of the distal tibia (Figure 20a). A follow-up repeated measures ANOVA indicated a significant effect of footwear on $DORSI_{PEAK}$ ($F(2.414,52) = 11.471$, $p < .001$, $\eta^2 = .181$, *Power* = .997). A post hoc test using the LSD criterion yielded a significant difference between BF and FF1 ($p < .001$) as well as BF and FF2 ($p < .001$) but no significant difference between BF and FF3 ($p = .059$) was noted. Statistical significance was also found between FF1 and FF3 ($p = .001$), and FF2 and FF3 ($p = .016$); however, there was no significant difference between FF1 and FF2 ($p = .311$). The OFM generates a larger value for greater dorsiflexion (or the angular distance between the tibia and the foot is smaller), so the BF condition yielded the greatest dorsiflexion angle compared to all the flip-flop conditions during the swing phase except FF3. There was no difference between FF1 and FF2 for $DORSI_{PEAK}$. Results indicate that dorsiflexion was more prominent in BF and FF3 conditions than FF1 and FF2 (Figure 36).

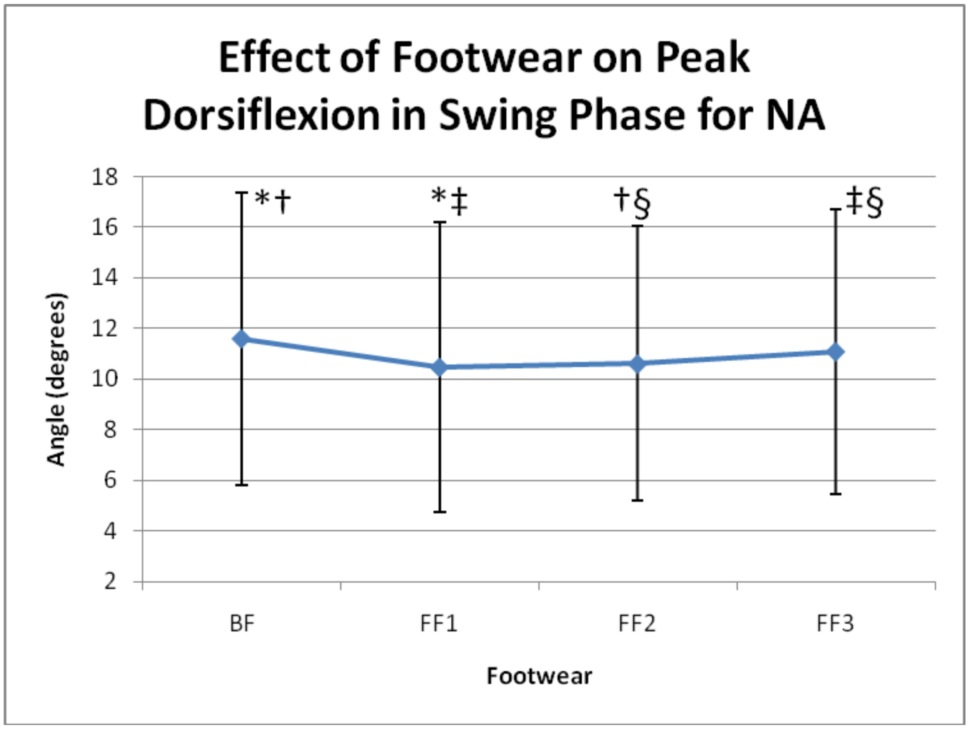


Figure 36 – Effect of footwear on peak dorsiflexion during the swing phase for NA participants.
$^{*\,\dagger}p < .001$. $^{\ddagger}p = .001$. $^{\S}p = .016$.

Peak Hallux Extension

A follow-up repeated measures ANOVA indicated no significant effect of footwear on HX_{PEAK} (F(3,52) = 1.249, p = .294, η^2= .023, *Power* = .330). There was no difference in peak hallux extension, after toe off, between any footwear condition (Figure 37).

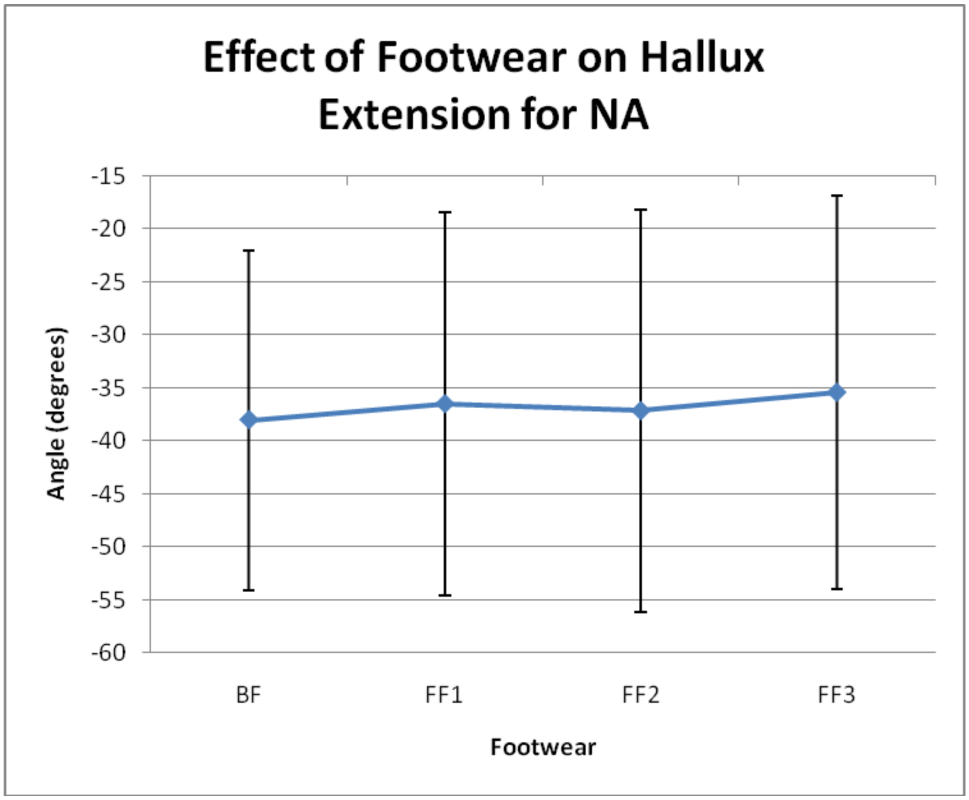


Figure 37. Effect of footwear on hallux extension for NA participants, p = .294.

Section 3: Footwear Arch and Arch Foot Type Kinematics

The second research question proposed by the current study was to examine the effects of a thong style flip-flop arch support on gait kinematics of individuals classified with either LA, NA, or HA feet. This section includes the results for that research question. The multivariate repeated measures ANOVA with a P value set *apriori* at $p <$ .05 yielded no significant interaction effect of footwear and foot arch type on the kinematic variables SL, EV, EV_{PEAK}, PRO, PRO_{PEAK}, $DORSI_{PEAK}$, and HX_{PEAK}, (Wilks's $\Lambda = .589$, $F(28,1124) = 1.342$, $p = .139$, $\eta^2 = .233$, *Power* = .929). There was a significant main effect of footwear on the kinematic variables SL, EV, EV_{PEAK}, PRO, PRO_{PEAK}, $DORSI_{PEAK}$, and HX_{PEAK}, (Wilks's $\Lambda = .187$, $F(14,62) = 19.269$, $p < .001$, $\eta^2 = .813$, *Power* = 1.000). The multivariate tests yielded no significant effect of foot arch type on the kinematic variables SL, EV, EV_{PEAK}, PRO, PRO_{PEAK}, $DORSI_{PEAK}$, and HX_{PEAK} (Wilks's Λ = .731, $F(14,138) = 1.670$, $p = .069$, $\eta^2 = .145$, *Power* = .874). Descriptive statistics are presented in Appendix B.

Stride Length

A follow-up repeated measures ANOVA indicated no significant interaction effect of footwear and arch type on SL ($F(4,77) = .442$, $p = .778$, $\eta^2 = .012$, *Power* = .152) (Figure 38).

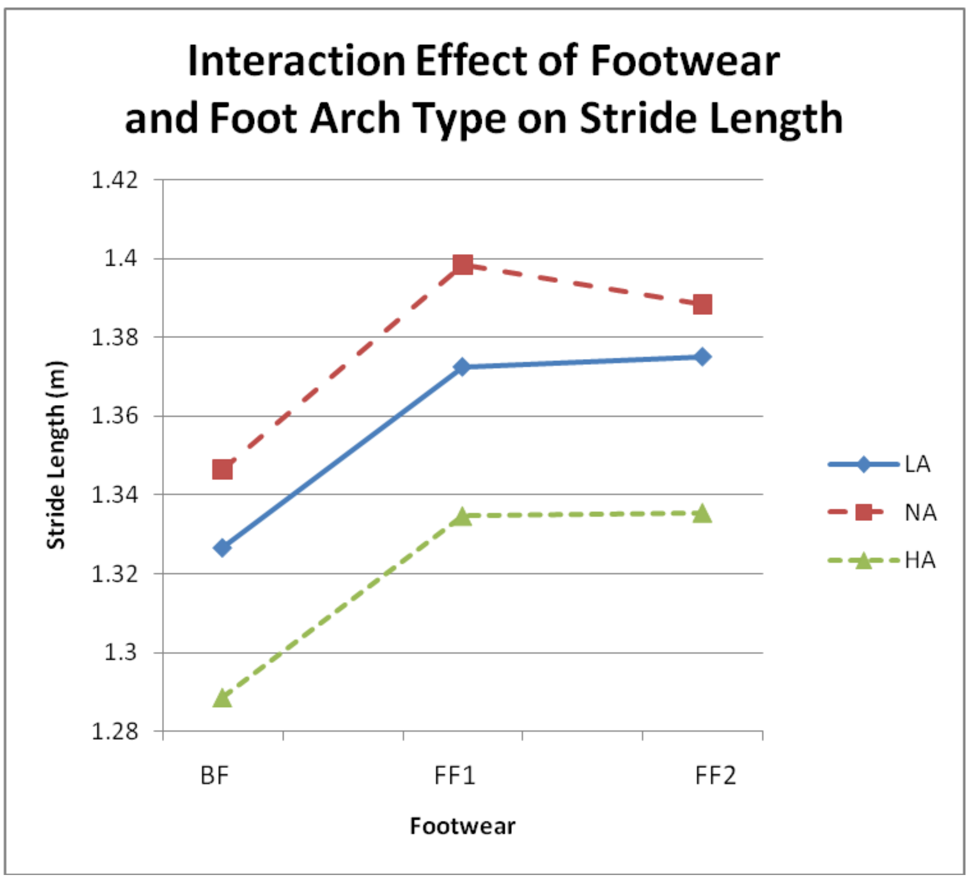


Figure 38. Interaction effect of footwear and foot arch type on stride length, $p = .778$.

There was a main effect of footwear on SL (F(2,77) = 52.090, $p < .001$, $\eta^2 = .410$, *Power* = 1.000). A post hoc test using the LSD yielded a significant difference between barefoot and both flip-flop conditions: BF and FF1 ($p < .001$) and BF and FF2 ($p < .001$). Both flip-flop conditions yielded longer stride lengths when compared to walking barefoot but there was no significant difference between the flip-flop conditions: FF1 and FF2 ($p = .644$) (Figure 39).

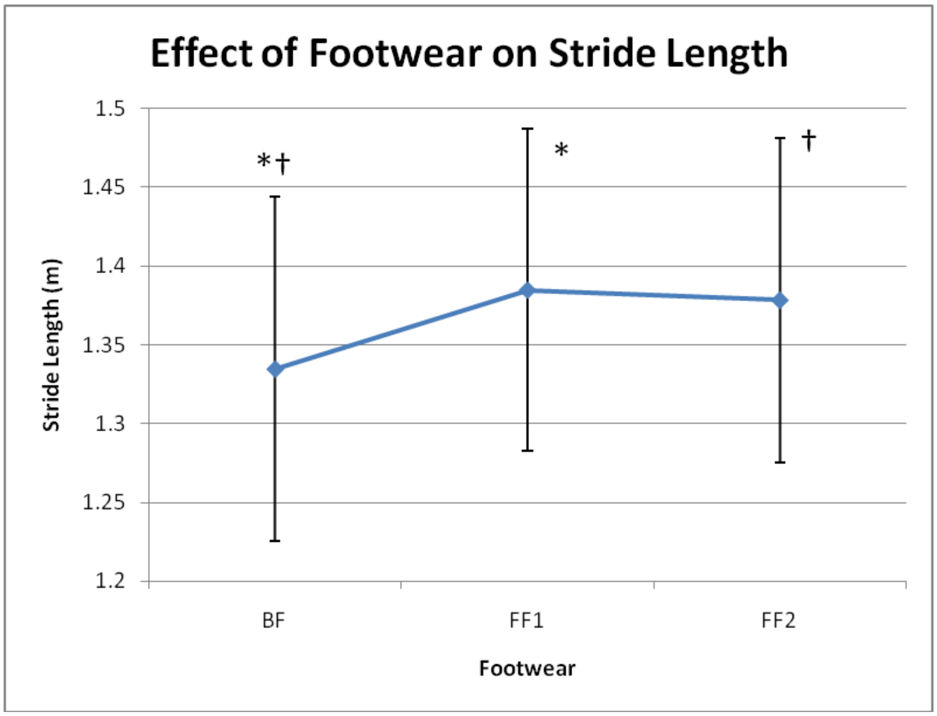


Figure 39. Effect of footwear on stride length.
$^{*\dagger}p < .001$.

There was no significant main effect of foot arch type on SL (F(2,77) = 1.531, p = .223, $\eta^2 = .039$, *Power* = .316) indicating an individual's arch type does not affect stride length (Figure 40).

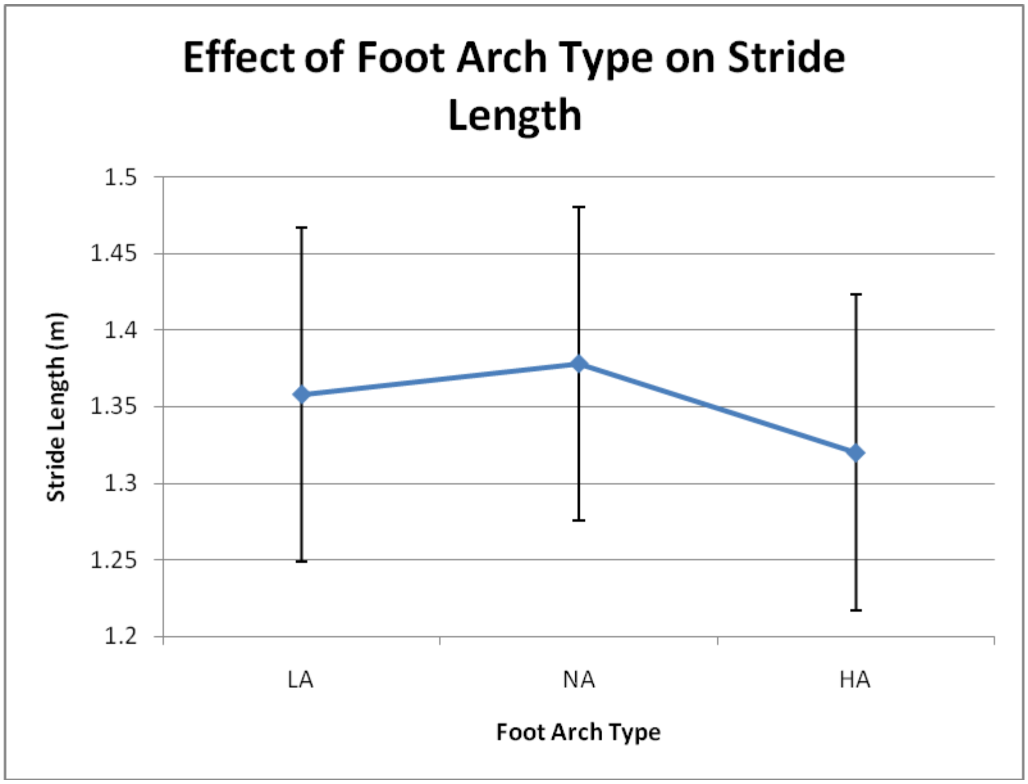


Figure 40. Effect of foot arch type on stride length, p = .223.

Eversion

A follow-up repeated measures ANOVA indicated a significant interaction effect of footwear and arch type on EV (F(2.863,77) = 4.403, p = .007, η^2= .105, *Power* = .851) (Figure 41). The interaction effect implies that different foot arch types produced different EV angles across each type of footwear. Means and standard deviations (SD) are presented in Table 6. To investigate the interaction, an independent samples t test was performed for each footwear condition and a paired samples t test was performed for each foot arch type.

The independent samples t test yielded a significant effect of arch type on eversion for all footwear conditions. The LA individuals had statistically less EV than NA individulas for BF (t(65) = 2.293, p = .025), FF1 (t(65) = 2.146, p = .036) and FF2

($t(65) = 2.343$, $p = .022$). The LA individuals also had statistically less EV than HA individuals for BF ($t(23) = 2.579$, $p = .017$), FF1 ($t(23) = 2.341$, $p = .028$) and FF2 ($t(23) = 2.305$, $p = .031$). There was no significant difference between NA and HA individulas for any of the footwear conditions; BF ($t(62) = 1.192$, $p = .238$), FF1 ($t(62) = .544$, $p = .589$) and FF2 ($t(62) = .043$, $p = .966$). These findings imply that LA individuals have less EV when compared to NA and HA individulas.

The paired samples *t* test yielded no significant difference between any of the footwear conditions for LA individuals; BF and FF1 ($t(12) = -.704$, $p = .495$), BF and FF2 ($t(12) = -.884$, $p = .394$), and FF1 and FF2 ($t(12) = -.354$, $p = .729$). There was also no significant difference between FF1 and FF2 for NA ($t(52) = 1.551$, $p = .127$) or HA ($t(10) = -1.391$, $p = .194$) individuals. There was a significant difference between BF and both flip-flop conditions (FF1 and FF2) in NA ($t(52) = -3.983$, $p < .001$ and $t(52) = -2.502$, $p = .016$ respectively) and HA individuals ($t(10) = -3.111$, $p = .011$ and $t(10) = -2.854$, $p = .017$ respectively). These findings imply that footwear has no effect on LA individuals, and that walking barefoot is the same as walking in flip-flops for LA individuals. However, flip-flops limit eversion in NA and HA individuals, but there is no difference between flip-flops.

Table 6
Means and standard deviations for eversion angles during midstance

	LA		NA		HA	
	M	*SD*	*M*	*SD*	*M*	*SD*
BF	-1.7259	2.3508	-3.9999	1.7759	-5.4741	1.7808
FF1	-1.5309	3.4970	-3.3892	3.0966	-3.9541	3.1652
FF2	-1.4814	4.7688	-3.5519	3.3282	-3.5964	2.7926

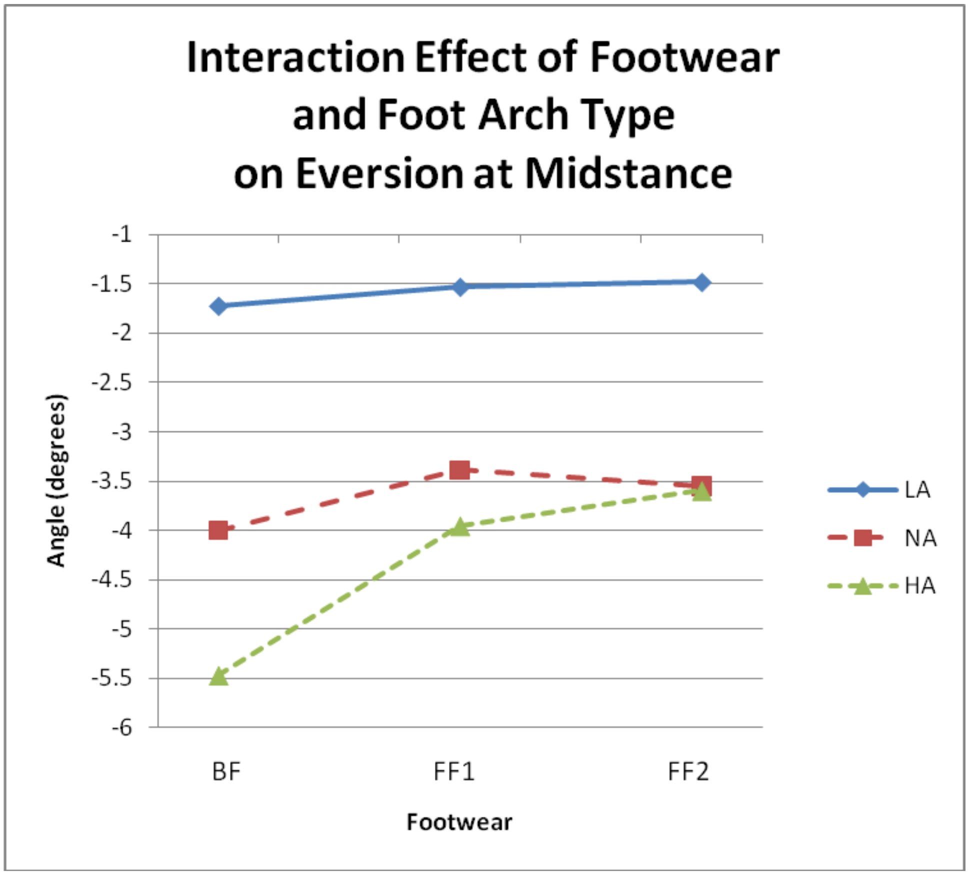


Figure 41. Interaction effect of footwear and foot arch type on eversion during midstance, p = .007.

There was a simple main effect of footwear on EV (F(1.431,77) = 16.824, p < .001, η^2= .183, *Power* = .997). A post hoc test using the LSD criterion yielded a significant difference between BF and FF1 (p < .001) and BF and FF2 (p < .001). There was no significant difference between flip-flop conditions, FF1 and FF2 (p = .447) (Figure 42). Based on the OFM output, a smaller number means the ankle has greater eversion than a larger number (the angle between the calcaneus and the tibia is less); therefore, these data suggest that ankle eversion is greatest in the BF condition with less eversion seen in FF1 and FF2 conditions. These results also imply that during the midstance phase, the ankle everts the same amount in FF1 as in FF2.

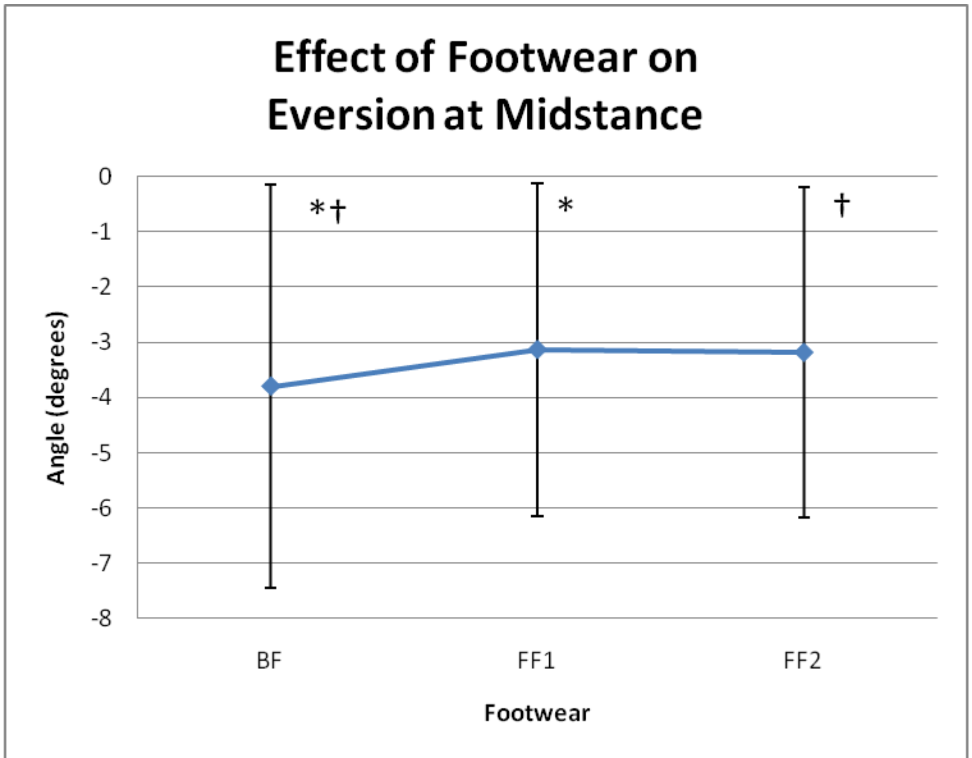


Figure 42. Effect of footwear on eversion during midstance. *† $p < .001$.

There was also a significant simple main effect of arch type on EV ($F(2,77) = 3.139$, $p = .049$, $\eta^2 = .077$, *Power* = .586). A post hoc test using the LSD yielded a significant difference between individuals with LA and NA ($p = .028$) and LA and HA ($p = .029$). This statistical significance suggests that individuals with HA and individuals with NA have greater eversion angles than individuals with LA. There was no significant difference between NA and HA ($p = .498$) (Figure 43).

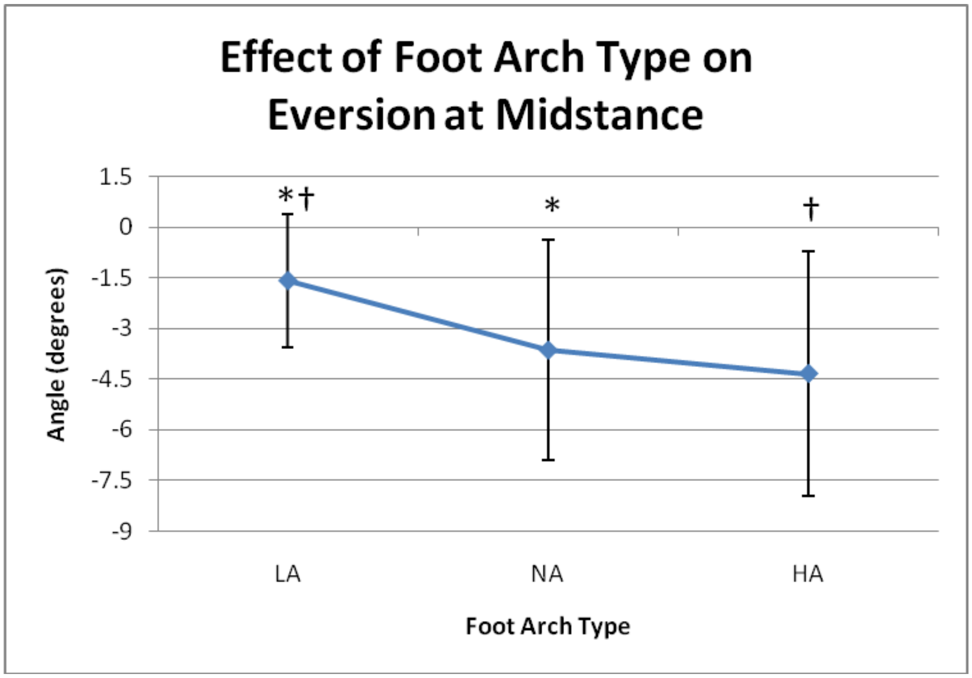


Figure 43. Effect of foot arch type on eversion during midstance.
$^{*}p$ = .028. $^{\dagger}p$ = .029.

Peak Eversion

A follow-up repeated measures ANOVA indicated a significant interaction effect of footwear and arch type on EV_{PEAK} (F(3.383,77) = 2.856, p = .034, η^2= .071, *Power* = .710) (Figure 44). The interaction effect indicates that different foot arch types demonstrated different EV_{PEAK} across each type of footwear. Means and standard deviations (SD) are presented in Table 7. To investigate the interaction, an independent samples t test was performed for each footwear condition and a paired samples t test was performed for each foot arch type.

The independent samples t test yielded a significant effect of arch type on peak eversion for all footwear conditions. The LA individuals had less EV than NA individulas for BF (t(33.343) = 2.837, p = .008), FF1 (t(41.192) = 2.455, p = .018) and

FF2 ($t(34.505) = 2.528$, $p = .016$). The LA individuals also had less EV than HA individuals for BF ($t(23) = 2.317$, $p = .030$) and FF1 ($t(23) = 2.258$, $p = .034$); howeverthere was no difference for FF2 ($t(23) = 1.932$, $p = .066$). There was no significant difference between NA and HA individulas for any of the footwear conditions; BF ($t(62) = 1.026$, $p = .309$), FF1 ($t(62) = 1.017$, $p = .313$) and FF2 ($t(62) = .197$, $p = .844$). These findings imply that LA individuals have less EV_{PEAK} when compared to NA and HA individulas for BF and FF1, but not for FF2. These findings also show that there was no fifference between NA and HA individuals in EV_{PEAK} for any of the footwear conditions.

The paired samples *t* test yielded a significant difference between BF and FF1 for all arch types; LA ($t(12) = -2.255$, $p = .044$), NA ($t(52) = -6.135$, $p < .001$), HA ($t(10) = -3.438$, $p = .006$). There was also a significant difference between BF and FF2 for all arch types; LA ($t(12) = -2.548$, $p = .026$), NA ($t(52) = -5.794$, $p < .001$), HA ($t(10) = -3.163$, $p = .010$). There was no significant diffrenece between the flip-flop conditions (FF1 and FF2) for any of the arch types; LA ($t(12) = -.366$, $p = .721$), NA ($t(52) = .253$, $p = .802$), HA ($t(10) = -1.945$, $p = .080$). These findings imply that EV_{PEAK} is increased in flip-flop conditions when compared to BF for all foot arch types; however, there is no differenec bewteen flip-flop types on EV_{PEAK}.

Table 7

Means and standard deviations for peak eversion angles

	LA		NA		HA	
	M	*SD*	*M*	*SD*	*M*	*SD*
BF	-4.5516	2.7618	-7.0753	4.4770	-8.9217	6.3625
FF1	-3.7909	2.2093	-5.8325	4.2634	-7.3334	5.3396
FF2	-3.6919	2.4541	-5.8766	4.0965	-6.1420	3.8672

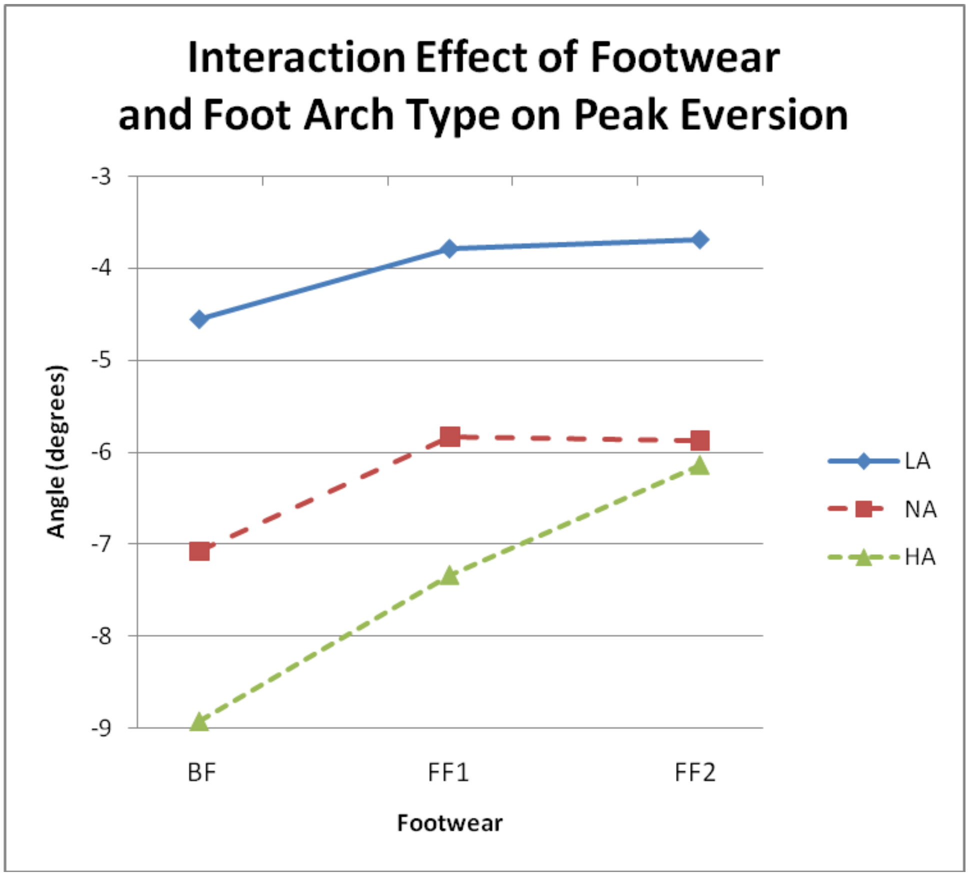


Figure 44. Interaction effect of footwear and foot arch type on peak eversion, p = .034.

There was a simple main effect of footwear on EV_{PEAK} (F(1.828,77) = 30.994, p < .001, η^2= .292, *Power* = 1.000). A post hoc test using the LSD yielded a significant difference in EV_{PEAK} between each footwear condition: BF and FF1 (p < .001) and BF and FF2 (p < .001), and FF1 and FF2 (p = .031) (Figure 45). Results imply that the type of footwear affects EV_{PEAK} as BF resulted in greater eversion angles than FF1 and FF2 with the smallest eversion angles occurring in FF2.

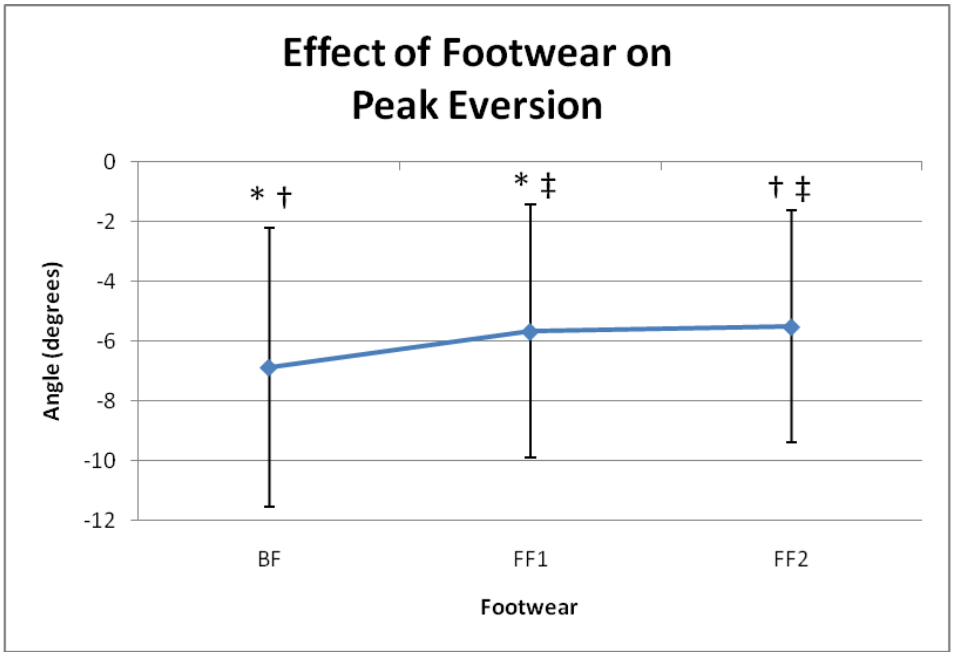


Figure 45. Effect of footwear on peak eversion.
$^{*\,\dagger}p < .001$. $^{\ddagger}p = .031$.

There was no simple main effect of foot arch type on EV_{PEAK} ($F(2,77) = 2.462$, $p = .092$, $\eta^2 = .062$, *Power* = .481). Implying that EV_{PEAK} was no different between individuals with LA, NA, and HA (Figure 46).

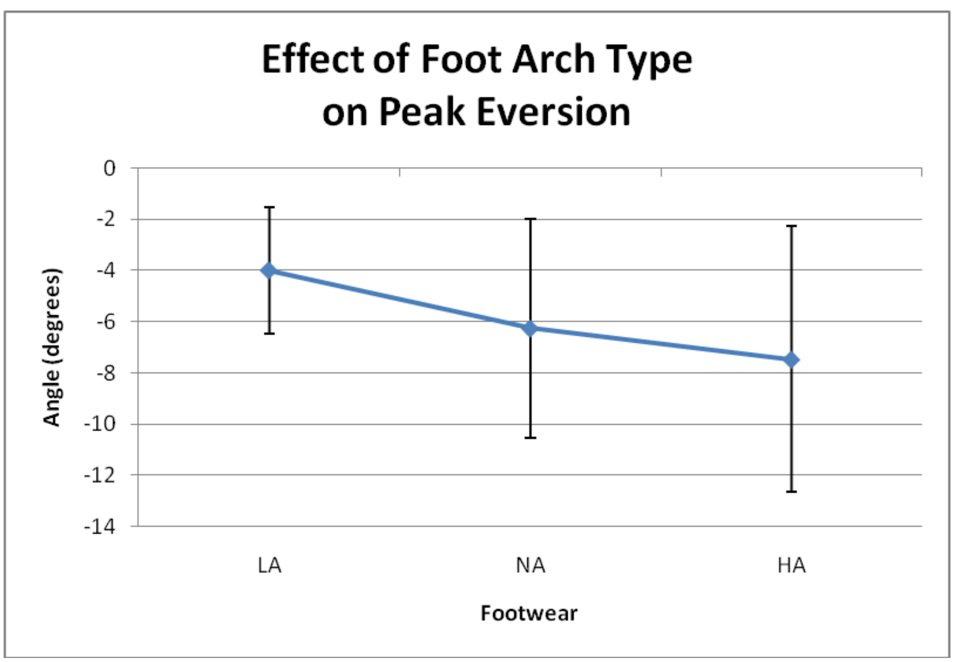


Figure 46. Effect of foot arch type on peak eversion, $p = .092$.

Pronation

A follow-up repeated measures ANOVA indicated no significant interaction effect of footwear and arch type on PRO (F(3.510,77) = 2.344, p = .066, η^2= .059, *Power* = .625), (Figure 47). There was also no significant main effect of footwear on PRO (F(1.755,77) = .755, p = .456, η^2= .010, *Power* = .168), (Figure 48).

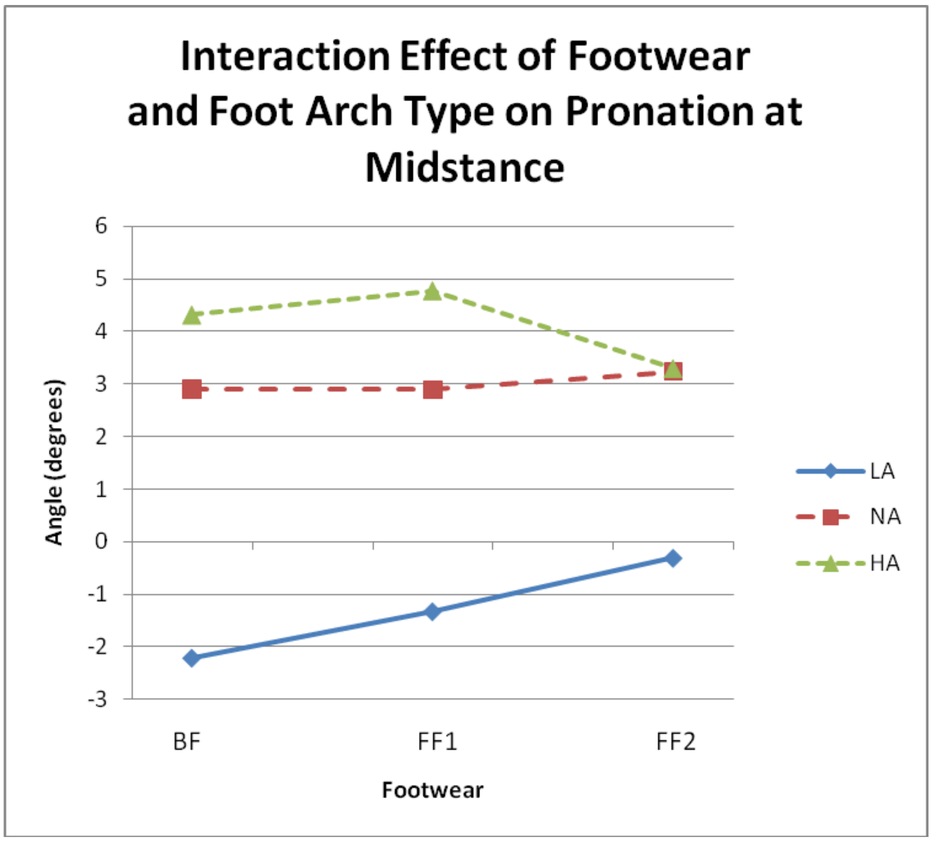


Figure 47. Interaction effect of footwear and foot arch type on pronation at midstance, p = .066.

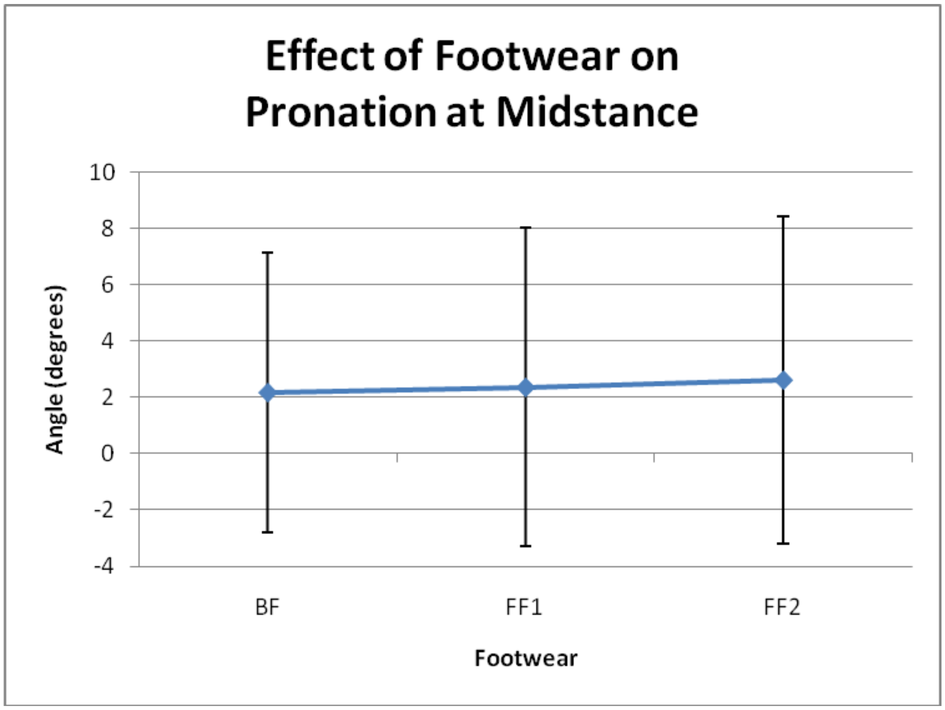


Figure 48. Effect of footwear on pronation during midstance, p = .456.

There was a significant main effect of foot arch type on PRO (F(2,77) = 4.249, p = .018, η^2= .102, *Power* = .727). A post hoc test using the LSD criterion yielded a significant difference between individuals with LA and NA (p = .009), and LA and HA (p = .014). There was no significant difference between individuals with NA and HA (p = .519) (Figure 49). Based on the OFM, a smaller value means greater pronation; therefore, individuals with a LA exhibit greater degrees of pronation during the midstance phase when compared to individuals with NA and HA. There was no difference in the degree of pronation during midstance for individuals with NA and HA.

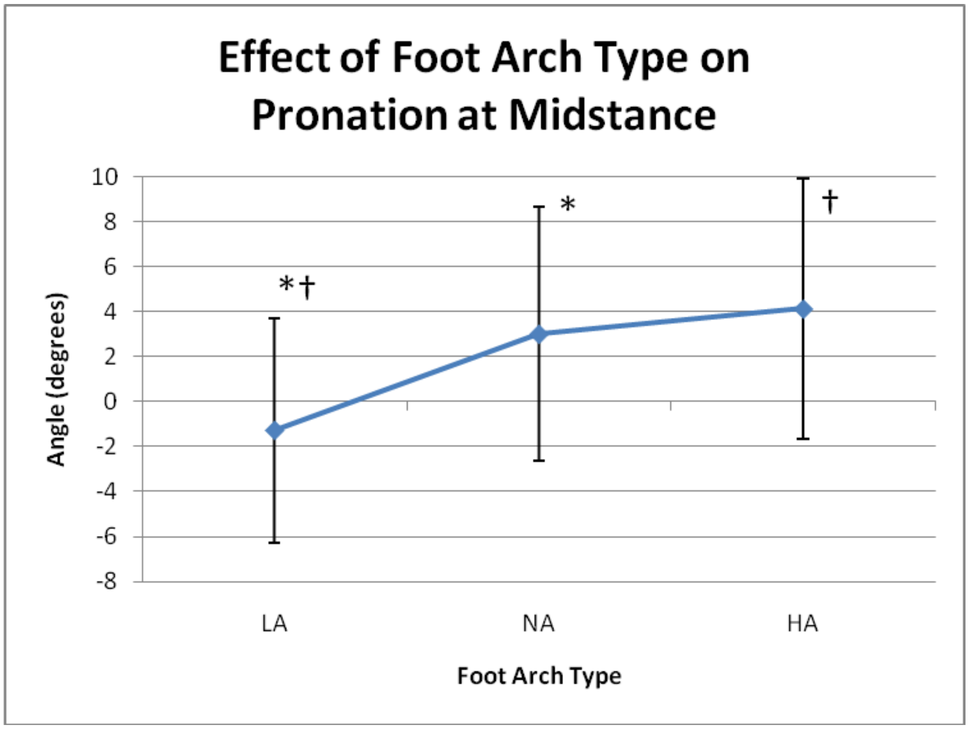


Figure 49. Effect of foot arch type on pronation during midstance.
$^{*}p$ = .009. $^{\dagger}p$ = .014.

Peak Pronation

A follow-up repeated measures ANOVA indicated no significant interaction effect of footwear and arch type on PRO_{PEAK} (F(3.495,77) = 1.490, p = .215, η^2= .038, *Power* = .420) (Figure 50). Nor was there a significant main effect of footwear on PRO_{PEAK} (F(1.747,77) = .519, p = .572, η^2= .007, *Power* = .129) (Figure 51).

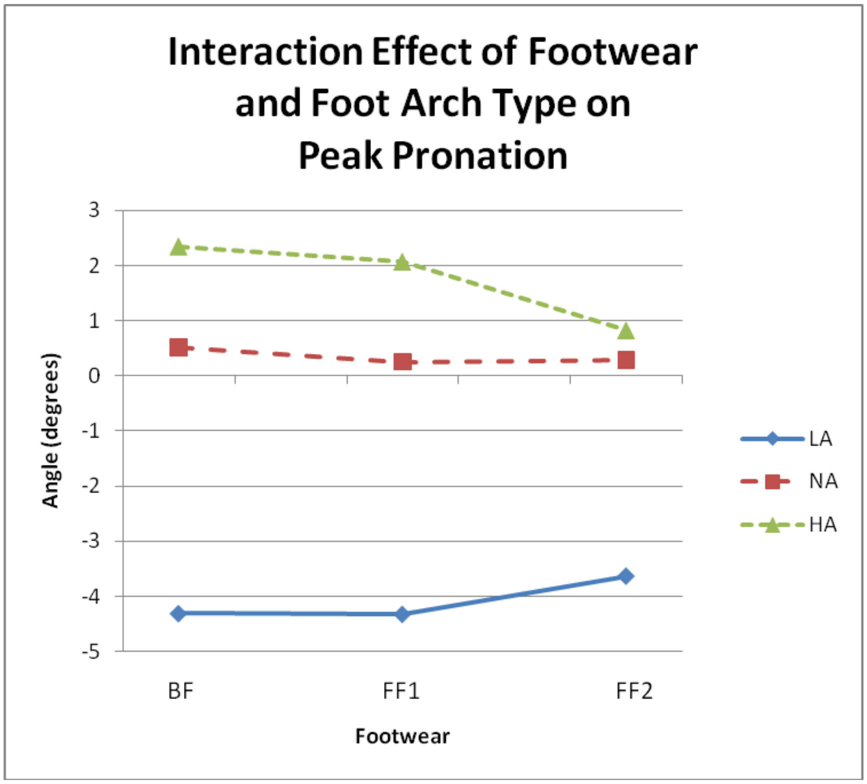


Figure 50. Interaction effect of footwear and foot arch type on peak pronation, *p* = .215.

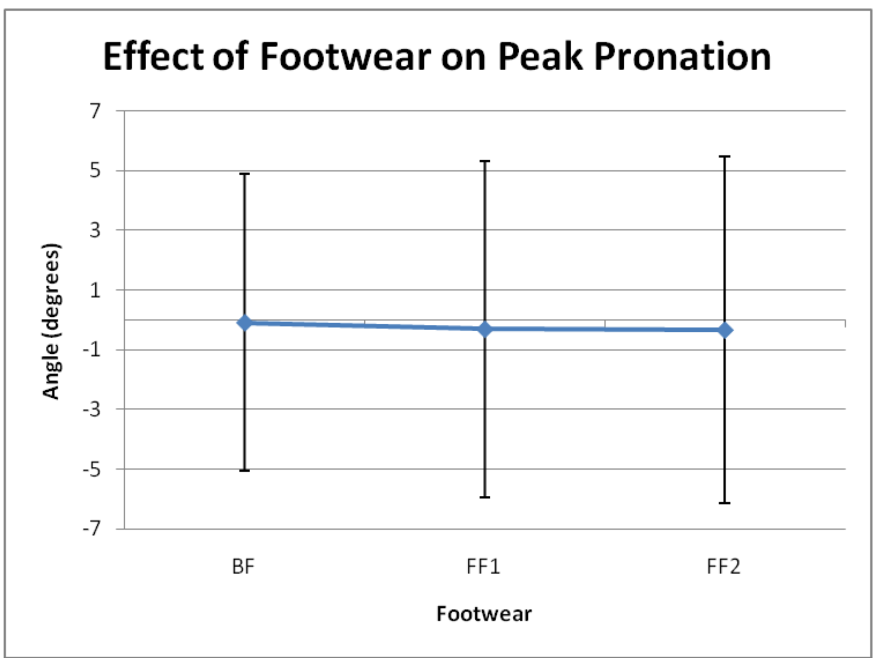


Figure 51. Effect of footwear on peak pronation, *p* = .572.

There was a significant main effect of foot arch type on PRO_{PEAK} ($F(2,77) = 5.030$, $p = .009$, $\eta^2 = .118$, *Power* = .801). A post hoc test using the LSD criterion yielded a significant difference between individuals with LA and NA ($p = .006$), and LA and HA ($p = .007$). There was no significant difference between individuals with NA and HA ($p = .420$) (Figure 52). Based on the OFM, a smaller value means greater pronation; therefore, individuals with a LA exhibit greater degrees of peak pronation when compared to individuals with NA and HA. There is no difference in the degree of peak pronation for individuals with NA and HA.

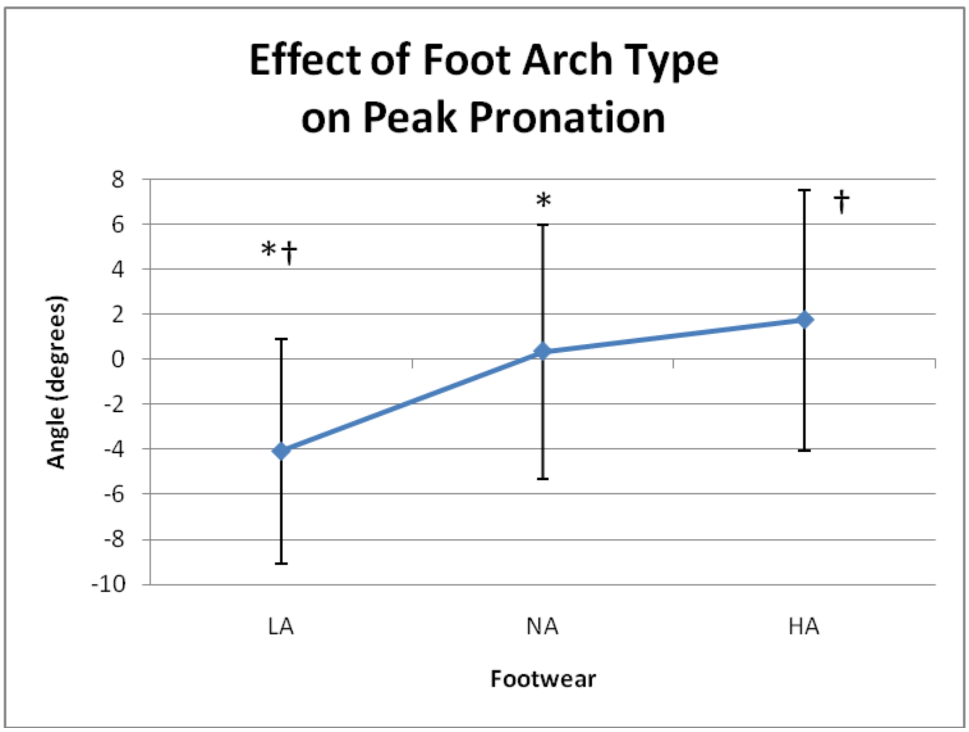


Figure 52. Effect of foot arch type on peak pronation.
$^*p = .006$. $^†p = .007$.

Dorsiflexion

A follow-up repeated measures ANOVA indicated a significant interaction effect of footwear and arch type on DORSI ($F(3.383,77) = 2.856$, $p = .034$, $\eta^2 = .071$, *Power* = .710) (Figure 53). The interaction effect indicates that the different foot arch types behaved differently across each type of footwear. Means and standard deviations (SD) are presented in Table 8. To investigate the interaction, an independent samples *t* test was performed for each footwear condition and a paired samples *t* test was performed for each foot arch type.

The independent samples *t* test yielded no significant effect of arch type on $DORSI_{PEAK}$ for all footwear conditions. The non significance between LA and NA individuals for each footwear condition was BF ($t(65) = -.915$, $p = .358$), FF1 ($t(65) = -1.041$, $p = .302$), and FF2 ($t(65) = -.939$, $p = .351$). The non significance between LA and HA individuals for each footwear condition was BF ($t(23) = -1.250$, $p = .224$), FF1 ($t(23) = -.891$, $p = .382$), and FF2 ($t(22.295) = -.048$, $p = .962$). The non significance between NA and HA individuals for each footwear condition was BF ($t(62) = -.273$, $p = .786$), FF1 ($t(62) = .195$, $p = .846$), and FF2 ($t(62) = .828$, $p = .411$). These findings show that there was no difference between individuals with any type fo foot arch on $DORSI_{PEAK}$ for any of the footwear conditions.

The paired samples *t* test yielded a significant difference between BF and FF1 for all arch types; LA ($t(12) = 3.333$, $p = .006$), NA ($t(52) = 4.728$, $p < .001$), HA ($t(10) = 4.418$, $p = .001$). There was also a significant difference in $DORSI_{PEAK}$ between BF and FF2 for all arch types; LA ($t(12) = 2.294$, $p = .041$), NA ($t(52) = 4.233$, $p < .001$), HA ($t(10) = 6.403$, $p < .001$). Dorsiflexion was larger in the BF conditions compared to the flip-flop conditions. There was no significant diffrenece between the flip-flop conditions (FF1 and FF2) for LA ($t(12) = -.510$, $p = .616$) or NA individuals ($t(52) = -.825$, $p = .413$); however, there was a significant difference bewteen FF1 and FF2 for HA individuals ($t(10) = 2.396$, $p = .038$). The FF2 condition resulted in a smaller $\text{DORSI}_{\text{PEAK}}$ in HA individuals compeared to NA individuals. These findings imply that $\text{DORSI}_{\text{PEAK}}$ is decreased in flip-flop conditions when compared to BF for all foot arch types; however, the only differenec bewteen the flip-flop conditions for $\text{DORSI}_{\text{PEAK}}$ occurs in HA individuals.

Table 8
Means and standard deviations for peak dorsiflexion

	LA		NA		HA	
	M	*SD*	*M*	*SD*	*M*	*SD*
BF	10.2916	3.5986	11.5962	4.9255	12.0198	3.1990
FF1	9.0312	3.2574	10.4696	4.8756	10.1704	3.0585
FF2	9.2479	3.5726	10.6015	5.0598	9.3048	2.3065

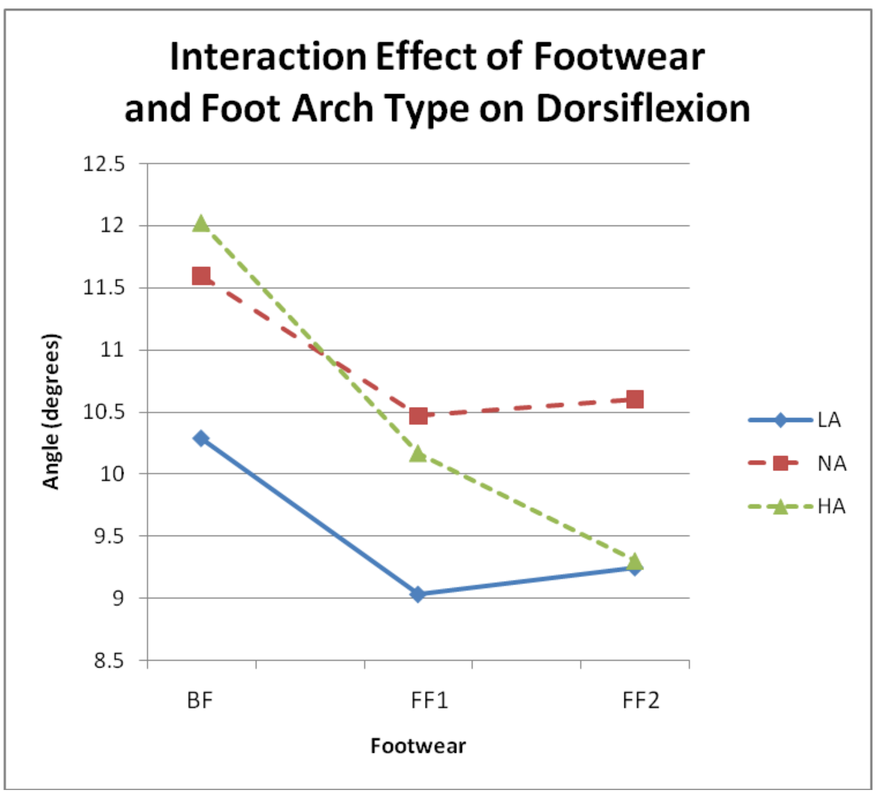


Figure 53. Interaction effect of footwear and foot arch type on dorsiflexion, $p = .034$.

There was a significant simple main effect of footwear on $DORSI_{PEAK}$ ($F(1.691,77) = 30.461$, $p < .001$, $\eta^2 = .289$, *Power* = 1.000). A post hoc test using the LSD criterion yielded a significant difference between BF and FF1 ($p < .001$), and BF and FF2 ($p < .001$). There was no significant difference between FF1 and FF2 ($p = .311$). In the OFM, a larger value means a greater dorsiflexion (or the angular distance between the tibia and the foot is smaller); therefore, the BF condition yielded the greatest dorsiflexion angle compared to both flip-flop conditions during the swing phase. Greater dorsiflexion is noted in the BF condition when compared to all flip-flops conditions (Figure 54).

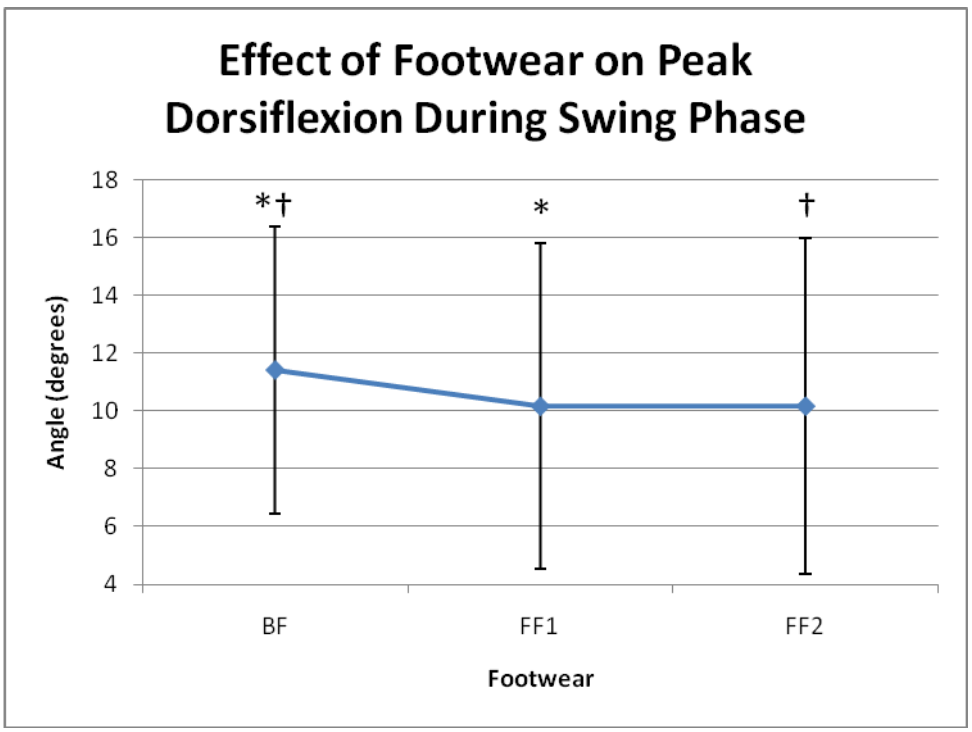


Figure 54. Effect of footwear on peak dorsiflexion during the swing phase.
$^{*\dagger}p < .001$.

There was no significant simple main effect of foot arch type on $DORSI_{PEAK}$ ($F(2,77) = .535$, $p = .588$, $\eta^2 = .014$, *Power* = .135). This suggests that a person's arch type does not affect peak dorsiflexion during the swing phase of gait (Figure 55).

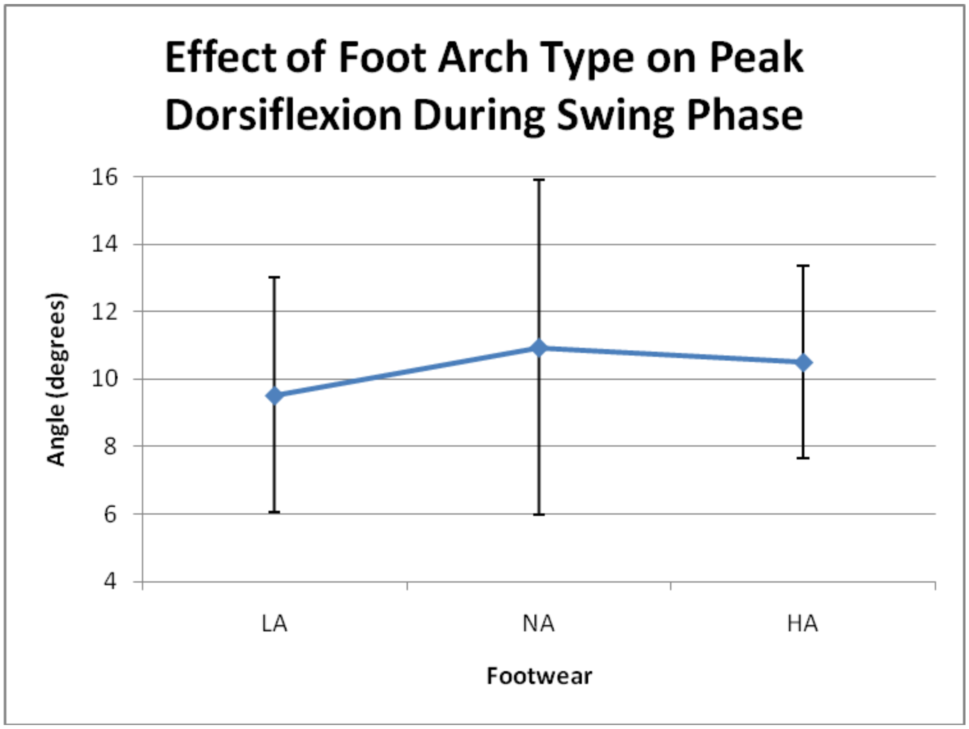


Figure 55. Effect of foot arch type on peak dorsiflexion during swing phase, $p = .588$.

Peak Hallux Extension

A follow-up repeated measures ANOVA indicated no significant interaction effect of footwear and arch type on HX_{PEAK} ($F(4,77) = .581$, $p = .677$, $\eta^2 = .015$, *Power* = .189), (Figure 56). There was no significant main effect of footwear on HX_{PEAK} ($F(2,77) = 1.012$, $p = .366$, $\eta^2 = .013$, *Power* = .224). There was also no significant main effect of foot arch type on HX_{PEAK} ($F(2,77) = .620$, $p = .541$, $\eta^2 = .016$, *Power* = .150). There is no difference in peak hallux extension after toe off between any of the footwear conditions nor foot arch types (Figures 57 and 58).

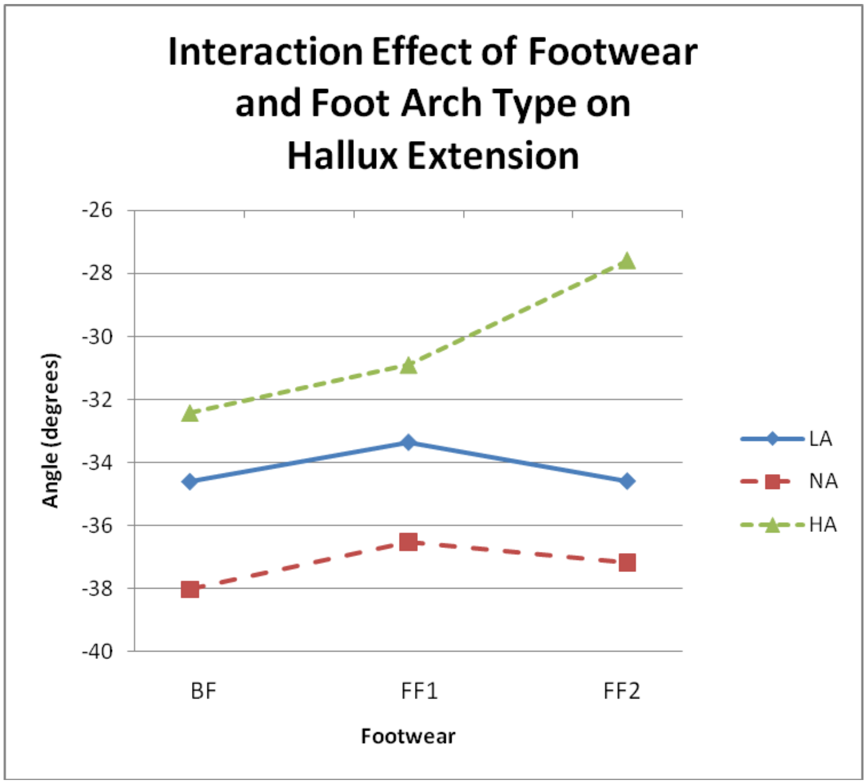


Figure 56. Interaction effect of footwear and foot arch type on peak hallux extension, p = .677.

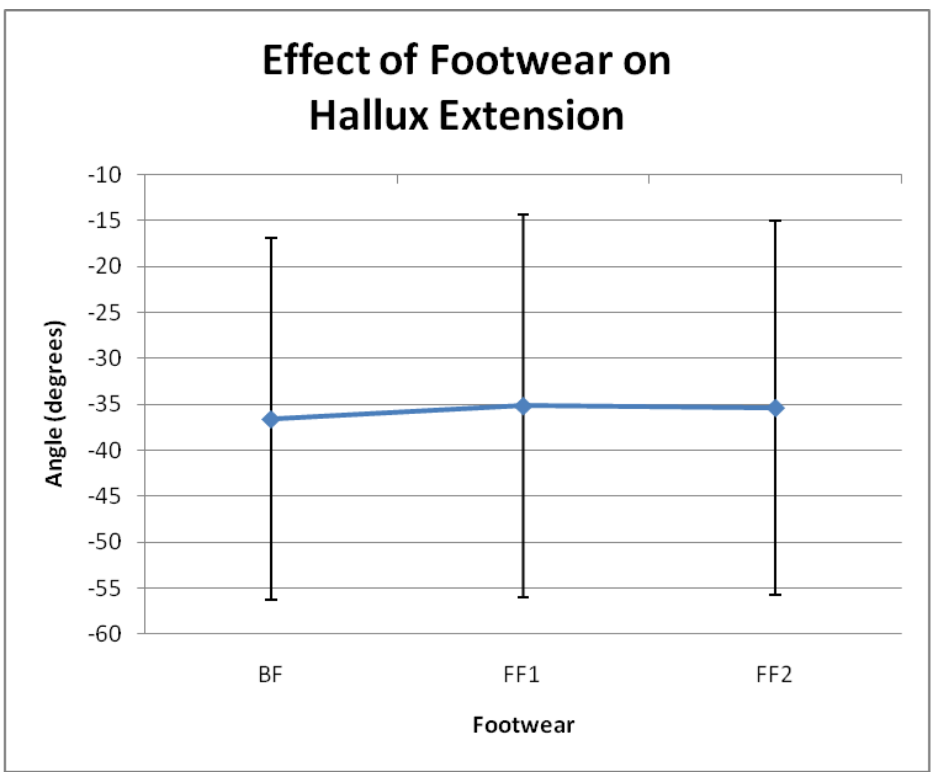


Figure 57. Effect of footwear on peak hallux extension, p = .366.

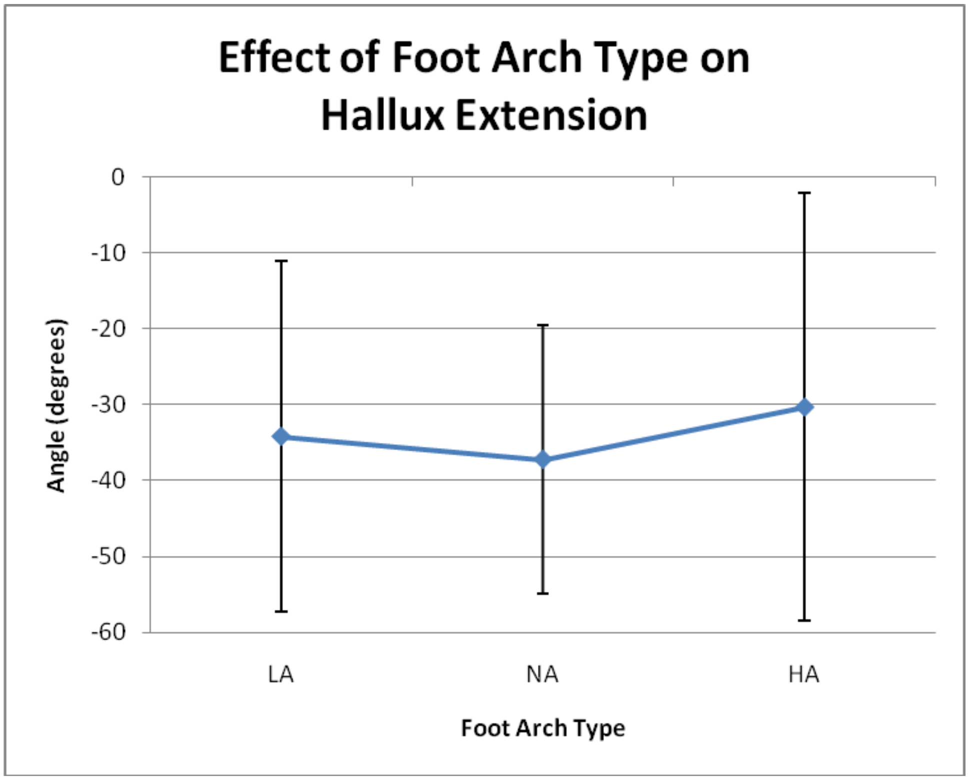


Figure 58. Effect of foot arch type on peak hallux extension, $p = .541$.

Section 4: Normal Arch and Surface Electromyography

The third research question proposed by the current study was to examine if there was an increase in muscular activity of the tibialis anterior (TA) during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA. This section includes the results for that research question. Fifty-three individuals of the sample population were classified as having a NA; however, for the sEMG variables, data for two participants was omitted due to erroneous data making the sample population $N = 51$. A representative graph of the raw sEMG data of the TA is in Appendix D. The multivariate repeated measures ANOVA with a *p* value set *apriori* at $<$ .05 yielded a significant effect of footwear on sEMG of the TA during the swing phase

(Wilks's Λ = .385, $F(6,45) = 11.957$, $p < .001$, $\eta^2 = .615$, *Power* = 1.000. Follow-up repeated measures ANOVA's showed a significant main effect of footwear on the average sEMG activity ($sEMG_{AVG}$) and peak sEMG ($sEMG_{PEAK}$) of the TA muscle during the swing phase (Table 9). The average sEMG was calculated as the average TA activity over the swing phase during the gait cycle in the non support leg. The peak sEMG was calculated as the maximum peak of the TA at any point during the swing phase of the gait cycle in the non support leg. Descriptive statistics are presented in Appendix C.

Table 9
Effect of footwear on $sEMG_{AVG}$ and $sEMG_{PEAK}$ for NA participants

	N	df	*F*	*p*	η^2	*Power*
$sEMG_{AVG}$	51	2.517	33.914	< .001	.404	1.000
$sEMG_{PEAK}$	51	3	6.537	< .001	.116	.969

Average sEMG

A post hoc test using the LSD criterion yielded a significant difference in $sEMG_{AVG}$ between BF and each of the flip-flop conditions: FF1 ($p < .001$), FF2 ($p < .001$), and FF3 ($p < .001$). However, there was no significant difference in $sEMG_{AVG}$ between FF1 and FF2 ($p = .292$), FF1 and FF3, ($p = .056$), or FF2 and FF3 ($p = .352$). The means and standard deviations are found in Figure 59. These data illustrate that $sEMG_{AVG}$ increased during the swing phase when wearing flip-flops compared to barefoot. The data also indicate that there is no significant difference in the activity of the TA between any of the flip-flop conditions. The increased mean activity of the TA indicates that the tibialis anterior provides a greater contribution to the dorsiflexion moment at the ankle during all the flip-flop conditions compared to the barefoot condition.

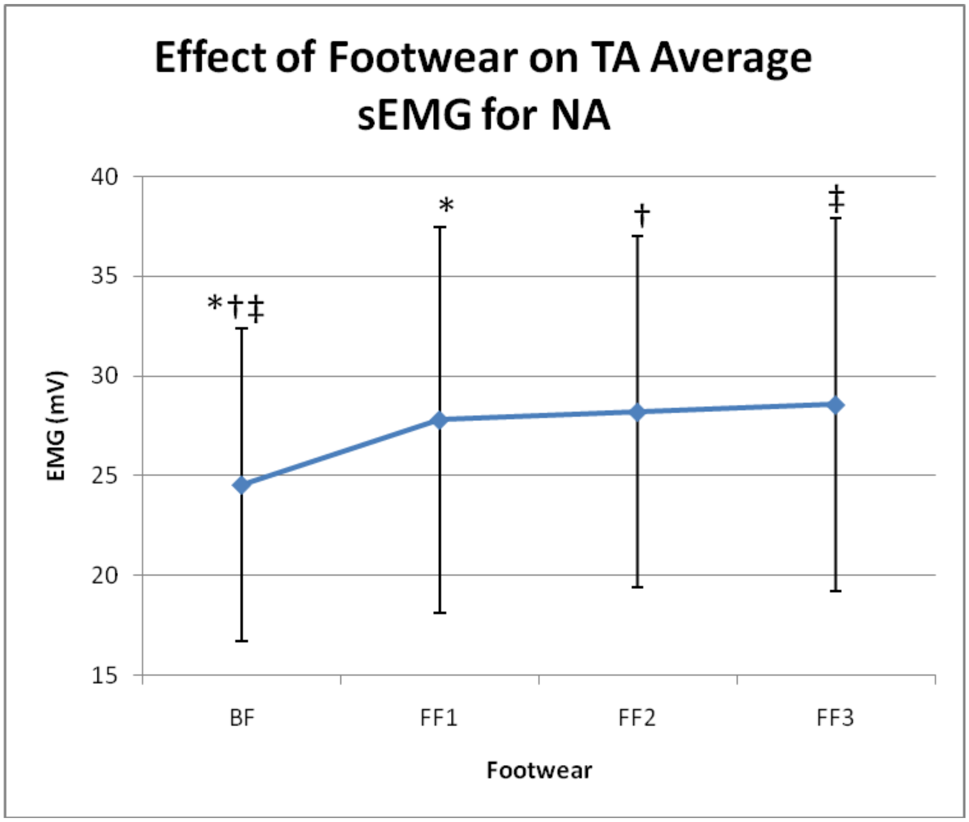


Figure 59. Effect of footwear on tibialis anterior average sEMG activity during the swing phase for NA participants.
$^{*\dagger\ddagger}p < .001$.

Peak sEMG

A post hoc test using the LSD criterion yielded a significant difference in $sEMG_{PEAK}$ between BF and all the flip-flop conditions: FF1 (p <.001), FF2 (p = .003), and FF3 (p = .001). There was no significant difference in $sEMG_{PEAK}$ between any of the flip-flop conditions: FF1 and FF2 (p = .393), FF2 and FF3 (p = .528), and FF1 and FF3

(p = .855) (Figure 60). These data illustrate that $sEMG_{PEAK}$ is increased during the swing phase when wearing flip-flops compared to barefoot, but there is no difference in the peak sEMG amplitude of the TA during the swing phase of gait between any flip-flop condition. This implies that the contribution of the peak TA activity to the dorsiflexion moment at the ankle is greater when wearing flip-flops than BF, and there is no significant difference in the contribution of the TA to the dorsiflexion moment between different types of flip-flops.

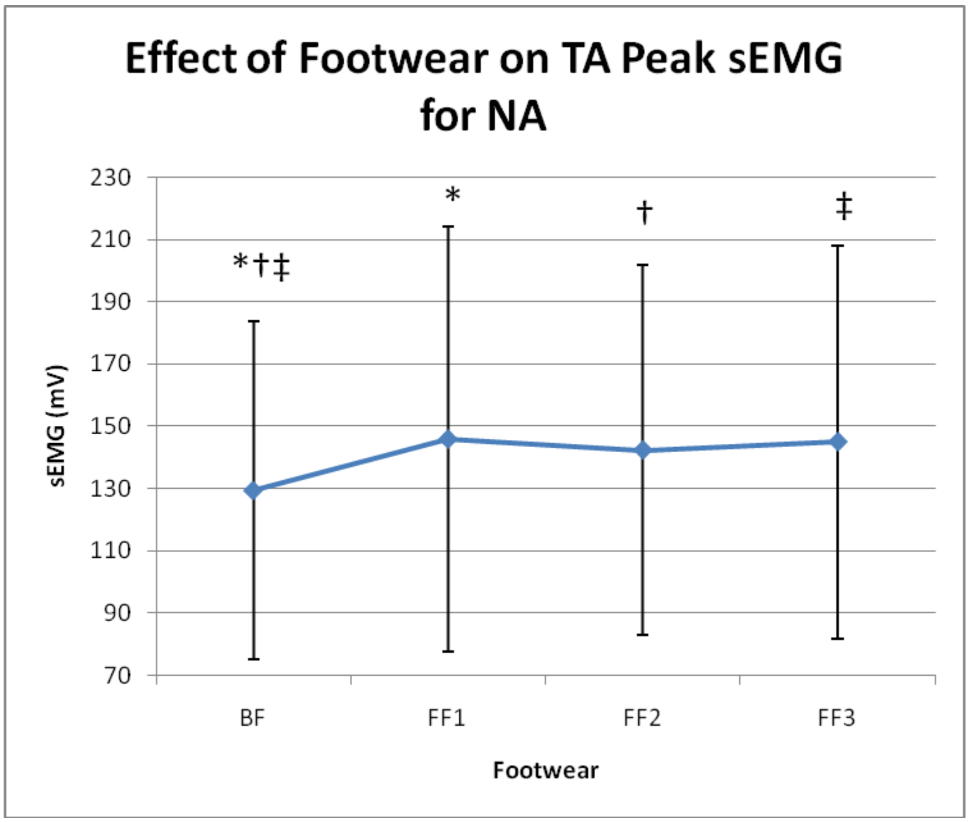


Figure 60. Effect of footwear on tibialis anterior peak sEMG amplitude during the swing phase for NA participants.
$^{*}p < .001$. $^{\dagger}p = .003$. $^{\ddagger}p = .001$.

CHAPTER V. DISCUSSION

The purpose of this investigation was three-fold: (1) to examine the effects that different components (i.e. arch support, midtarsal support, toe ridge, and longer/thicker straps) of the thong style flip-flop have on gait kinematics in individuals classified with normal arched (NA) feet; (2) to investigate the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified with either low (LA), normal (NA), or high arched (HA) feet; and (3) to determine if there is an increase in muscular activity of the tibialis anterior (TA) during the swing phase of gait when wearing thong style flip-flops in individuals classified with NA. This chapter is divided in to 5 sections. Each of the first 3 sections addresses one of the research questions of the overall study and the findings for each research question individually. Section 1 discusses the effect of footwear on gait kinematics in individuals classified with NA. Section 2 discusses the effect of footwear arch on gait kinematics in individuals classified with LA, NA, and HA. The author wanted to specifically investigate the effect or the footwear arch support on individuals with difference foot arch types; therefore, the second research question was designed specifically for that research question. However, the arch support was part of the first research question and consequently the first and second research questions have similar outcomes. It should be noted that there are overlapping findings between section 1 and section 2 and as such, the same explanations for the outcomes mentioned in section 1 may be presented again in section 2. In addition, there are conflicting findings between section 1 and 2, so some explanations may be contrary, however Section 4 will provide the synthesis of all of the findings. Section 3 discusses the effect of footwear on TA surface electromyography (sEMG) in individuals classified with NA. Section 4 provides a summary of the findings combining the results of all three research questions. Section 5 includes final conclusions and recommendations for future research.

Section1: Footwear and Gait Kinematics

The purpose of this portion of the current study was to investigate the effects of different components (i.e. arch support, midtarsal support, toe ridge, and longer/thicker straps) of the thong style flip-flop on gait kinematics in individuals classified with NA feet. The gait kinematics of interest included stride length (SL), eversion at midstance (EV), peak eversion during the stance phase (EV_{PEAK}), pronation at midstance (PRO), peak pronation during the stance phase (PRO_{PEAK}), peak dorsiflexion during swing phase ($DORSI_{PEAK}$), and peak hallux extension in the beginning swing phase (HX_{PEAK}).

For purposes of discussion in this section, the kinematics observed in the BF condition will be considered a "normal gait." This is based on the assumption that gait kinematics in the barefoot condition are how the ankle and foot behave without any other factor to influence the foot and ankle kinematics. If any of the kinematic variables differ from the BF condition, this could present potential problems because the foot kinematics have been altered by the introduction of the flip-flop in such a manner that structural properties of the foot are not functioning as it would without the flip-flop (i.e. BF). Table 10 illustrates all flip-flop conditions compared to barefoot for each kinematic variable measure. Table 11 illustrates the flip-flop conditions compared to each other.

Table 10
Effect of flip-flops compared to barefoot on all variables in NA individuals

Kinematic variable	FF1	FF2	FF3
SL	↑	↑	↑
EV	↓	↓	↔
EV_{PEAK}	↓	↓	↓
PRO	↔	↔	↔
PRO_{PEAK}	↔	↔	↔
$DORSI_{PEAK}$	↓	↓	↔
HX_{PEAK}	↔	↔	↔

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

Table 11
Comparison of flip-flops on all variables in NA individuals

Kinematic variable	Arch Support (FF2)[*]	Arch Support with Addition Features (FF3)[*]	Higher Arch Support with Addition Features (FF3)[†]
SL	↓	↔	↔
EV	↔	↑	↑
EV_{PEAK}	↔	↑	↑
PRO	↔	↔	↔
PRO_{PEAK}	↔	↔	↔
$DORSI_{PEAK}$	↔	↑	↑
HX_{PEAK}	↔	↔	↔

Note. ↔ = no difference. ↑ = increase. ↓ = decrease. [*]compared to FF1. [†]compared to FF2

Stride Length

There have been several studies that have identified the influence of footwear on stride length (Mundermann et al., 2003; Oeffinger et al., 1999; J. F. Shroyer & Weimar, *In Press;* Wolf et al., 2008). Oeffinger et al. (1999) and Wolf et al. (2008) found that stride length increases when wearing footwear versus walking barefoot (Oeffinger et al., 1999; Wolf et al., 2008). It is suggested that different types of footwear alter stride length and research shows that lighter footwear will result in shorter stride lengths when compared to heavier footwear (Mundermann et al., 2003). A more recent study investigated the effects of flip-flops on gait kinematics and found that when compared to athletic shoes, flip-flops resulted in decreased stride length (J. F. Shroyer & Weimar, *In Press*). Researchers postulate that a direct correlation between the decreased mass of the slipper/flip-flop causes decreased inertia of the distal segment (the foot) during the swing phase of gait. An increased inertia imposed by the heavier footwear is thought to carry the foot further forward (Mundermann et al., 2003; Oeffinger et al., 1999; J. F. Shroyer & Weimar, *In Press*).

In the present study, there was a significant main effect of footwear on SL. Stride length was shorter in BF compared to all flip-flop conditions (Figure 31). This shorter SL when BF suggests that walking BF is not similar to walking in flip-flops with regards to SL (Table 10). Mean differences between BF and the flip-flop conditions were: FF1 (.05205 m), FF2 (.04186 m), and FF3 (.05111 m) (Appendix A). Based on the smaller SL in BF when compared to all flip-flop conditions, the findings of this current study are congruent with the previous research by Mundermann et al. (2003). However, for the flip-flop conditions there was a significant difference between FF1 and FF2. Flip-flop 2 resulted in a shorter SL than FF1, which contradicts previous literature as the average mass for sizes 6 -10 for each flip-flop type were, FF1 (92.1 g), FF2 (95.6 g), and FF3 (142.2 g). This does not support the findings that the increased SL is due to an increase in inertia of the foot from the increased mass of the footwear; FF2 was heavier than FF1, yet FF2 yielded a shorter SL than FF1. In addition, FF3 had an averaged mass that was 50.1 g heavier than FF1, and there was no significant difference in stride length between these 2 conditions. While the findings of the present study support the inertia conclusion in order to explain the difference between the BF and flip-flop conditions, findings associated within the flip-flop conditions also suggest that something other than inertia offers a complete explanation for the change in SL noted across flip-flops in NA individuals. Since the only difference, other than mass, between FF1 and FF2 was arch support, it may be reasonable to conclude that arch support produced adaptations in the gait that caused the decreased SL in FF2 when compared to FF1. Based on the above findings, it is reasonable to suggest that arch support in FF2 may have caused the foot to slide laterally; however, this was not recorded. This sliding made the footwear “feel” less stable and caused individuals to put their foot down sooner than was seen in FF1. Further investigation is needed to understand the mechanism for the decreased SL in FF2 due to arch support, but it may be concluded that wearing flip-flops is not like walking in barefoot with regard to SL.

Eversion at Midstance

The eversion angle (EV) is defined as the relationship between the tibia and the hindfoot at the ankle in the frontal plane about the anterior/posterior axis (Figure 20b). Results showed a significant main effect of footwear on EV in NA individuals (Figure 32). There was no significant difference in EV between BF and FF3; nor FF1 and FF2, however BF and FF3 were both significantly different from FF1 and FF2 (Tables 10 and 11). These findings imply that flip-flops with no arch (FF1) and those with an arch support (FF2) influenced the position of the calcaneus in the same manner, and that FF3 (which has an arch support, heel cup, midtarsal support and thicker straps) resembles walking barefooted, with respect to EV. It was hypothesized that EV would be limited in FF3; however, the pronounced heel cup found only in FF3 may provide an explanation.

The heel cup provided a ridge on the medial and lateral side of the hindfoot that could have provided an altered calcaneal position to account for the increased EV in comparison to FF1 and FF2. Not only does the heel cup provide a possible explanation for why EV seen when wearing FF3 is different from FF1 and FF2, it may also explain why FF3 is similar to BF. It is possible that the ankle motion was similar in all flip-flop conditions; however, the heel cup in FF3 may have caused the foot to be everted the same magnitude as the difference observed between BF and FF1 and FF2. The eversion caused by the heel cup made it appear that FF3 caused the foot to evert the same as when walking BF. It should also be noted that from an anecdotal finding, it appeared that in BF and FF3 individuals rolled the foot more like a "normal gait." During heel strike of "normal gait," the center of pressure (CoP) should initially move laterally and then medially; contrastingly, individuals walking in FF1 and FF2 conditions did not follow the "normal gait" pattern. This anecdotal observation of "normal" versus "abnormal" gait by researchers supports why eversion was limited in FF1 and FF2 when compared to BF and FF3. Future investigations should evaluate ground reaction forces and CoP path during the stance phase to determine if this above anecdotal observation is correct. In conclusion, the EV of individuals during the FF3 condition was the same as the EV in BF; whereas FF1 and FF2 cause smaller EV.

Peak Eversion

Peak eversion (EV_{PEAK}) occurred between heel off and toe off during the stance phase. Eversion is defined as the relationship between the tibia and the hindfoot at the ankle in the frontal plane about the anterior/posterior axis (Figure 20b). When wearing flip-flops, there is a period of time in which the heel of the foot comes off the flip-flop, but the flip-flop remains in contact with the ground. It is during this slice of time that EV_{PEAK} occurred and was one reason why this current study investigated not only peak eversion, but also eversion at midstance. At midstance, the foot is in full contact with the flip-flop. When EV_{PEAK} occurred, the foot was not in complete contact with the flip-flop. This observed foot to flip-flop separation is unique to flip-flop wear and it was hypothesized that critical information could be gained from investigating this period.

The results showed a significant main effect of footwear on EV_{PEAK} in NA individuals (Figure 33). Peak eversion in BF was significantly greater than all flip-flop conditions, suggesting that flip-flops limit ankle eversion (Table 10). It is possible that the contact of the mid and forefoot with the flip-flop had some effect on motion at the ankle by limiting eversion. An explanation could be that the Y-strap, common to all thong style flip-flops that runs from between the hallux and the second phalange to both the medial and lateral side of the foot limits the eversion of the hindfoot. The Y-strap runs across the midfoot and hindfoot and may prevent the hindfoot from moving more laterally than when walking BF therefore limiting the amount of possible eversion. Flip-flop 3 resulted in statistically significant more EV_{PEAK} when compared to FF1 and FF2, but unlike EV, FF3 was significantly different from the BF condition (Tables 10 and 11). These inter-flip-flop findings imply that structural differences of FF3 caused the ankle to evert more than in the FF1 and FF2 conditions. The influence of arch support may be ruled out because EV_{PEAK} was not different between FF1 and FF2. In addition, the heel cup may be ruled out as a potential influence as the foot is away from the heel during this interval. Therefore, it seems that the architectural components that most likely contributed to the difference were the midtarsal support and toe ridgebar which were present in FF3 only. Although, the heel cup may also contribute in that it position the foot prior to separation in such a manner to affect EV_{PEAK}. It is hypothesized that the midtarsal support could have limited the inversion as the CoP moved more medially. The toe ridgebar may also provide an increase gripping of the toes, in particular the great toe, in both the saggital and transverse planes. It is this toe gripping that may have rotated the foot more medially about the long axis of the foot and produced an apparent increased eversion angle for the FF3 condition. As with EV, the effect of limited EV_{PEAK} in flip-flop conditions could be the manner in which the individuals manipulate CoP on the foot. The flip-flop is not affixed securely to the foot, so individuals may adapt a safety mechanism to prevent excessive medial shifts in the CoP. Preventing the medial CoP shifts results in a less dramatic "roll the foot" when wearing flip-flops than that seen during the BF condition. The explanation is plausible but future research needs to incorporate ground reaction forces and comparisons of CoP path to support or discredit this supposition.

Pronation at Midstance

Pronation, for the purposes of this study, was defined as the relationship between the forefoot relative to the hindfoot about the long axis of the forefoot (Figure 21). It was hypothesized that the heel cup of FF3 would provide additional arch support for the medial arch of the foot and limit pronation of the foot when compared to BF, FF1, and FF2. Results were contrary to the hypothesized findings. There was no significant main effect of footwear on pronation during the midstance phase of the gait cycle in NA individuals (Figure 34 and Table 11). Results suggest that walking in flip-flops is like walking barefoot, with regard to pronation at midstance (Table 10). It was hypothesized that the external arch support provided by the architecture of FF2 and FF3 would support the medial arch of the foot and resist pronation of the foot when compared to the FF1 and BF condition. One explanation stems from the lack of adherence of the foot to the flip-flop and may provide insight into the contrary results. Since the foot is not affixed to the footwear, the foot may move laterally away from the medial arch support of FF2 and FF3. Presumably, these arch supports are present to prevent excessive pronation of the foot, however, in light of the current PRO findings, appear to be ineffective. Secondly, the arch of NA individuals may have functioned to dissipate forces experienced during stance phase. To achieve this dissipation of forces the medial arch of the foot collapses to some degree. However, due to the rigidity of the medial arch in NA individuals, the arch may not have collapsed enough to require the additional support provided by the arch support of the flip-flop. In conclusion, NA individuals did not need the external arch support provided by the footwear and therefore there was no effect of footwear on NA individuals.

Peak Pronation

Peak pronation (PRO_{PEAK}) occurred between heel off and toe off during the stance phase of the support leg. Pronation was defined as the relationship between the forefoot relative to the hindfoot about the long axis of the forefoot (Figure 21). One unique feature of the flip-flop is that there is a period of time when the heel of the foot comes off the flip-flop, but the flip-flop remains in contact with the ground. This time span is when peak pronation occurred and was one reason why this current study investigated average pronation at midstance as well as peak pronation.

It was hypothesized that PRO_{PEAK} would be related to a medial roll of the forefoot as the foot pivots about the Y-strap that is located between the hallux and second phalange of the foot. Additionally, it was hypothesized that the arch support would cause the foot to rotate in the frontal plane differently in each of the flip-flop conditions than in BF. In the current study, there was no significant main effect of footwear on PRO_{PEAK} in NA individuals (Figure 35). As with pronation at midstance (PRO), there was no difference between any footwear conditions for peak pronation. This statistical insignificance suggests that for the PRO_{PEAK} variable walking in flip-flops is like walking barefoot (Table 10). Based on the above finding anecdotal claims that walking in flip-flops is comparable to walking barefoot, in regards to PRO_{PEAK}, may be correct. As with PRO, it may be that the foot arch of NA individuals did not collapse enough to require additional support provided by the arch support of the flip-flop that would prevent excessive pronation further supporting the findings that there was no difference between walking BF and walking in flip-flop conditions for PRO_{PEAK}.

Peak Dorsiflexion

Dorsiflexion, for the purposes of this study, was defined as the relationship of the hindfoot relative to the tibia about the transverse axis of the distal tibia (Figure 20a). A previous study suggested that when wearing sneakers the ankle angle has greater dorsiflexion during the swing phase of gait when compared to flip-flops (J. F. Shroyer & Weimar, *In Press*). The present study shows that there was a significant main effect for footwear on peak dorsiflexion during the swing phase of the gait cycle in NA individuals. The BF condition resulted in greater dorsiflexion of the ankle when compared to FF1 and FF2, but not FF3 (Table 10).

Although barefoot and sneakers are not the same, it is interesting to compare the previous research by Shroyer and Weimar (*In Press*) and note that both walking barefooted and in sneakers demonstrated an increase in dorsiflexion during the swing phase when compared to walking in flip-flops. However in the current study FF1 and FF2 resulted in decreased $DORSI_{PEAK}$, but FF3 did not. These results eliminate the arch support as the reason for these differences as there was no statistical difference between $DORSI_{PEAK}$ while walking in FF1 and FF2, but do suggest that other architectural differences between FF1, FF2 and FF3 may play a role in the discrepancies in dorsiflexion. Flip-flop 3 displayed a specific architectural difference with regards to the Y-strap; the Y-straps are wider and run more posterior along the foot in FF3. This positioning of the strap and may have caused an increase in the contact surface area of the foot when compared to the Y-straps of FF1 or FF2. By having this positional advantage, FF3 may have kept the flip-flop closer to the foot, protecting the heel, and allowing for greater $DORSI_{PEAK}$. Due to the proposition that the heel of the flip-flop moved further away from the heel of the foot during the swing phase for FF1 and FF2, it is reasonable to conclude that the participant did not dorsiflex to such a degree as to potentially expose the heel to contact with the ground. Therefore, the participant decreased dorsiflexion in FF1 and FF2 conditions to ensure that the heel of the foot would remain closer to the heel pad of FF1 and FF2.

Peak Hallux Extension

Previous research that investigated plantar/dorsiflexion between flip-flops and sneakers found that plantar flexion was increased in flip-flops (J. F. Shroyer & Weimar, *In Press*). Researchers hypothesized that increased plantar flexion in flip-flops could be attributed to the contraction of the flexor digitorum longus (FDL) and flexor hallicus longus (FHL) in an attempt to use the phalanges to grip the flip-flop and prevent the flip-flop from coming off the foot. Once the phalanges begin to flex, the FDL and the FHL contribute to an implied ankle plantar flexion moment because the FDL and FHL do not only cross the metatarsal phalange joint but also the ankle joint (J. F. Shroyer & Weimar, *In Press*). Based on this hypothesis about toe flexion, it was anticipated that when walking in flip-flops, the phalanges would flex more in FF1 and FF2 than in BF and FF3.

Hallux extension is defined as the relationship of the hallux relative to the forefoot about the transverse axis of the distal forefoot (Figure 22). The current study demonstrated that there was no effect of footwear on HX_{PEAK} during the swing phase of gait in NA individuals (Figure 37). This statistical nonsignificance suggests that walking in flip-flops is like walking barefoot with regard to HX_{PEAK} (Table 10). As already discussed in the previous section on dorsiflexion, greater dorsiflexion was observed in BF and FF3 compared to FF1 and FF2. Thus, plantar flexion was increased in FF1 and FF2 compared to BF and FF3. The reason for less plantar flexion in BF could be attributed to the fact that there was no flip-flop to "grip;" less hallux flexion in FF3 could have been due to structures such as the depressed heel cup and a raised toe ridge that would aid the phalanges in gripping the flip-flop. While there was no difference in HX_{PEAK} between any footwear conditions, this finding does not dispel the increased activity of the FHL and FDL and increased plantar flexion moment theory. It is possible that the proposed implied moment may be present without any observable joint action at the 1st metatarsalphalangeal joint. Future research should investigate whether there were any differences in the flexion/extension of the 2-5 metatarsalphalangeal joints and/or increased activity in the FHL and/or FDL. In addition, inverse dynamics could be applied to determine if the implied plantar flexion moment is present during swing phase.

Section 2: Footwear Arch and Foot Arch Type

The effect of footwear on individuals with varying medial arch types has been investigated in previous research (Butler et al., 2006; Butler et al., 2007; Hillstrom et al., 2005; Kim et al., 2005); however, there is a gap regarding the influence of the flip-flop, on gait kinematics in individuals with different arch types. The purpose of this portion of the current study was to investigate the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified as having LA, NA, or HA feet. Gait kinematics of interest included SL, EV, EV_{PEAK}, PRO, PRO_{PEAK}, $DORSI_{PEAK}$, and HX_{PEAK}. Means, standard deviations, and p-values for all kinematic variables are presented in Appendix B.

For purposes of discussion in this section, the gait observed when BF for NA individuals will be considered a “normal gait” for all foot arch types. Individuals with a low arch have less foot rigidity and subsequently increased foot motion during gait. Conversely, individuals with high arches have increased foot rigidity and subsequently decreased foot motion. Both increased and decreased foot motion have been identified as a risk factor for potential foot, ankle, and lower leg injuries (Butler et al., 2006; Butler et al., 2007; Kaufman et al., 1999; Knapik et al., 1999). If any of kinematic variables differ from the BF condition for NA individuals, potential problems could arise due to altered foot kinematics brought about by the introduction of the flip-flop in such a manner that structural properties of the foot are not functioning as they would without footwear (i.e. BF).

Stride Length

Results of this portion of the study showed no interaction effect of footwear and foot arch type on SL meaning that individuals with different foot arch types did not demonstrate different SL across the footwear conditions (Figure 38). Results, however, did show a significant main effect of footwear on SL where BF resulted in a shorter SL when compared to both FF1 and FF2. When data was collapsed across arch types no significant main effect of foot arch type was noted on SL (Figures 39 & 40). The effect of footwear on SL in this current study is supported by a study that investigated the effects of flip-flops and sneakers on gait kinematics. The previous study found that when compared to athletic shoes flip-flops resulted in decreased SL (J. F. Shroyer & Weimar, *In Press*). Researchers explained this shorter SL as the product of a direct correlation between the decreased mass of the flip-flop and decreased inertia of the distal segment (the foot) during the swing phase of gait. The increased inertia of the heavier shoe is thought to carry the foot further forward (Mundermann et al., 2003; Oeffinger et al., 1999; J. F. Shroyer & Weimar, *In Press*). By reviewing the inertia explanation, the findings of this portion of the present study in which the added mass of the flip-flops resulted in an increased SL when compared to barefoot is plausible.

Eversion at Midstance

It was hypothesized that more EV would be seen in LA individuals, less EV would be seen in HA individuals, and that NA individuals would show EV between that of LA an HA individuals. The previous hypothesis is based on prior research that has noted that pes planus individuals have increased foot motion where as pes cavus individuals have decreased foot motion (Kaufman et al., 1999; Knapik et al., 1999). Results of the present study showed an interaction effect of footwear and foot arch type on EV (p = .007) indicating that individuals with different foot arch types demonstrated different eversion angles across footwear conditions (Figure 41). Footwear had no effect on EV in LA individuals, suggesting that walking in flip-flops is like walking BF for LA individuals. Figure 41 indicates that the interaction occurs between the effect of footwear and arch type on eversion for FF1 and FF2 specifically between NA and HA individuals. Flip-flops limited eversion in NA and HA individuals when compared to BF. This limiting ability of flip-flops suggests that individuals with HA should wear flip-flops, specifically flip-flops without an arch support in order to most resemble normal gait. Individuals with NA had EV in FF2 that more closely resembled the BF condition; therefore, individuals with NA should wear flip-flops with an arch support.

There was less EV in LA individuals in all footwear conditions when compared to both NA and HA individuals. It is interesting to note that there was no difference in EV between NA and HA individuals for all footwear condition. This finding may seem in error until it is considered that all angles are calculated from a static position. In the static position, individuals with LA would be in an everted static position compared to HA individuals, whom would be in an inverted static position (Figure 61). Therefore, it is reasonable to conclude that compared to the static position, HA individuals were able to evert more because the LA individuals were already everted in the static position.

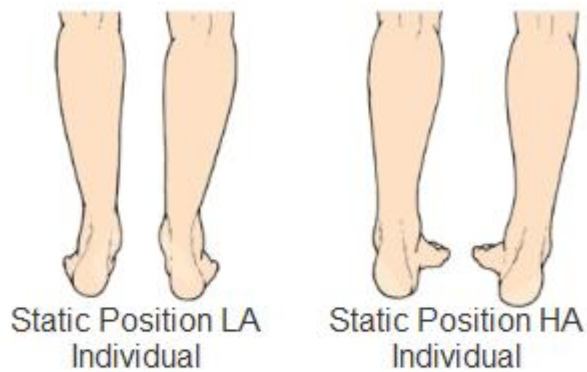


Figure 61. Drawing representation of static position for LA and HA individuals

Peak Eversion

Results showed an interaction effect of footwear and foot arch type on EV_{PEAK} indicating that individuals with different foot arch types produced different EV_{PEAK} across the footwear conditions. By examining the graph of the interaction (Figure 44) and the *t* tests, it is apparent that EV_{PEAK} is affected by flip-flops when compared to all foot arch types. Peak eversion was larger in BF when compared to FF1 and FF2 for LA, NA, and HA individuals; however, there was no significant difference between FF1 and FF2 for any of the foot arch types. These findings suggest that walking in flip-flops is not like walking barefoot, and there is no effect of flip-flop arch support on EV_{PEAK}. The findings also show that flip-flops limit peak eversion of the foot for all arch types. When comparing NA and HA individuals, there was no difference in EV_{PEAK} for any of the footwear condition. This suggests that NA and HA individuals have similar peak eversion regardless of flip-flop type or barefoot. Based on these results for EV_{PEAK}, individuals with HA should wear flip-flops with or without an arch support to resemble a "normal gait." For NA individuals, FF2 resulted in a smaller mean difference from the BF condition so it could be recommended that NA individuals wear flip-flops with an arch support.

Pronation at Midstance

It was hypothesized that the arch support of FF2 would resist the pronation of the foot, particularly for individuals with LA. Results of this study showed no significant interaction effect of footwear and foot arch type on PRO indicating that individuals with different foot arch types did not produce different PRO across the footwear conditions (Figure 47). It should be noted, however, that the *p*-value of the interaction was $p = .066$ and as such was approaching significance; however, there was a low effect size ($\eta^2 = .056$) and moderate power (*Power* = .625). Observation of the interaction graph shows a trend of footwear limiting pronation in LA individuals, with the greatest reduction observed in FF2. This reduction in pronation suggesting that arch support may be limiting PRO in LA individuals; however, the interaction was not statistically significant, so this conclusion should be read with caution.

The present study also found no significant main effect of footwear on pronation at midstance ($p = .456$) (Figure 48). This statistical nonsignificance suggests that there was no difference in pronation of the foot between barefoot and flip-flops. It was expected that there would be no difference in BF and FF1 since FF1 does not have an arch support to limit pronation of the foot; however, lack of a significant difference between FF1 and FF2 was not expected. Interestingly, the non-significant effect of footwear on PRO is supported by a previous study that investigated the effects of sandal arch height on foot and ankle biomechanics in individuals with pes planus (Hillstrom et al., 2005). The study measured malleolar valgus index (MVI), which is considered to be an indication of pronation, during a static stance position. Hillstrom et al. (2005) found that lower values (less pronation) were observed in the sandal with the lowest arch height and higher (more pronation) in the sandal with the highest arch height. This was contrary to what Hillstrom, Song, Kim & Heilman had expected to observe, as researchers expected the sandal with the largest arch height to correct the pronation (Hillstrom et al., 2005). Midstance in normal gait is the point at which the vertical ground reaction force most resembles force from body weight on the foot. Therefore, it is appropriate to compare the findings of the footwear arch types during static stance trials by Hillstrom et al. (2005) to the current findings at midstance. The unexpected findings of the current study coincide with the unexpected findings of Hillstrom et al. (2005). Researchers involved with the previous study offered no explanation, but based on the current study, the arch support of FF2 may not have been high enough to limit pronation on the foot. Future studies should investigate the influence of higher arch supports.

There was a significant main effect of foot arch type on PRO with greater PRO in LA individuals when compared to NA and HA individuals (Figure 49); however, there was no difference in PRO between NA and HA individuals. The findings of this study supports the hypothesis that greater PRO is found in LA individuals when compared to

NA and HA individuals; but failed to support the hypothesis that there is a difference in PRO between NA and HA individuals. Previous literature noted that LA individuals have decreased foot rigidity and increased foot motion when loaded; contrastingly, HA individuals have increased foot rigidity and decreased foot motion when loaded (Kaufman et al., 1999; Knapik et al., 1999). Like Kaufman et al., (1999) and Knapik et al., (1999) the individuals with LA in the current study demonstrated increased foot motion identified by larger PRO angles. The similar behavior for PRO in NA and HA individuals could be explained by similar stiffnesses of the foot arches in these individuals. Future research should incorporate arch stiffness as well as arch height. In addition, the foot arch support may have been too stiff that body weight during midstance was not enough to induce a large enough arch collapse to result in a significant statistical difference between NA and HA for PRO angles. Foot arch support and stiffness in NA and HA individuals makes footwear arch support superfluous to individuals with NA and HA.

Peak Pronation

The results of the current study showed no interaction effect of footwear and foot arch type on PRO_{PEAK} indicating that individuals with different foot arch types did not show different PRO_{PEAK} across footwear conditions (Figure 50). Nor was there a significant main effect of footwear on PRO_{PEAK} for all foot arch types (Figure 51). An explanation for the similar PRO_{PEAK} angles seen in each of the three conditions, BF, FF1, and FF2, is that the hindfoot comes off the flip-flop during the time PRO_{PEAK}

occurs. For this reason, the structural components of the flip-flops did not affect motion of the foot at the time during which PRO_{PEAK} was determined.

There was a significant main effect of foot arch type on PRO_{PEAK} (Figure 52). Greater PRO_{PEAK} was found in LA individuals when compared to NA and HA individuals. It is interesting to note that there was no statistical difference in PRO_{PEAK} between NA and HA individuals. The findings of this study supports our hypothesis that more PRO_{PEAK} was seen in LA individuals compared to NA and HA individuals. It was also hypothesized that there would be a difference in PRO_{PEAK} between NA and HA individuals. Results of the current study were found to be contrary to the second part of our hypothesis; there was no statistical difference in PRO_{PEAK} between NA and HA individuals. Kaufman et al. (1999) and Knapik et al. (1999) state that LA individuals have decreased foot rigidity and increased foot motion when loaded; whereas, HA individuals have increased foot rigidity and decreased foot motion when loaded (Kaufman et al., 1999; Knapik et al., 1999). This increased foot motion and decreased rigidity explains why individuals with LA demonstrated increased PRO_{PEAK}. Similarly to PRO, the observation of PRO_{PEAK} in NA and HA individuals suggests that the relative common rigidities of the foot in NA and HA individuals also elicit similarities in how the heel comes off the flip-flop. The similar heel motion may be why there was no difference in PRO_{PEAK} between NA and HA individuals.

Dorsiflexion

The results of this study showed an interaction effect of footwear and foot arch type on $DORSI_{PEAK}$ (p = .034); indicating that individuals with different foot arch types recorded different $DORSI_{PEAK}$ across footwear conditions. There were no significant differences between foot arch types for each footwear condition. This suggests that each foot arch type (LA, NA, and HA) have similar dorsiflexion angles at the ankle for each footwear condition. Hoevwer, the footwear conditions have an individual effect on each foot arch type. Flip-flops result in decreased dorsiflexion for all arch type when compare to BF. This suggests that walking barefoot is not the same as walking in flip-flops with regard to dorsiflexion. There was no difference between FF1 and FF2 for $DORSI_{PEAK}$ in LA and NA individuals. This indicates that arch support has no effect on $DORSI_{PEAK}$ in individuals with LA or NA. However, there was a significant difference in FF1 and FF2 fo HA individuals. The FF2 resulted in a decreased $DORSI_{PEAK}$ in HA individuals when compared to FF1. This suggets that the arch support had an effect on HA individulas and limited the $DORSI_{PEAK}$.

It was not anticipated that the presence of an arch support would have any influence on dorsiflexion, regardless of foot arch height. Based on these results in HA individuals, a flip-flop with an arch support yielded a gait that deviated more from the "normal gait" of BF in NA individuals. Based on the findings, a flip-flop without an arch support may be better for HA individuals than a flip-flop with an arch support. The smallest mean difference between flip-flop to BF in NA individuals was for FF2, suggesting that if NA individuals wear a flip-flop, it should be one with an arch support.

Hallux Extension

Results showed no significant interaction effect of footwear and foot arch type on HX_{PEAK} indicating that individuals with different foot arch types did not record different

HX_{PEAK} across the footwear conditions (Figure 56). The current study also found no significant main effect of footwear or foot arch type on HX_{PEAK} (Figures 57 and 58). The implication of these findings is that there is no difference in the flexion/extension of the hallux between BF, FF1 and FF2, for LA, NA, and HA individuals. As mentioned in the discussion section 1, it was hypothesized that an increase in toe flexion from the contribution of the FHL and FDL would be observed and help to explain the changes seen in dorsiflexion when subjects walked in sneakers compared to flip-flops (J. F. Shroyer & Weimar, *In Press*); interestingly, this incongruency in dorsiflexion angles was not observed in the current study. The purpose of this study was to investigate the influence of the presence of an arch support on the proposed variables and as such it was not anticipated that an arch support would affect HX_{PEAK}. Future research needs to investigate the activity of the FHL and FDL as well as the flexion/extension of the 2^{nd} through 5^{th} phalanges of the foot.

Section 3: Surface Electromyography (sEMG)

Average sEMG and Peak sEMG

Previous research has shown decreased dorsiflexion in flip-flops when compared to sneakers (J. F. Shroyer & Weimar, *In Press*). The authors concluded that one explanation for the decreased dorsiflexion in flip-flops when compared to sneakers could be the increased activity of the FHL and FDL to flex the phalanges to grip the flip-flop during the swing phase in an attempt to keep the flip-flop on the foot. This increased activity of the toe flexors would subsequently cause an increased plantar flexion moment at the ankle resulting in decreased dorsiflexion (J. F. Shroyer & Weimar, *In Press*). It was hypothesized that if there was an increased plantar flexion moment caused by the FHL and FDL that there may be an increase in the dorsiflexor muscles to counter act the increased plantar flexion moment. One of the major dorsiflexors in the lower leg is the tibialis anterior (TA). As a result this study investigated the average ($sEMG_{AVG}$) and peak ($sEMG_{PEAK}$) sEMG of the TA during the swing phase of gait in NA individuals. The swing phase consisted of the time period from toe off to heel contact of the ipsilateral foot.

The present study found that there was a main effect of footwear on both $sEMG_{AVG}$ and $sEMG_{PEAK}$ in NA individuals. Specifically, there was a significant difference between BF and all flip-flop conditions, FF1, FF2, and FF3, with flip-flops resulting in an increase in $sEMG_{AVG}$ and $sEMG_{PEAK}$ when compared to BF (Figures 59 and 60). However, there was no difference in $sEMG_{AVG}$ or $sEMG_{PEAK}$ between any of the flip-flop conditions. These findings support a study that investigated the effect of foot orthotic wedging on sEMG of lower leg muscles. Researchers found an increase in TA activity during the gait cycle in all footwear conditions compared to barefoot. Like the current study, there was also no significant difference between the four footwear conditions when orthotics were worn (Murley & Bird, 2006).

Several explanations can be made in interpreting the findings of the current study. First, the increased mass of the flip-flops when compared to BF would result in increased activity of the TA. In this case, the TA would have to dorsiflex a segment with more mass resulting in increased muscle activity. However, since there was no difference in $sEMG_{AVG}$ and/or $sEMG_{PEAK}$ between the lightest flip-flop (FF1 = 92.1 g) and the heaviest flip-flop (FF3 = 142.2 g), this explanation loses credibility. Second, flip-flops add to the length of the leg segment. This is not to say that flip-flops actually make the leg longer, but when wearing footwear, the bottom of the foot is not at the end of the distal segment. Individuals must consider the addition of the sole of the footwear; therefore, the ankle is required to have greater dorsiflexion to clear the walking surface. This supposition also looses credibility because previous research has shown that less dorsiflexion was experienced in flip-flops when compared to sneakers (J. F. Shroyer & Weimar, *In Press*). Last, theoretically, the flip-flop may require more dorsiflexion because it falls off the hindfoot during the swing phase resulting in an individual having an even longer segment needing clearance over the walking surface during swing phase. In conclusion, it seems that there are two plausible explanations for the increased $sEMG_{AVG}$ and $sEMG_{PEAK}$: (1) there is an increase plantar flexion moment caused by the FHL and FDL when wearing flip-flops thus, dorsiflexors (ie. TA) must be more active to counteract this plantar flexion moment and (2) the flip-flop falling away from the heel of the foot, may require the ankle to have greater dorsiflexion in an attempt to bring the sole of the flip-flop in contact with the sole of the foot prior to floor contact.

Section 4: Summary

The purpose of this section is to summarize the findings of the present study and combine the results from the three separate research questions. This summary includes two parts:

- Part 1: summarized discussion comparing the effects that different components of the thong style flip-flop have on gait kinematics in individuals classified with NA feet and the effects that a thong style flip-flop arch support has on gait kinematics of individuals classified with either LA, NA, or HA feet.

- Part 2: summarized discussion comparing the muscular activity of the tibialis anterior (TA) of the ankle during the swing phase of gait when wearing thong style flip-flops and dorsiflexion angle in individuals classified with NA feet.

Summary: Part 1

Stride Length

When looking at both research questions flip-flops resulted in longer SL when compared to BF. In the second research question, there were no statistical differences between the flip-flop conditions. Together, these finding support that SL is affected by the mass of the footwear (Oeffinger et al., 1999); however, when only NA individuals were considered in the first research question of this study, there was a significant greater SL observed in FF1 than FF2 (Figure 31). This greater SL was not observed in the second research question that investigated LA, NA, and HA individuals (Figure 39). The inertia hypothesis (Mundermann et al., 2003; Oeffinger et al., 1999; J. F. Shroyer & Weimar, *In Press*) identified to explain SL differences is not supported by the current research project as there was no difference between the lightest flip-flop (FF1) and the heaviest flip-flop (FF3). Also, FF2 resulted in a smaller stride length than FF1, even though FF2 had more mass. So, in one regard this study has supported previous literature in that when compared to BF, footwear increased SL. However, it also appears that inertia may not completely explain the differences in SL. The author hypothesizes that people may take shorter strides when walking barefoot to decrease the negative acceleration at foot/ground contact. The decreased negative acceleration at contact results in less force at the unprotected barefoot. With a decreased SL, it is thought by the author that there is also a smaller peak velocity during swing phase. This smaller velocity would be preferable when the foot makes contact with the ground, especially when BF, because a concomitant negative force is required to bring the foot to rest. The greater this force is at contact with the ground, the more shear force will be exerted to the base of the foot which could lead to blisters and/or bruising from walking barefoot. Future research should investigate the stride length between people who walk in barefoot often versus those who are usually shod and include kinetic variables such as anterior/posterior forces to investigate the decreased force theory.

Eversion at Midstance

Based on the current study, individuals with different foot arch types produced different EV angles across the footwear conditions (Figure 41). These results suggest that flip-flops limited EV more in NA and HA individuals when compared to BF. Also, FF2 (with an arch support) limited EV the most for HA individuals so that their gait resembled “normal” gait. When HA individuals wore flip-flops they produced an EV angle closer to the EV angle observed in BF for NA individuals; therefore, it would be most beneficial for HA individuals to wear flip-flops, specifically flip-flops with an arch support than flip-flops without an arch support or barefoot when considering only EV.

The first research question and the second research question of the current study support each other in that flip-flops (FF1 and FF2) limited EV when compared to barefoot. Results obtained during the FF3 condition did not limit EV (Figure 32).

Ultimately, not all flip-flops were found to limit EV in individuals with NA; it was predicted that EV would be limited in all flip-flop conditions including FF3. It was thought that limited foot motion, especially for LA and NA individuals would be beneficial due to the increased risk for lower leg and foot problems associated with excessive foot motion. Increased ankle EV in FF3 may be due to the pronounced heel cup that provides a ridge on the medial and lateral side of the rear of the foot. In theory, these medial and lateral ridges cause an implied eversion angle on the foot from the flip-flop. The pronounced heel cup may also explain the similarities observed in BF and FF3 since the heel cup may have caused the foot to be everted the same magnitude as the difference observed between BF and FF1 and FF2. This potential EV caused by the heel cup may have made it seem that FF3 caused the foot to evert the same as BF. Furthermore, based on anecdotal observation, BF and FF3 individuals rolled the foot more like a "normal" gait. During "normal" gait the center of pressure (CoP) initially proceeds laterally and as the foot progresses through midstance to heel off and toe off, the CoP proceeds more medially. This "normal" gait motion was not observed in individuals while wearing FF1 and FF2. This anecdotal observation by the researchers could explain why eversion was limited in FF1 and FF2 when compared to BF and FF3. Future research should evaluate ground reaction forces and CoP path during the stance phase to determine if this anecdotal observation is correct.

Peak Eversion

Based on the current study, individuals with different foot arch types produced different EV_{PEAK} across the footwear conditions (Figure 44). Overall, EV_{PEAK} is limited in flip-flops compared to barefoot for all arch types. The effect of FF1 and FF2 was the same for LA and HA individuals; however, FF2 had larger mean difference for HA individuals (2.78 deg) compared to FF1 (1.59 deg). A possible explanation could be that the Y-strap that runs from between the hallux and second phalange to both the medial and lateral side of the foot limits EV_{PEAK} of the hindfoot. The strap runs across the midfoot and hindfoot and may prevent the hindfoot from moving laterally and as a result may limit eversion of the ankle; whereas the Y-strap of FF1 did not cross the hindfoot. These findings also demonstrate that walking in flip-flops is not similar to walking barefoot when evaluating EV_{PEAK}. For HA individuals, FF1 resulted in EV_{PEAK} that was most similar to that observed in the BF condition for NA individuals; therefore, flip-flops without and arch support are recommended for HA individuals..

Pronation at Midstance and Peak Pronation

The implications of the findings for PRO and PRO_{PEAK} were similar, so these two variables have been combined for discussion purposes in this section. It was expected that the footwear arch supports of FF2 and FF3 would have limited PRO and PRO_{PEAK}; however, the findings were contrary to expected results. It was interesting to note that the interaction effect was approaching statistical significance and that there was a trend of decreased PRO for LA individuals in FF1 and FF2 (Figure 47). One reason the flip-flops arch support did not limit PRO and PRO_{PEAK} is that the foot is not confined to the flip-flop as is the case in sneakers. This freedom of the foot to move on the flip-flop could be the reason behind the flip-flops inefficiency at controlling foot motion. Results of the second research question showed a significant main effect of foot arch type on PRO and

$PRO_{PEAK.}$ Low arch individuals had greater PRO (Figure 49) and PRO_{PEAK} (Figure 52) than NA and HA individuals. It was expected that LA individuals would have greater PRO and PRO_{PEAK} than NA, and NA individuals to have greater PRO and PRO_{PEAK} than HA individuals. These expectations were drawn based on the idea that individuals with LA are characterized as having increased foot motion in the frontal plane and HA individuals are characterized as having decreased foot motion in the frontal plane. Contrastingly, results showed no difference in NA and HA individuals for PRO and PRO_{PEAK}. Therefore the implication is that flip-flops do not limit pronation of the foot. Specifically, the arch support does not have any influence on pronation of the foot. This suggests that flip-flop pronation kinematics are like walking barefoot. It may be that in the flip-flop condition, the foot is not confined to a specific spot on the footwear and therefore, the flip-flop cannot effectively limit motion of the foot.

Peak Dorsiflexion

Based on the current study, individuals with different foot arch types reported different $DORSI_{PEAK}$ across the footwear conditions (Figure 53). The results of the interaction suggest that flip-flops limited $DORSI_{PEAK}$ in all arch types when compared to BF, and that FF2 (with an arch support) imposed the greatest limitation on $DORSI_{PEAK}$ for HA individuals.

The findings of both the first and second research question show that flip-flops limited DORSI when compared to BF, and the arch support of the flip-flop did not affect DORSI (Figures 36 and 54). Considering the first research question, FF3 behaved in a similar manner to BF possibly due to features such as the heel cup and the Y-straps, which are wider and run more posterior along the foot. Structural positioning of the Y-straps may increase the contact surface area of the foot in FF3 more than the Y-straps of FF1 and FF2. These structural features may have kept the flip-flop closer to the foot and allowed for dorsiflexion similar to that observed in the BF condition. These findings also shed doubt on the inertia explanation for increased SL for footwear conditions. If inertia increase was the reason for differences observed in kinematics between barefoot and footwear, the increased mass of the flip-flops would have contributed to an increased moment and consequently increased dorsiflexion. However, this was not the case.

Peak Hallux Extension

The results for HX_{PEAK} yielded no significant effect of footwear or foot arch type (Figures 37, 57, and 58). It was hypothesized that there would be decreased hallux extension in the flip-flop conditions in attempt of the phalanges to grip the flip-flop. The results were contrary to what was hypothesized. It is plausible that there was increased activity of the FHL and FDL resulting in an increased plantar flexion moment but without any observable joint action at the 1st metatarsalphalangeal joint. The increased activity could have resulted in movement at the ankle joint and not the 1st metatarsalphalangeal joint. Also, the hypothetical “gripping” of the toes could be occurring not at the hallux but at the 2nd through 5th metatarsalphalangeal joints. Future research should investigate the flexion/extension of the 2nd through 5th metatarsalphalangeal joints as well as activity of the FHL and FDL.

Summary: Part 2

Tibialis Anterior sEMG and Dorsiflexion

Previous research has shown decreased dorsiflexion during swing phase of the ankle in flip-flops when compared to sneakers (J. F. Shroyer & Weimar, *In Press*). Authors concluded that one explanation of decreased dorsiflexion seen in flip-flops when compared to sneakers could be the increased activity of the FHL and FDL in order to flex the phalanges and grip the flip-flop during swing phase in an attempt to keep the flip-flop on the foot. This increased activity of the toe flexors would subsequently cause an increased plantar flexion moment at the ankle resulting in decreased dorsiflexion (J. F. Shroyer & Weimar, *In Press*). It was hypothesized that if there was an increased plantar flexion moment caused by the FHL and FDL there may be an increase in the dorsiflexor muscle activity to counter act the increased plantar flexion moment. One of the major dorsiflexors in the lower leg is the tibialis anterior (TA). The current study found that there was a main effect of footwear on both $sEMG_{AVG}$ and $sEMG_{PEAK}$ in NA individuals. There was a significant difference between BF and all flip-flop conditions: FF1, FF2, and FF3. The flip-flops conditions resulted in an increase in $sEMG_{AVG}$ and $sEMG_{PEAK}$ when compared to BF (Figures 59 and 60); however, there was no difference in $sEMG_{AVG}$ or $sEMG_{PEAK}$ for any of the flip-flop conditions. This suggests that walking in flip-flops is not analogous to walking barefoot. Also, the lack of a statistical difference between flip-flop types negates the thought that a heavier “shoe” creates the need for increased TA activity and further strengthens the idea that the FDL and FHL are more active in flip-flops conditions and cause a plantar flexion moment.

The current study also found a significant main effect of footwear on $DORSI_{PEAK}$. The BF condition resulted in an increased $DORSI_{PEAK}$ compared to FF1 and FF2; however, no significant difference between FF3. A drawing representing the combined main effects of $sEMG_{AVG}$, $sEMG_{PEAK}$, and $DORSI_{PEAK}$ is presented in Figure 62. Note that sEMG activity was increased in all flip-flops conditions compared to BF. Also, FF1 and FF2 had similar dorsiflexion angles; BF and FF3 had similar dorsiflexion angles. It is counter intuitive that there would be increased dorsiflexion muscle activity (i.e. the TA) and less dorsiflexion, yet this is the case for the BF condition. Further, results show that even though FF1 and FF2 had an increased contribution to a dorsiflexion moment provided by the TA, there was less dorsiflexion. It is hypothesized that another factor, such as the activity of the FHL and the FDL, is providing a plantar flexion moment that must be overcome by the TA, and overcoming this plantar flexion moment by the TA results in the increased activity of the TA and a smaller dorsiflexion angle.

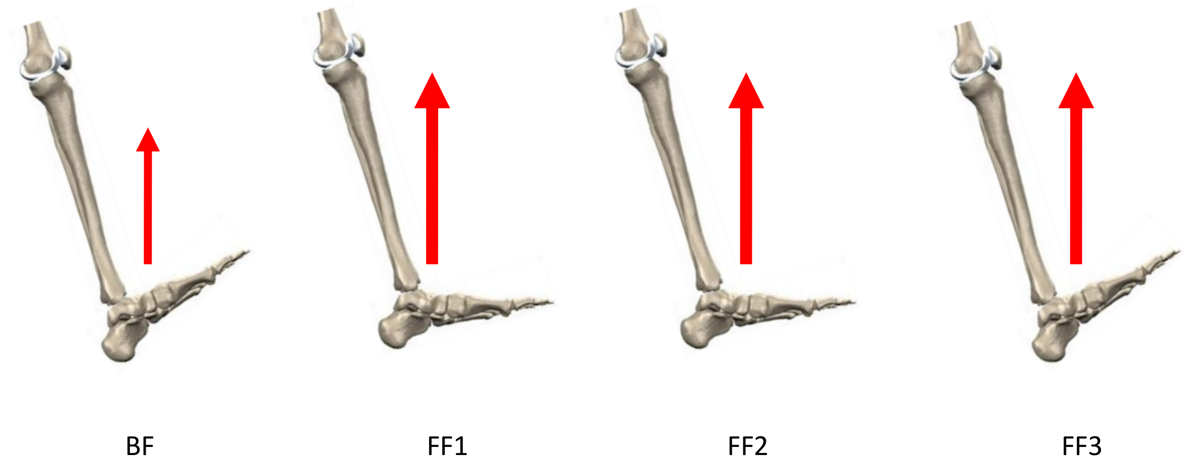


Figure 62. Drawing representation of ankle dorsiflexion and tibialis anterior (TA) surface electromyography activity (sEMG).
Note. Pictured ankle angles represent relative differences in dorsiflexion angle between footwear. Arrows represent both $sEMG_{AVG}$ and $sEMG_{PEAK}$. Larger arrow represents greater EMG activity. .

Section 5: Conclusions and Future Research

This final section will discuss the conclusions drawn from this current study and directions for future research. This section will be divided into four sections: (1) the effect of different thong flip-flops on gait kinematics in NA individuals, (2) the effect of thong style flip-flop arch support on gait kinematics in LA, NA, and HA individuals, (3) the effect of thong style flip-flops on muscular activity the TA during the swing phase in NA individuals, and (4) avenues for future research.

Flip-flops on Gait Kinematics in NA Individuals

It can be concluded from the current research that footwear, specifically flip-flops, affects certain gait kinematics in college aged females. There are anecdotal claims that walking in flip-flops is analogous to walking barefoot. Based on seven kinematic variables, the authors conclude that walking in flip-flops is not analogous to walking barefoot. The current study investigated three different types of flip-flops. It was observed that specific components that are common to flip-flops had an effect on whether or not the flip-flops influenced gait to be more like the gait observed in barefoot. Common features of flip-flops that were investigated included an arch support, heel cup, midtarsal support, larger and thicker straps that proceeded more posteriorly, and a toe ridge. It was found that the flip-flop that had all of these components resulted in a gait that more resembled barefoot walking than other flip-flops, however, no flip-flop completely epitomized walking barefoot. Flip-flops that did not have any of these features or only an arch support were even less like walking barefoot than was the flip-flop with all features. In conclusion, authors recommend that in order to resemble walking barefoot while wearing flip-flops, college aged females with NA should wear flip-flops that have a pronounced arch support, toe ridge, midtarsal support, heel cup and thicker straps that proceed past the mid-point of the flip-flop.

In addition to answering the question, "Is walking in flip-flops like walking barefoot?" the authors wanted to investigate how different components of commercially available flip-flops affect gait kinematics of college aged females with NA. When evaluating the affect of an arch support, the only variable that was different was stride length. It was hypothesized that an arch support would limit foot motion specifically in the frontal plane and would produce differences in eversion and pronation variables; however, the current study concludes that an arch support in flip-flops does not significantly affect gait kinematics. It was also hypothesized that flip-flops with a larger arch support, heel cup, midtarsal support, toe ridge and larger straps would reduce foot motion, specifically pronation and foot eversion, but there was no difference between this type of flip-flop and other flip-flops for pronation. Additionally, flip-flops with a larger arch support, heel cup, midtarsal support, toe ridge and larger staps resulted in increased eversion. As a result, it may be concluded that the flip-flop that had a pronounced arch support, toe ridge, midtarsal support, heel cup and thicker straps had a counter intuitive effect on gait kinematics and caused increased foot motion. Results of this study were interesting in that the arch support did not limit forefoot or hindfoot motion. It is concluded that because the foot is not fixed on the flip-flop and free to move that the structural components lose their ability to affect foot motion as designed. In conclusion, walking in flip-flops is *not* like walking barefoot. If walking barefoot is considered to be optimal, then further research is required to design flip-flops that produce more natural gaits.

Based on the current findings of the first research question, the author has suggested appropriate flip-flops for NA individuals for each kinematic variable measured (Table 12). It should be noted that BF would result in the most "normal" output for the kinematic variable; however, the author wanted to include only recommendations for flip-flops. Based on Table 12, FF2, the flip-flop with an arch support, is recommended for all foot arch types.

Table 12
Recommendation of flip-flop for NA individuals based on kinematic variables

Kinematic variable	NA
SL	—
EV	FF3
EV_{PEAK}	FF3
PRO	—
PRO_{PEAK}	—
$DORSI_{PEAK}$	FF3
HX_{PEAK}	—
Overall recommendation	FF3

Note. — means not able to recommend

Flip-flops and Foot Arch Type on Gait Kinematics

Increased foot motion has been linked to increased risk for injury in individuals with pes planus (Kaufman et al., 1999; Knapik et al., 1999). In the athletic shoe industry, there are running shoes designed specifically for certain foot types. Motion control shoes are designed for individuals with flat feet and cushioning shoes are designed for individuals with high arches (Butler et al., 2006; Butler et al., 2007). The authors wanted to investigate if there was an effect of arch support in flip-flops on individuals with low, normal or high arches. For all arch types, the authors conclude that flip-flops limited eversion and dorsiflexion; however, there was no effect on pronation of the foot in college aged females. It may also be concluded that an arch support had the greatest affect in limiting foot motion in college aged females with high arches. For college aged females with high arches, the arch support limited peak eversion and eversion at midstance as well as dorsiflexion. There seemed to be no effect of the flip-flop arch support on college aged females with low or normal arches; however, across all arch types, the flip-flop arch support did limited eversion. The present study also concludes that a flip-flop arch support had no effect on forefoot motion such as pronation; the arch supports inability to limit foot motion is interesting. In addition to the idea that the foot is not constrained enough on the flip-flop to allow the arch to be effective, the arch height of FF2 was only 9 mm and may have not been large enough to restrict or otherwise affect foot motion. The premise of this investigation was to determine if a flip-flop with or without an arch support was more beneficial to individuals with LA, NA, or HA. Based on the current findings, the author has suggested appropriate flip-flops for LA, NA, and HA individuals for each kinematic variable measured (Table 13). It should be noted that BF would result in the most "normal" output for the kinematic variable; however, the author wanted to include only recommendations for flip-flops. Based on Table 13, FF2, the flip-flop with an arch support, is recommended for all foot arch types.

Table 13
Recommendation of flip-flop for LA, NA, and HA individuals based on kinematic variables.

Kinematic variable	LA	NA	HA
SL	—	—	FF2
EV	—	FF2	FF1
EV_{PEAK}	—	—	FF1
PRO	FF2	FF1	FF2
PRO_{PEAK}	FF2	FF2	FF2
$DORSI_{PEAK}$	—	—	—
HX_{PEAK}	—	FF2	—
Overall recommendation	FF2	FF2	FF2

Note. — indicates not able to recommend

Flip-flops on Surface Electromyography in NA Individuals

It can be concluded from the present research that footwear, specifically flip-flops, affects lower leg muscular activity in college aged females. Flip-flops resulted in an increased peak and average activity of the TA during the swing phase of the gait cycle; whereas, barefoot resulted in decreased TA activity. Based on the varying masses of the flip-flops and that there was no difference in either peak or average sEMG of the TA, the author concludes that a variance in mass of the footwear was not the cause of the increased activity of the TA. Thus, there must be some other variable that is causing the discrepancy. Previous research has shown decreased dorsiflexion during the swing phase of the ankle in flip-flops when compared to sneakers (J. F. Shroyer & Weimar, *In*

Press) and it was concluded that the decreased dorsiflexion in flip-flops when compared to sneakers could be explained by a possible increase in activity of the FHL and FDL to flex the phalanges to grip the flip-flop during the swing phase. It was thought that increased flexion was necessary to keep the flip-flop on the foot and subsequently cause an increased plantar flexion moment at the ankle resulting in decreased dorsiflexion (J. F. Shroyer & Weimar, *In Press*). The current study showed that FF1 and FF2 resulted in more decreased $DORSI_{PEAK}$ than did BF but had an increase in muscle activity of the TA. It is expected that increased activity of the TA would yield greater $DORSI_{PEAK}$. Based on the counterintuitive findings of the current study, the author concludes that there is another factor that is causing a counter plantar flexion moment at the ankle. It is suspected that the FHL and FDL may be causing an implied plantar flexion moment at the ankle in an attempt to flex the phalanges to grip the flip-flop.

Future Research

Flip-flops are a common footwear option and there are numerous claims that flip-flops are harmful for foot and lower leg health. Besides the current study and two previous studies (Carl & Barrett, 2008; J. F. Shroyer & Weimar, *In Press*), the author is unaware of any other studies that specifically investigated the effect of thong style flip-flop on gait kinematics. This study focused primarily on gait kinematics; however, future research should include the influence of flip-flops on gait kinetics. A portion of the current study included the effects of flip-flops on muscular activity, but only one muscle was evaluated. Future research should include other muscles of the foot and lower leg, specifically the FHL, FDL, peroneus longus, peroneus brevis, soleus, and gastrocnemius.

Though the current study found significant differences between flip-flops and barefoot kinematic variables, future research needs to address the long term affects of these differences. Longitudinal studies are needed to determine how the measured acute differences manifest into possible chronic orthopedic problems. Also, all flip-flops in the current study were new. Future research needs to address the effects of wear on the performance of the flip-flops on all kinematic and kinetic variables previously mentioned.

Finally, future research should evaluate different types of flip-flops. Only three types of flip-flops were evaluated in the current study. There are numerous other styles available commercially for consumers, and if the scientific community is going to investigate which flip-flop is more beneficial for foot, ankle, and lower leg health more research on all types of flip- flops is needed.

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Appendix A

ANOVA Table for the Effect of Footwear on Gait Kinematics

	Mean	SD	*p*	η^2	*Power*
SL (m)					
BF	1.35	.11			
FF1	1.40	.11	< .001	.517	1.000
FF2	1.39	.11			
FF3	1.40	.11			
EV (deg)					
BF	-4.00	3.5			
FF1	-3.39	3.1	< .001	.156	.991
FF2	-3.55	3.2			
FF3	-4.11	3.3			
EV_{PEAK} (deg)					
BF	-7.08	4.5			
FF1	-5.83	4.3	< .001	.287	1.000
FF2	-5.88	4.1			
FF3	-6.47	4.0			
PRO (deg)					
BF	2.90	5.8			
FF1	2.83	5.7	.630	.011	.168
FF2	3.24	5.4			
FF3	2.74	5.6			
PRO_{PEAK} (deg)					
BF	.520	5.8			
FF1	.253	5.6	.412	.018	.259
FF2	.288	5.2			
FF3	.772	5.5			
$DORSI_{PEAK}$ (deg)					
BF	11.60	4.9			
FF1	10.47	4.9	< .001	.181	.997
FF2	10.60	5.1			
FF3	11.09	5.2			
HX_{PEAK} (deg)					
BF	-38.02	16.0			
FF1	-36.52	18.1	.294	.023	.330
FF2	-37.17	19.0			
FF3	-35.44	18.6			

Appendix B

ANOVA Table for the Effect of Footwear and Arch Type on Gait Kinematics

	LA		NA		HA		*p*		
	Mean	SD	Mean	SD	Mean	SD	Interaction	Footwear	Arch Type
SL (m)									
BF	1.33	.09	1.35	.11	1.29	.11			
FF1	1.37	.08	1.40	.11	1.33	.08	.778	< .001	.223
FF2	1.38	.09	1.39	.11	1.34	.09			
EV (deg)									
BF	-1.73	2.4	-4.00	3.5	-5.47	4.8			
FF1	-1.53	1.7	-3.39	3.1	-3.95	3.3	.007	< .001	.049
FF2	-1.48	1.8	-3.55	3.2	-3.60	2.8			
EV_{PEAK} (deg)									
BF	-4.55	2.8	-7.08	4.5	-8.92	6.4			
FF1	-3.79	2.2	-5.83	4.3	-7.33	5.3	.034	< .001	.092
FF2	-3.69	2.5	-5.88	4.1	-6.14	3.9			
PRO (deg)									
BF	-2.21	5.0	2.90	5.8	4.32	5.3			
FF1	-1.33	4.9	2.83	5.7	4.78	5.8	.066	.456	.018
FF2	-.309	5.1	3.24	5.4	3.30	6.3			
PRO_{PEAK} (deg)									
BF	-4.31	4.1	.520	5.8	2.34	5.7			
FF1	-4.32	4.4	.253	5.6	2.07	5.4	.215	.572	.009
FF2	-3.63	4.5	.288	5.2	.823	5.9			
$DORSI_{PEAK}$ (deg)									
BF	10.3	3.6	11.6	4.9	12.0	2.2			
FF1	9.0	3.3	10.5	4.9	10.2	3.1	.034	< .001	.588
FF2	9.2	3.6	10.6	5.1	9.3	2.3			
HX_{PEAK} (deg)									
BF	-34.6	25.4	-38.0	16.0	-32.4	27.9			
FF1	-33.4	22.6	-36.5	18.1	-30.9	30.9	.677	.366	.541
FF2	-34.6	21.1	-37.2	19.0	-27.6	25.8			

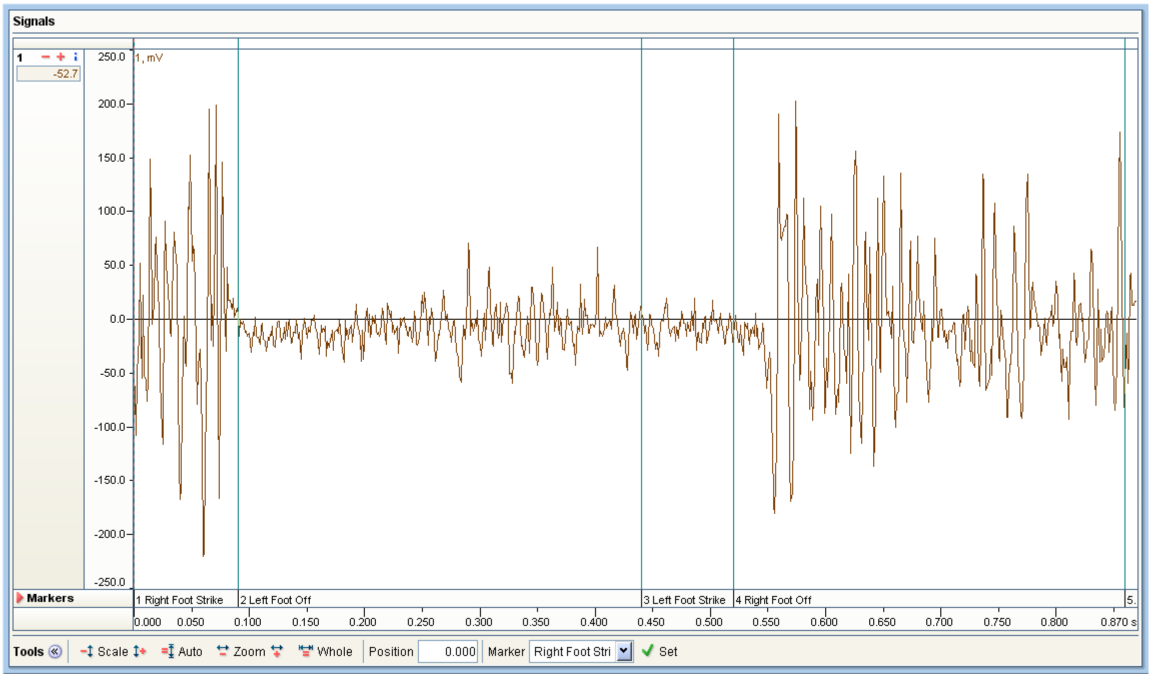
Appendix C

ANOVA Table for the Effect of Footwear on sEMG

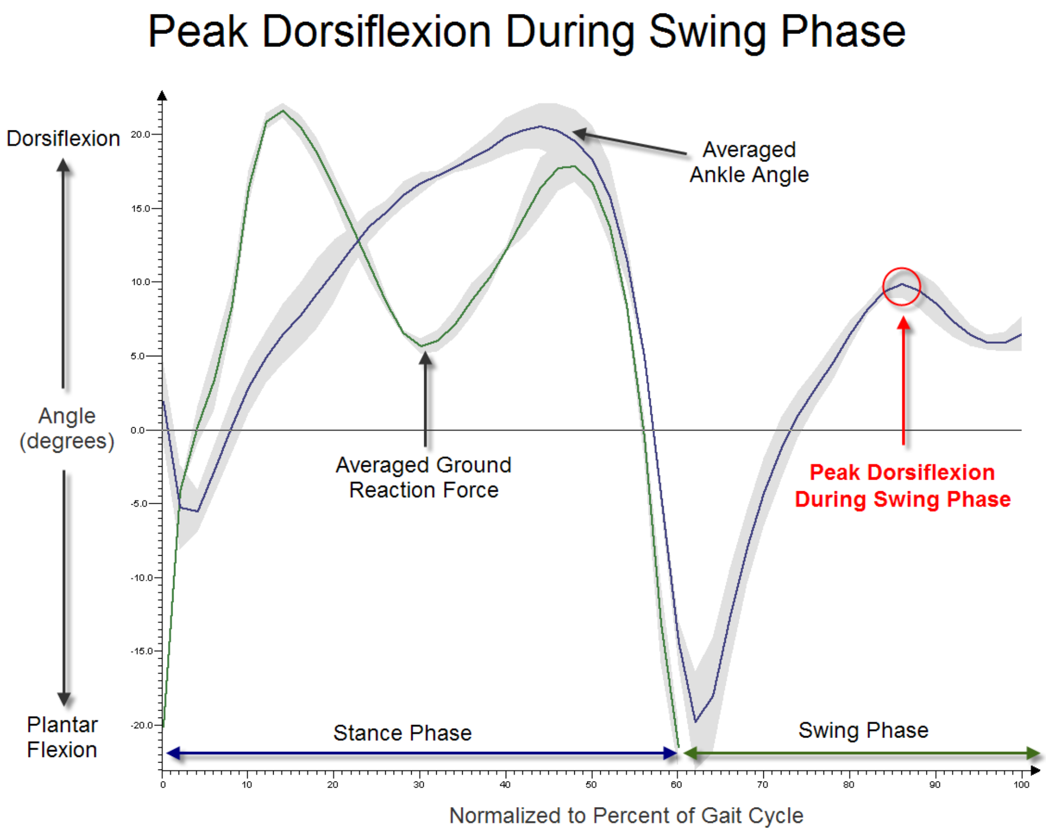
	Mean	SD	*P*	η^2	*Power*
$sEMG_{AVG}$ (mV)					
BF	24.55	7.83			
FF1	27.81	9.66	< .001	.404	1.000
FF2	28.21	8.80			
FF3	28.57	9.36			
$sEMG_{PEAK}$ (mV)					
BF	129.5	54.2			
FF1	145.8	68.3	< .001	.116	.969
FF2	142.3	59.4			
FF3	145.0	63.2			

Appendix D

Raw Surface Electromyography Signal of Tibialis Anterior for One Stride Length

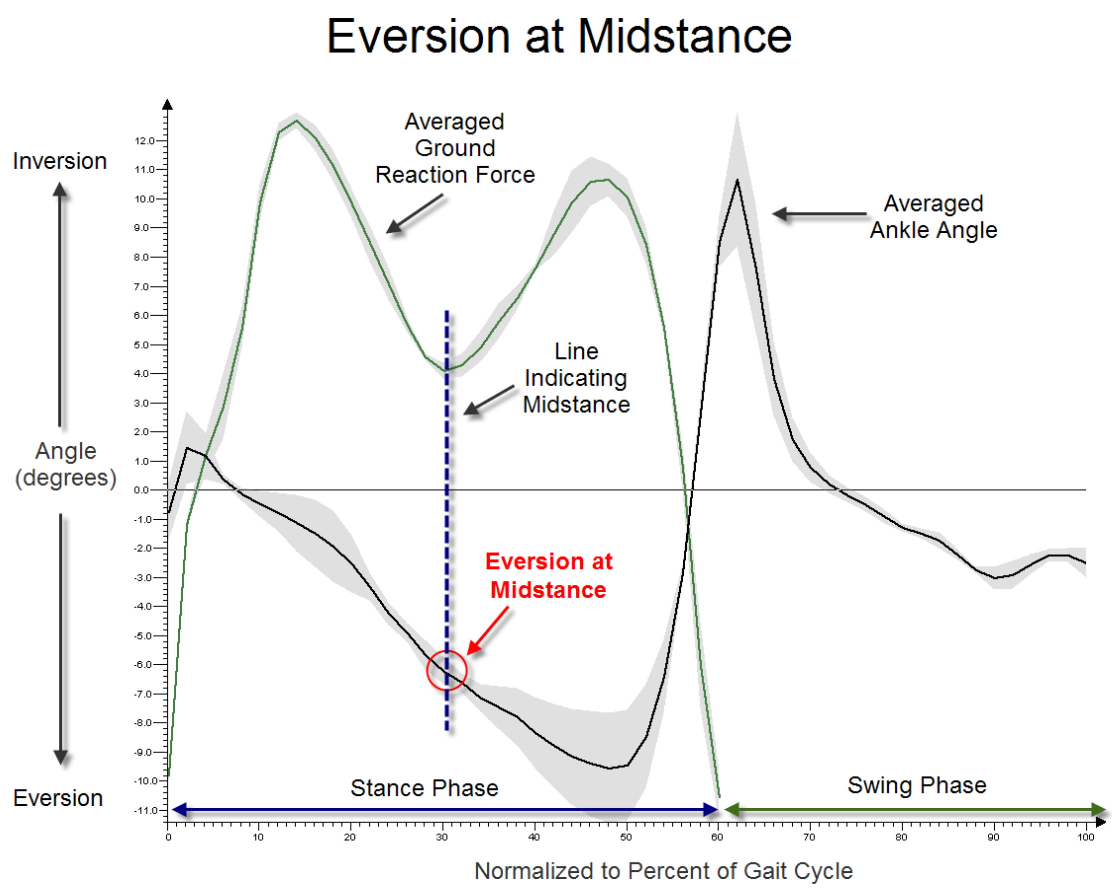


Appendix E

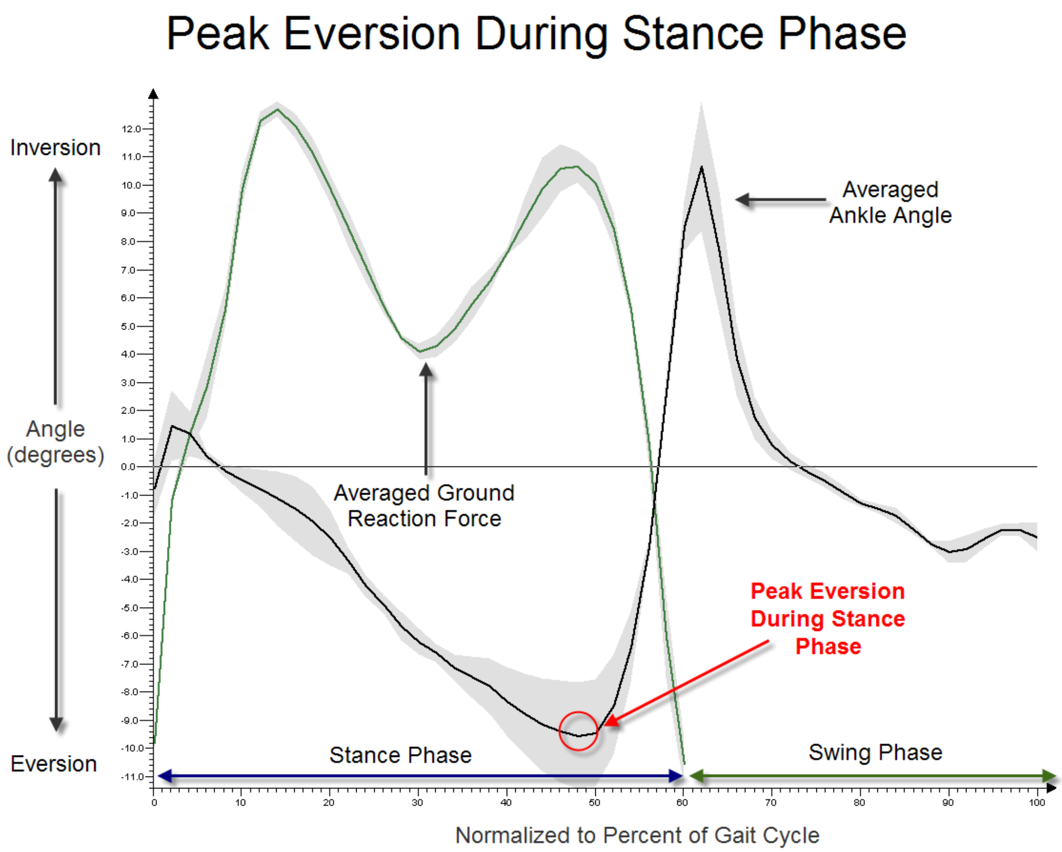


Peak Dorsiflexion During Swing Phase
Dorsiflexion
Angle
(degrees)
Plantar
Flexion
20.0
15.0
10.0
5.0
0.0
-5.0
-10.0
-15.0
-20.0
Averaged
Ankle Angle
Averaged Ground
Reaction Force
Peak Dorsiflexion
During Swing Phase
Stance Phase
Swing Phase
0
10
20
30
40
50
60
70
80
90
100
Normalized to Percent of Gait Cycle

Appendix F

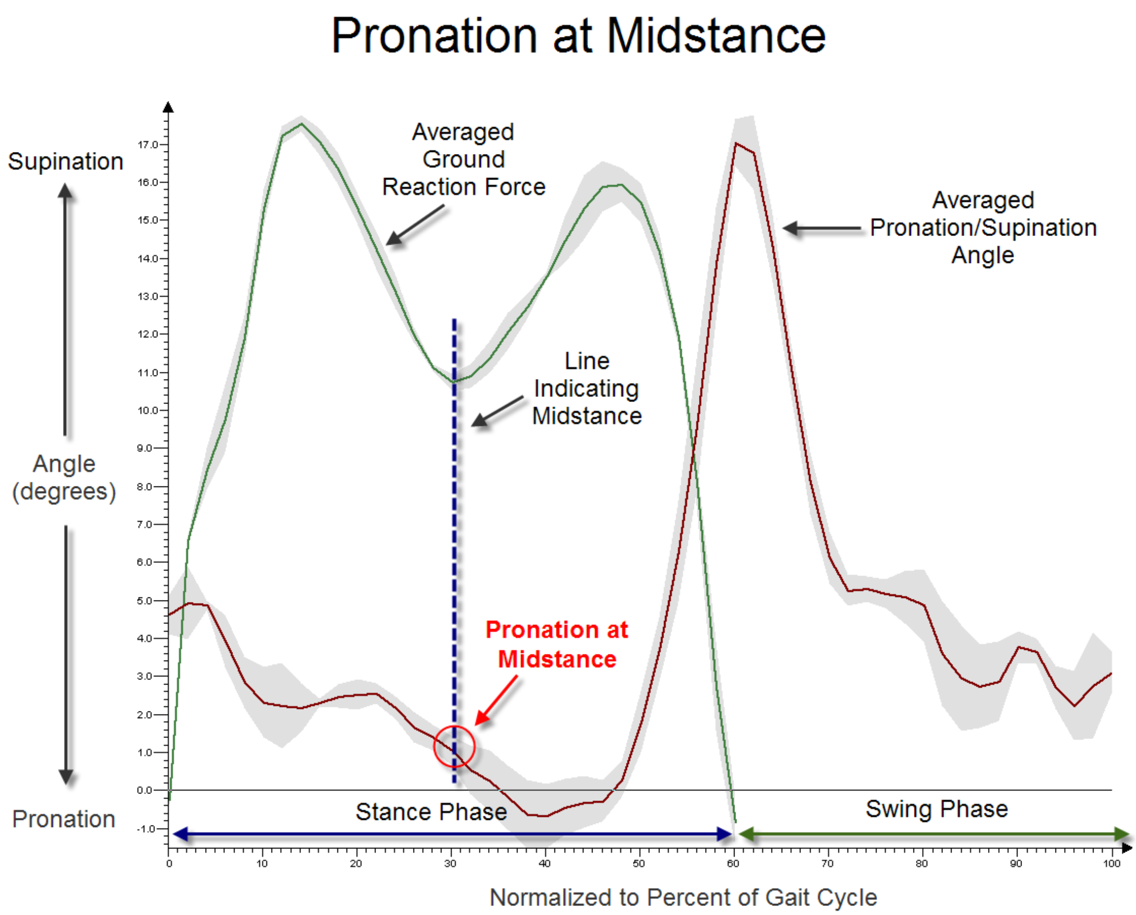

Eversion at Midstance
Inversion
Angle
(degrees)
Eversion
Averaged
Ground
Reaction Force
Averaged
Ankle Angle
Line
Indicating
Midstance
Eversion at
Midstance
Stance Phase
Swing Phase
Normalized to Percent of Gait Cycle

Appendix G



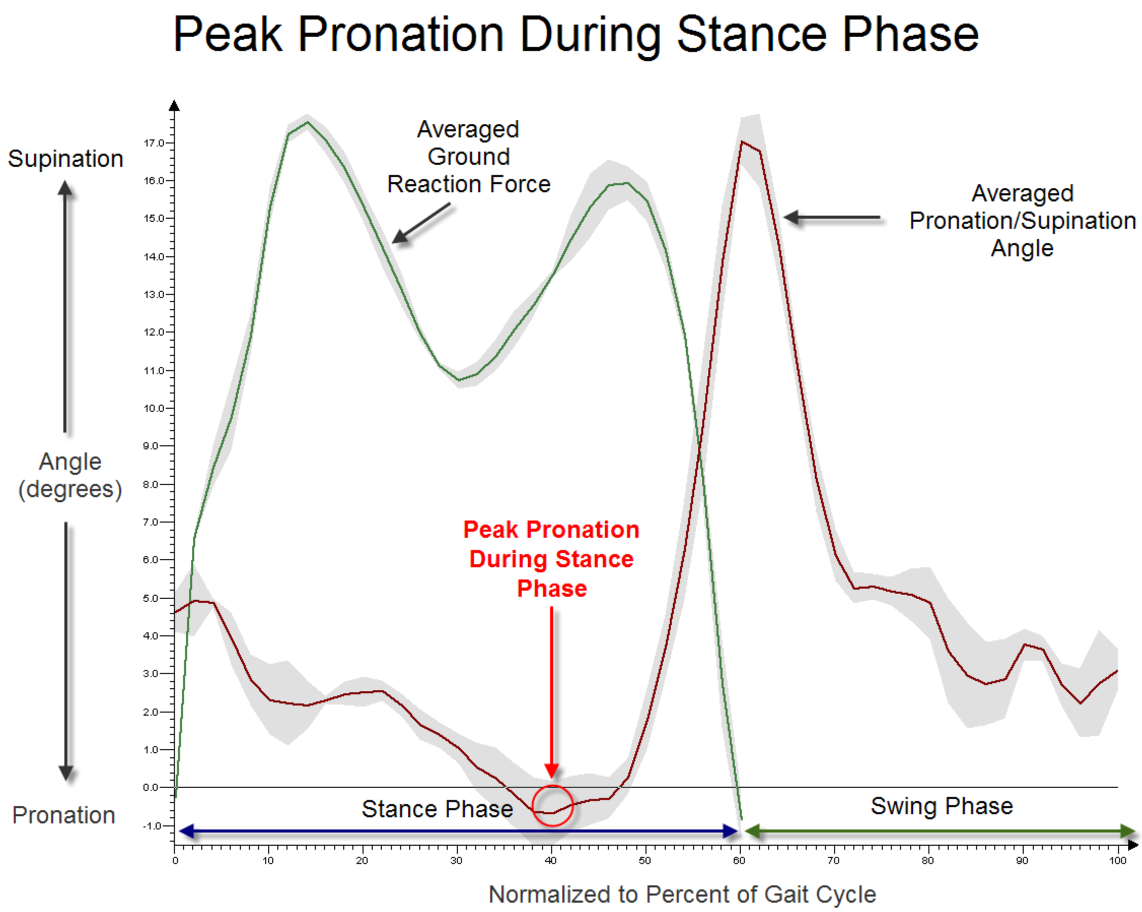
Peak Eversion During Stance Phase
Inversion
Angle (degrees)
Eversion
Averaged Ankle Angle
Averaged Ground Reaction Force
Peak Eversion During Stance Phase
Stance Phase
Swing Phase
Normalized to Percent of Gait Cycle

Appendix H



Pronation at Midstance
Supination
Angle (degrees)
Pronation
Averaged Ground Reaction Force
Line Indicating Midstance
Pronation at Midstance
Averaged Pronation/Supination Angle
Stance Phase
Swing Phase
17.0
16.0
15.0
14.0
13.0
12.0
11.0
10.0
9.0
8.0
7.0
6.0
5.0
4.0
3.0
2.0
1.0
0.0
-1.0
0
10
20
30
40
50
60
70
80
90
100
Normalized to Percent of Gait Cycle

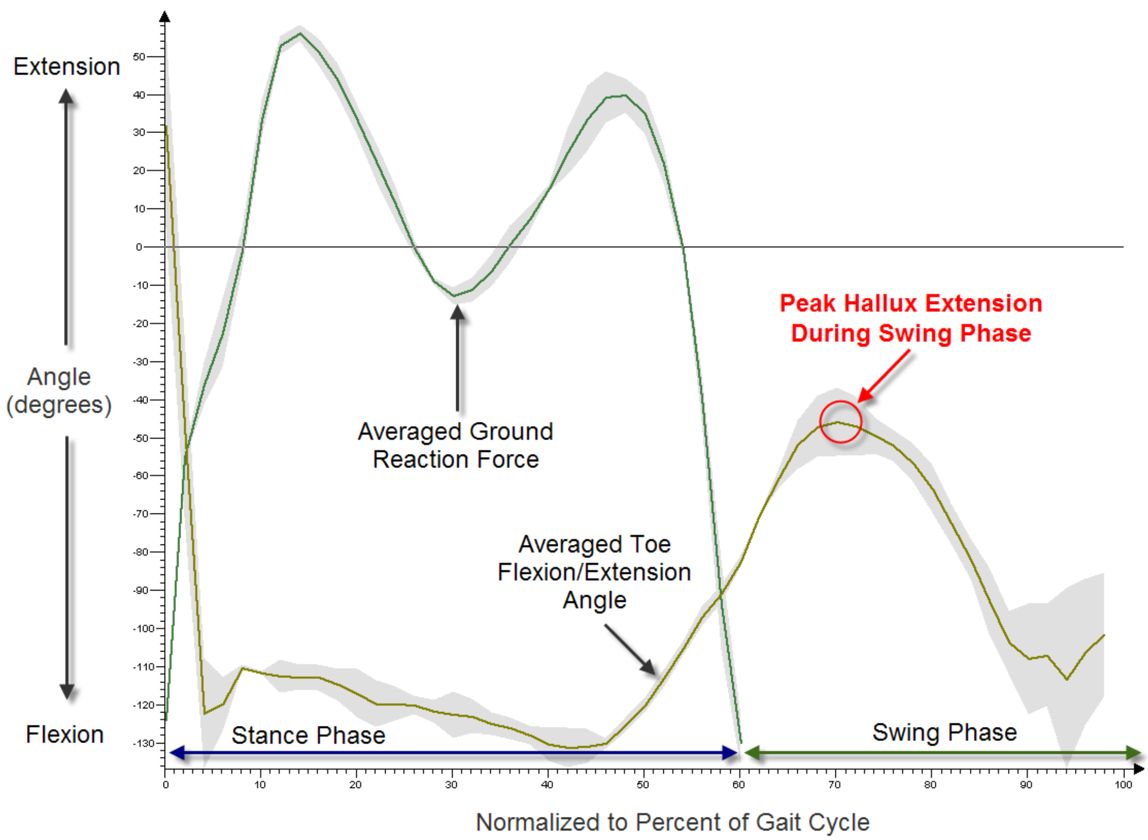
Appendix I



Peak Pronation During Stance Phase
Supination
Angle (degrees)
Pronation
Averaged Ground Reaction Force
Averaged Pronation/Supination Angle
Peak Pronation During Stance Phase
Stance Phase
Swing Phase
17.0
16.0
15.0
14.0
13.0
12.0
11.0
10.0
9.0
8.0
7.0
6.0
5.0
4.0
3.0
2.0
1.0
0.0
-1.0
0
10
20
30
40
50
60
70
80
90
100
Normalized to Percent of Gait Cycle

Appendix J

Peak Hallux Extension During Swing Phase



Appendix K

Institutionally Approved Informed Consent Document

INFORMED CONSENT
for a Research Study Entitled

"Influence of Thong Flip-flops on Gait Kinetics, Kinematics and Lower Leg Electromyography"

You are invited to participate in a study that compares the human gait of people with high and low arches while wearing flip-flops with no, medium and high arch support. Justin F. Shroyer, Dr. Wendi Weimar, Dr. Chip Wade, Joanna Booker and Andrea Sumner are conducting this study. We hope to compare foot motion of individuals with low, normal, and high arches while walking in various types of thong flip-flops. This study may benefit society in general by contributing to the body of knowledge regarding how wearing flip-flops with different structural features affect human gait.

You were selected as a possible participant because you meet the following criteria:

1. Between the ages of 19-25
2. No history of surgery in the lower extremities in the last year
3. No history of injury to the lower extremities within the previous year.

If you decide to participate, we will ask you to report to the Sport Biomechanics Laboratory, room 1127 in the Memorial Coliseum for two separate contact days. The first contact day will last approximately 30 minutes, including paper work. On the second contact day (testing day) you will arrive at the lab and you will perform a maximal volitional isometric contraction (MVIC) for the ankle musculature. This will include raising your toes up as hard as you can for 5 seconds. Then two electrodes will be placed on your shin and one on your knee to record muscle activity. After electrode placement, retroreflective markers (little reflective balls) will be placed on your right leg; 10 on your foot, 2 on your ankle, 3 on your shin, two on your knee, two on your hips, two on you abs and one on your lower back. Next, you will be assigned to a group that will indicate in which order your will be asked to wear and walk in the flip-flops (FF) (FF1, FF2, FF3, FF4). Your gait data will then be collected while wearing the various types of flip-flops as you walk approximately 6 m and strike a force platform at a self-selected pace, in bare feet. Next, three trials of each type of flip-flop will be performed. Between each set of three trials for a flip-flop type, you will be asked to do a barefoot trial. At the conclusion of this test, you will be allowed to keep all four pairs of flip-flops.

There are minimal risks associated with participation in this study. The risks that may be present are similar to that if you were walking in a hallway. Other risks include an adverse response to the adhesive on the electrodes and reflective markers; however, these risks are similar to risks associated with applying athletic tape. Also, the MVIC could result in a musculoskeletal injury such as a strain, sprain or muscle soreness, which are possible in any type of lifting or athletic activity. Though the potential for injury is minimal, in the event of injury resulting from participation in this study, you will be financially responsible for any medical costs incurred through participation in this study. The Auburn University Medical Clinic and/or East Alabama Medical Center will be available for minor risk injuries.

A phone will be available at all times for 911 emergencies. You will be allowed to discontinue participation at any time for any reason without penalty. If you have any questions or problems after you leave the laboratory as a result of your participation in this study, please inform Justin Shroyer (telephone: 334-844-1468; email: shroyjf@auburn.edu).

Any information obtained in connection with this study with which you can be identified will remain confidential. Information collected through your participation may be published in a professional journal, and/or presented at a professional meeting, and if so, none of your identifiable information will be included.

Your decision whether or not to participate in this study will not jeopardize your future relations with Auburn University or the Department of Kinesiology. If you have any questions we invite you to ask them now. If you have questions later, we will be happy to answer them (telephone: 334-844-1468; email: shroyjf@auburn.edu). You will be provided a copy of this form to keep for your records.

For more information regarding your rights as a research participant you may contact the Auburn University Office of Human Subjects Research or the Institutional Review Board by phone 334-844-5966 or e-mail at hsubjec@auburn.edu or IRBChair@auburn.edu.

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

______________________________ __________________

Participant's Name (Printed) Date

____

Participant's Signature

______________________________ __________________

Investigator obtaining consent Date

Appendix L

Participant Timeline for the Research Study Entitled:

Influence of Thong Flip-flops on Gait Kinetics, Kinematics and Lower Leg Electromyography

Visit One: Initial Screening	Time Commitment
Preliminary Medical Questionnaire	
Informed Consent	
Protocol Explanation/Demonstration/Practice	
Foot sizing and Arch Classification	
	30 minutes

Visit Two: Testing	
Kinematic marker and Electrode placement	
MVIC	
Gait testing	
Walk 6 m barefoot	
Walk 6 m flip-flop condition 1	
Walk 6 m flip-flop condition 1	
Walk 6 m flip-flop condition 1	
Walk 6 m barefoot	
Walk 6 m flip-flop condition 2	
Walk 6 m flip-flop condition 2	
Walk 6 m flip-flop condition 2	
Walk 6 m barefoot	
Walk 6 m flip-flop condition 3	
Walk 6 m flip-flop condition 3	
Walk 6 m flip-flop condition 3	
Walk 6 m barefoot	
Walk 6 m flip-flop condition 4	
Walk 6 m flip-flop condition 4	
Walk 6 m flip-flop condition 4	
	60 minutes

Appendix M

Preliminary Medical Questionnaire

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

Participant Number: ___________

YES	**NO**	
_____	_____	1) Are you under the age of 19?
_____	_____	2) Have you ever been told you have an inner ear disorder?
_____	_____	3) Have you ever had lower extremity surgery?
_____	_____	4) Do you presently have any lower extremity disorders?
_____	_____	5) Has your doctor ever said that you have heart trouble?
_____	_____	6) Have you ever had a heart murmur, rheumatic fever or respiratory problems?
_____	_____	7) Has your doctor ever told you that you have a muscle, bone or joint problem such as arthritis that had been aggravated by exercise, or might be made worse by exercise?
_____	_____	8) Have you ever felt faint, dizzy or passed out during or after exercise?
_____	_____	9) Have you ever felt pain, pressure, heaviness or tightness in the chest, neck, shoulders or jaws as a result of exercise?
_____	_____	10) Do you have any reason to believe that your participation in this investigative effort may put your health or well being at risk?
_____	_____	11) Do you need an aid to be able to stand on one leg or walk?
_____	_____	12) Are you currently taking any medication that you think might influence your ability to participate in this study?
_____	_____	13) Are you allergic to adhesives?