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# Foot strike patterns and body alignment effects on muscle activity during running.

Nicole Knapp
*University of Louisville*

# FOOT STRIKE PATTERNS AND BODY ALIGNMENT EFFECTS ON MUSCLE ACTIVITY DURING RUNNING

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Nicole Knapp
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FOOT STRIKE PATTERNS AND BODY ALIGNMENT EFFECTS ON MUSCLE ACTIVITY DURING RUNNING

Submitted by: Nicole Knapp

Nicole Knapp

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(Date)

by the Following Reading and Examination Committee:

Peter M Quesada

Dr. Peter M. Quesada, Thesis Director

Dr. Gina Bertocci

Dr. David Magnuson

Dr. Grady T. Holman

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ABSTRACT

The mechanics of the running stride was altered around the 1970s when shoes were released with an elevated heel (McDougall 2009). At this point in time a heel strike running style became increasingly common and has remained the popular pattern when running in traditional shoes. In fact, the exact same person has been seen to alter their stride when running in a traditional shoe where they heel strike verses barefoot where they take on a forefoot strike (Lieberman 2010).

When an individual heel strikes, the ground reaction force is greater and contains an initial peak which is void in a forefoot strike. During a heel strike, a lot of force is generated in the knee (Kulmala 2013). In fact, it has been recommended to alter a stride from heel strike to forefoot strike if an individual is experiencing knee pain. However, Kulmala identified an increase in stress on the Achilles tendon when an individual utilizes a forefoot strike which leads to a different set of injuries. Another common problem identified while performing a forefoot strike is an increase in the activation time of the gastrocnemius muscles (Ahn 2014).

The first aim of this study was created to identify the difference in muscle activity occurring during a heel strike pattern and a forefoot strike pattern. The second aim is designed to

identify if adding a forward body tilt while performing a forefoot strike will alter the muscles activated in hopes of reducing the amount of activation at the calf.

To assess the aims, a 3D motion capture analysis system as well as an EMG system were utilized. The EMG system was used to monitor tibialis anterior, medial gastrocnemius, lateral gastrocnemius, vastus lateralis, rectus femoris, semitendinosus, and biceps femoris muscle activity. Each of the 10 subjects underwent a testing period in which four different strides were assigned individually. Prior to data capture of each stride, the subject was allotted time to familiarize with the stride followed by two and a half minutes of running on the treadmill in which the last 15 seconds were recorded (EMG and motion capture). The motion capture data were processed to ensure the assigned stride was correctly performed. The EMG data was processed by a full wave rectification followed by a root mean square 50 ms moving window average and then normalized.

During terminal swing, the three forefoot strike conditions had higher gastrocnemius, hamstring, and quadriceps muscle activity compared to the heel strike condition. During late stance, the forefoot strike (FF), forefoot strike with an upright body position (FFU) and heel strike (HS) patterns had greater gastrocnemius activity compared to the forefoot strike with a forward body lean (FFL) condition. During initial swing, the gastrocnemius and hamstring muscles had a greater amount of activity in the FF conditions compared to the HS condition. Throughout early stance, midswing and terminal swing phases, the tibialis anterior had an increase in muscle activation when a HS was performed verses one of the three FF conditions. The tibialis anterior muscle demonstrated the least amount of variance due to its sole responsibility of dorsiflexion which had a greater need in the HS condition consistent with the results presented. The three FF conditions relied on plantarflexion to land toe first which resulted in increased gastrocnemius activity beginning in the initial swing phase and proceeded throughout terminal swing. A greater amount of hip extension in the three FF conditions resulted in an increased amount of activation of the hamstring muscles during terminal swing. Increased activation in the gastrocnemius and hamstring muscles for the FF conditions in terminal swing consequently created a tendency of knee flexion which resulted in an increased amount of activation for the quadriceps muscles for knee extension to occur.

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# l. Introduction

## A. <u>Background</u>

Running plays a different role in a variety of lives everywhere whether it be a form of exercise, a role in a sport, a type of escape, or a sense of adventure. Running has been a part of life for millions of years (Bramble 2004). There is a theory backed by evidence that running has been a part of the human evolution. Bramble and Lieberman identified the type of human running performed as “endurance running.” Endurance running is considered to be unique to humans where the run is at a slower pace, in comparison to other mammals, but for a much longer time. The human species have unique anatomical characteristics not seen in other mammals. Due to the bipedaler nature of the way humans run, the lower legs utilize a spring system. Tendons attach muscle to bone and act as the spring system. The main tendons utilized to reduce the metabolic cost of running are the Achilles tendon, iliotibial tract, and the peroneus longus. Another key feature to this anatomical spring system is the plantar arch. The arch also acts like a spring through its elastic properties and ability to reduce the metabolic cost of running. Through noticing these running and human specific features, Bramble and Lieberman concluded by stating, “Today, ER (endurance running) is primarily a form of exercise and recreation, but its roots may be as ancient as the origin of the human genus, and its demands a major contributing factor to the human body form” (Bramble 2004), (Lieberman 2007).

Chris McDougall wrote a novel titled “Born to Run.” This book was published in May 2009 and began a debate in the running world as the evolution of the shoe is explained. Throughout the novel, McDougall follows a tribe of Tarahumara runners in Mexico. McDougall describes how Nike created a shoe with an elevated heel in the 1970s which introduced a new stride of running where the heel strikes first. Since this initial shoe, more models of shoes followed this design of an elevated heel. McDougall points out the lack of science behind this elevated heel and the “unnatural” strike pattern occurring alongside this type of shoe. In fact, he talks about an ultramarathon race where the Tarahumara runners are given a pair of modern day shoes to run the race in instead of their tire made sandals. Less than half way through the race, the runners were in so much pain they decided to strip their shoes off. They found a tire and created a pair of sandals (no elevated heel) to finish the race in (McDougall 2009).

Another runner in McDougall’s novel is a runner named Ted. Ted visited various doctors who all warned him of the dangers of what running could do to his aging body. He was told there was no hope for him in the sport of running. Custom orthotics were created for him, he purchased high stability shoes and so forth; however, all of this continued to give him problems when he would run. One day he ran to the mailbox without any shoes on and he noticed all his pains were relieved. This started his career of running as he became known for being a barefoot runner (McDougall 2009).

Throughout the various stories McDougall included in his novel, the running world started a minimalist trend with attempts to try to get this “natural” foot strike back. The shoe industry began producing lower drop shoes. The drop size is referred to the height of the heel in comparison to the height of the toe box. A “traditional” shoe, one made by Nike in the 1970s, has a 12 mm drop size (i.e. the heel is elevated 12 mm above the toe box) (McDougall 2009). Altra, a shoe company out of Colorado, created 0 mm drop shoes. Brooks, a shoe company out of Washington, created a "pure series" shoe line featuring 4 mm drop shoes. Hoka, a shoe company out of Colorado, created 4 mm drop shoes. Saucony, a shoe company out of Boston, created 4 mm drop shoes as well as some 0 mm drop shoes. Newton, a shoe company out of Colorado, created shoes that contained lugs under the forefoot to help the runner perform a more forefoot running stride. "Fivefingers" were created to mimic barefoot running as closely as possible where the shoes looked similar to toe socks. Merrell, a shoe company out of Michigan, created 0 mm drop shoes. Inov-8, a shoe company out of the United Kingdom, created 0 mm drop shoes with the rubber outsole created containing a barefoot pattern to mimic being barefoot. All these shoe companies produced these shoes claiming them to give the runner a more "natural" stride where they are more likely to perform a forefoot stride.

## B. Scientific Research

Alongside the shoe industry, research over this "natural" stride issue began to arise as well. Researchers began testing various components via kinematics, kinetics, surveys, etc. Strike patterns were identified with respect to forces. The impact of running surface has been researched. The effect of wearing a modern day shoe has been researched. Muscle activity during different strides has been researched. All of this has been done with hopes of identifying a method to reduce the likelihood of injury.

## 1. **Running Surface**

The reality of modern day life has led to concrete becoming the main running surface. In McDougall's novel, he studied a tribe who did not live with this reality. Research was performed to discover if the running surface altered the shoe necessities of an individual due to the ground reaction forces.

Dixon and colleagues performed a study looking at the effects of the running surface. Six runners ran with a heel strike pattern across a force plate covered with three different surface materials while kinematic and force data were collected. The various surfaces contained an asphalt surface, a rubber modified asphalt surface and a synthetic sports surface. A drop test was also performed on each surface. Throughout this test a 6.8 kg spherical head form was dropped onto the various surface materials from a height of 10 cm. An accelerometer was attached to the head form to obtain peak g. Throughout the analysis, the drop test identified a significant difference in the impact absorbing ability where the rubber modified asphalt surface had the lowest peak g (55) followed by the synthetic surface (105) and finally the asphalt surface producing the highest peak g (300) signifying a low impact absorbing ability. However, the six subjects had varying results as to which surface produced the greatest amount of peak impact force. Through investigation of the kinematic data, it was suggested that "joint movements have contributed to providing cushioning of the impact force, resulting in similar loading rates despite differences in surface compliance" (Dixon 2000).

Ferris and colleagues hypothesized an automatic change in leg stiffness when a runner knew they were about to run on a new running surface. To test their hypothesis, six runners ran on two different surfaces (a soft rubber surface and a hard rubber surface) under four different conditions (completely soft rubber, completely hard rubber, first half soft and second half hard rubber, and first half hard and second half soft rubber) while obtaining force plate and kinematic data. A computer simulation was also created to depict what would be expected to happen if the runner did not change leg stiffness. The computer simulations provided results identifying a biomechanical demand to adjust leg stiffness based on surface. If the leg stiffness did not change, the computer simulation identified a loss of running mechanics throwing the runner off balance. As expected, leg stiffness was altered by the runner as they ran from one surface to the next. Even though leg stiffness adjusted, they also noted that "the only consistent difference between surfaces was a substantial reduction in the initial impact peak on the soft surface" (Ferris 1999). Interestingly, this same reduction in initial impact peak is seen in a forefoot strike pattern (Lieberman 2010).

Hardin and colleagues were interested with how a runner's kinematics changed with regards to footwear, running surface, and duration of the run. To gain understanding, heel strike runners ran in a controlled shoe with two different midsole stiffnesses and over two different surface stiffnesses while oxygen consumption and kinematic data were recorded. The midsole stiffness was seen to solely effect the kinematics at the ankle while the harder surface was seen to correlate to a more extended position of the hip and knee during ground contact with an increased peak angular velocity in all the joints. A harder surface was also found to reduce the oxygen intake. It was concluded that kinematic adaptions occur when midsole hardness is altered as well as when surface hardness is altered (Hardin 2004).

These studies provided information as to the implications of the running surfaces in a heel strike condition. One study identified less force associated with a modified (softer) surface. They also observed an inconsistency from individual to individual shedding light on the probability of joints providing some of the cushioning for the surface (Dixon 2000). Likewise, another study found a significant reduction found in initial impact peak when running on a softer surface (Ferris 1999). Finally, another study interested in kinematic data identified a change in the hip and knee joints due to surface changes. A harder surface brought about an increased peak angular velocity in the joints which could account for the inconsistency found in the Dixon study as the joints adapted to the surface hardness to provide cushioning (Hardin 2004), (Dixon 2000). These studies show that the running surface material impacts the runners stride as the joints are altered to account for the surface. Therefore, the surface needs to be considered and held constant throughout data collection.

## 2. Barefoot Running

McDougall pointed out that relative to running, shoes are a new invention. In identifying this, he made the argument that running barefoot is more natural than running with shoes on. Further, he provided a testimony of Ted who was only able to run when he was barefoot (McDougall 2009). Researchers began to research his phenomena by running tests with barefoot conditions verses shod (wearing shoes) conditions.

Thompson and colleagues researched the effects of stride length while running barefoot verses in a shod condition. Eleven runners went through two sessions of data capture where their stride length and foot condition were altered. The stride length was controlled by various tape strips along the runway where 3D kinematic data was captured along with force plate data. The initial stride length was the individuals preferred stride length and then it was increased and decreased by 5 and 10 percent. When the runners ran barefoot, their preferred stride length was significantly shorter compared to their shod condition. The vertical ground reaction force was also consistently lower in the barefoot condition compared to the shod (Thompson 2014).

A study by Lieberman examined the differences of habitually shod individuals in the US, habitually barefoot individuals in Kenya, habitually shod individuals in Kenya, recently shod individuals in Kenya, and recently barefoot individuals in the US. Each of these groups (except the habitually barefoot runners in Kenya) performed tests in a barefoot and shod condition. When going from shod to barefoot, the running kinematics and biomechanics changed. More individuals displayed a heel strike while in the shod condition verse the barefoot condition. Furthermore, the habitually shod individuals from the US who continued to have a heel strike pattern in the barefoot condition did so with a less dorsiflexed foot meaning it was more parallel to the ground (Lieberman 2010).

## 3. Shod Running

Larson and colleagues were interested in the typical strike pattern of recreational runners. They gathered their data at the Manchester City Marathon where they filmed participants at two locations (10 km and 32 km) along the course. After reviewing all the film, the study identified the stride patterns for 936 runners at the 10 km mark. Of those

936 runners, 832 (89%) of them performed a rear foot strike. 286 of the 936 runners were filmed again at the 32 km mark. Of these 286 runners, 266 (93%) of them performed a rear foot strike at 32 km into the race. Larson further stated, “It is tempting to conclude that the high percentage of heel strikers observed in this sample indicates that this is the way that most humans prefer to run. However, a potentially confounding variable is that nearly all runners observed here (with only a few exceptions) were wearing highly cushioned running shoes with a raised heel” (Larson 2011).

Lohman and colleagues presented a center of pressure map. When a runner utilized a standard running shoe, the initial center of pressure was present near the calcaneus. Opposite of this, when the same runner ran in a minimal shoe or barefoot, the initial center of pressure was identified in the forefoot. This observation implies a difference in stride pattern when wearing a traditional shoe (Lohman 2011).

## 4. **Stride Pattern**

The studies mentioned in the previous sections suggested a difference in stride pattern when barefoot verses shod. To further understand the effects of the stride pattern, researchers performed studies interested in foot strike, muscle activity, and other components involved in stride pattern.

The primary component altered by footwear is the foot strike pattern. There are two main foot strike patterns: a heel strike pattern where the heel hits first and then a forefoot strike pattern where the forefoot strikes the ground first. Some researchers claim there is a mid-foot strike pattern as well where your mid-foot strikes first. The presence of the mid-foot strike pattern has been debated. Some researchers claim it to be its own category whereas others associate it with a forefoot strike pattern.

Lieberman obtained kinematic data such as hip angle, knee angle, ankle angle and so forth. He noted that the kinematic joint angles were similar between a forefoot strike and midfoot strike where as a heel strike produced significantly different angles. He further noted, "The data also highlight the general kinematic similarity between forefoot and midfoot striking, both of which differ from rearfoot striking in a number of respects" (Lieberman 2014). Therefore, the midfoot strike condition is linked into the forefoot strike condition throughout this study leaving two categories of foot strikes: forefoot and heel.

## C. Identifying Foot Strike

The determination of which foot strike pattern is occurring has been done in a variety of methods.

Larson and colleagues assessed the foot strike of participants during a race. They set up a high speed camera on a tripod at two different locations along the race. The camera was set at a capturing speed of 300 Hz. In order to identify foot strike, they simply visually assessed it by looking at the video footage. They were deciphering each foot strike between three possible strike classifications. A forefoot strike was determined if the forefoot hit the ground first followed by the heel. A mid-foot strike was determined if the whole foot (forefoot and heel) hit down together. Lastly, a heel strike was determined if the heel struck the ground first (Larson 2011).

Giandolini and colleagues attempted to identify foot strike pattern using accelerometers placed on the heel and metatarsal. Throughout the experiment, runners ran with their normal strike at their preferred speed and then ran with a forefoot strike, a mid-foot strike and finally a heel strike. Each runner had two accelerometers placed on their foot (one at the heel and one at the metatarsals) as well as a reflective marker over the accelerometer. Kinematic data and accelerometer data were recorded while the individuals ran on the treadmill. The time between the heel and metatarsal peak accelerations (THM) was calculated and was the measurement of interest. Giandolini noted that the THM calculation was only reliable for a heel strike pattern and could further be used to assess if the strike is a heel strike or not a heel strike. The THM calculated was highly correlated to the foot strike angle allowing THM to be an adequate way to identify a heel strike vs a non-heel strike when kinematics are not available (Giandolini 2013).

Hatala and colleagues were interested in the foot strike patterns across a habitually barefoot population. The barefoot population studied were runners in Dassanach near Kenya. To assess strike pattern, a plantar pressure pad was utilized. A heel strike was observed when the initial pressure was solely at the heel area. A forefoot strike was observed when the initial pressure was solely at the forefoot. Finally, a midfoot strike was classified when pressure was distributed between the forefoot and heel at initial contact (Hatala 2013).

Altman and Davis posed a question on the noise associated with force plate data on a treadmill proclaiming the foot strike pattern. Due to this question, they were interested in performing a study to identify if kinematics could be a reliable method to declare foot strike pattern. Each runner ran in a heel strike pattern, midfoot strike pattern, and a forefoot strike pattern with running shoes on. Following these three patterns, they ran one more time in a barefoot condition without a strike pattern assigned. Force plate data was collected as well as kinematic data utilizing a 3d camera set up and reflective markers. The two variables of interest was the strike index compared to the foot strike angle. The strike index was found from the force plate data where the center of pressure was identified as a longitudinal axis value and then was divided by the length of the foot and then multiplied by 100% to essentially identify how far up the foot the individual first had ground contact. Therefore, the strike index increases when you go from heel strike to midfoot strike and is greatest in a forefoot strike. The Foot strike angle is calculated utilizing two markers placed on the forefoot and ankle. The angle produced by these markers at initial contact minus the angle produced during normal stance is found to be the foot strike angle. Altman and Davis found a strong negative correlation between the strike index and foot strike angle with enough difference in between the strike patterns to conclude that strike pattern can be effectively identified using kinematics as a surrogate measure (Altman 2012).

Lieberman performed a study on Tarahumara runners. He was interested in identifying the difference in strike patterns between habitually shod runners and habitually minimal or barefoot runners. To analyze foot strike, he used an angle of incidence. Reflective markers were placed at the calcaneus and at the fifth metatarsal head. Runners were recorded at 240 Hz using a high speed camera. An angle of incidence at initial ground contact greater than 1 degree was considered to be a forefoot strike. An angle of incidence less than -1 degree was considered to be a heel strike. An angle of incidence between 1 degree and -1 degree was considered to be a mid-foot strike. To further validate the strike, a strike index was used where 1 represented a forefoot strike, 2 represented a mid-foot strike, and 3 represented a heel strike. The strike index values were averaged for each individual (Lieberman 2014).

Ahn and colleagues measured foot strike angle similar to the previous two studies mentioned. They used a high speed camera placed at the side of the runner. There was a marker placed at the heel and at the fifth metatarsal head. A horizontal vector was created from the calcaneus marker parallel the floor. Another vector was calculated from the heel marker to the toe marker. The angle made by these two vectors with the horizontal vector as the base line at 0 degrees is identified as the foot strike angle. Throughout the study, Ahn and colleagues determined a forefoot strike to be represented by a foot strike angle of less than 8 degrees positive and a heel strike to be represented by a foot strike angle of greater than 8 degrees. This was further confirmed using planter pressure sensors placed inside the socks (Ahn 2014).

## D. <u>Injury</u>

A study by Daoud and colleagues examined the injury rate with regards to the strike pattern. 52 runners were used to gather data where 36 of them had a heel strike and the other 16 were observed to have a forefoot strike. All of these runners were physically very similar, competitive runners. The runners were consistently diagnosed and watched from August 2006 to January 2011. Injury diagnosis was done by the same athletic trainer throughout the whole study. Injury type was analyzed over four different categories: repetitive injuries, traumatic injuries, forefoot strike predicted injuries, and heel strike predicted injuries. Throughout the data analysis, the results suggested that a heel striker is 2.6 times more likely to obtain a mild injury and 2.4 times more likely to obtain a moderate injury. There was not a significant difference between strike patterns and their relation to obtaining a traumatic injury. As expected, a heel striker was 2 to 4 times more likely to obtain a heel strike predicted injury whereas, there was no significant difference between strike pattern and obtaining a forefoot strike predicted injury. Therefore, this study concluded in suggesting an increase in injury likelihood when using a heel strike verses a forefoot strike pattern (Daoud 2012).

### 1. <u>Problems with Heel Strike Pattern</u>

Ground reaction force (GRF) is a variable of interest with regards to the various foot strike patterns. The GRF is measured using a force plate level to the ground as a runner is running. Lieberman and colleagues found a large difference in GRF patterns when different foot strikes were evaluated. There appeared to be two spikes in the GRF pattern when a heel strike was utilized. During a forefoot pattern, a gradual force slope was observed (Lieberman 2010). This same pattern was found in other studies and is represented in Figure 4 below (Giandolini 2013).

Bus performed a study identifying the biomechanical differences in older aged runners verses younger aged runners. He had two groups of runners who ran across a force plate while kinematic (3D motion analysis system) and kinetic (force plate) data were obtained. The kinematic data allowed assessment of joint angles while the kinetic data allowed assessment of vertical ground reaction forces. Through this study the older population was seen to have a shorter stride length, a slower preferred run pace, less knee flexion, and a higher vertical ground reaction force. Bus performed another trial where both populations had to run at the same speed. Through this trial, he noted that the increase in vertical ground reaction force in the older population was interesting. Further research is needed to determine why the increase is present; however, Bus suggested a loss of shock absorption due to aging could be the cause. He further presented that the heel pad absorbs the most amount of shock when the heel initially strikes and suggested that the heel pad may lose is shock absorption abilities throughout aging. If this were to be the case, would it be wise to transition from a heel strike to a forefoot strike?

Giandolini and colleagues observed the difference in loading rate in a heel strike compared to a mid-foot strike. They observed the vertical ground reaction forces and determined load rate from those values. Through their data they found a 50% decrease in load rate when a mid-foot strike is used in comparison to a heel strike. The heel strike pattern presented an impact peak when the heel immediately hit. This peak vanished when a mid-foot strike was used. They found this trend present in all nine subjects who partook in this study (Giandolini 2013).

Davis and colleagues performed a study where they tracked the injuries of 240 female heel strike runners for 2 years. In conclusion, the top five injuries identified over the two year period were listed. The most common injury was iliotibial band syndrome. The second most common was identified to be anterior knee pain. The third most common injury was a tie between a tibial stress fracture, tibial stress syndrome, and plantar fasciitis (Davis 2010). A study by Pohl et al identified risks associated with the onset of plantar fasciitis. Two main risks were identified being vertical ground reaction force load rates and a lower arch. The load rate component is mentioned prior to be significantly greater in a heel strike pattern (Pohl 2009). A study by Milner and colleagues identified risk factors incorporated with the onset of a tibial stress fracture. Once again, a main risk factor observed is the vertical ground reaction force load rate (Milner 2006).

Kulmala and colleagues performed a study interested in the amount of knee loading when running with a heel strike pattern verses a forefoot strike pattern. 19 female runners were carefully chosen to be the subjects in this study due to their ability to maintain a forefoot strike pattern. These runners were carefully matched with 19 heel strike runners. Kinematic and kinetic data were gathered as the subjects ran on a treadmill. Patellofemoral contact force, patellofemoral stress, knee moments and knee angles were calculated. The heel strike pattern produced a statistically significant increase in patellofemoral stress applied to the knee as well as higher patellofemoral contact force. This finding suggests that a forefoot strike pattern "may decrease the risk of developing running-related knee injuries" (Kulmala 2013).

## 2. <u>Problems with a Forefoot Strike Pattern</u>

Giandolini and colleagues utilized an EMG system to identify the muscle activity of the anterior tibialis, lateral gastrocnemius, vastus lateralis, and biceps femoris. In this study, nine habitually heel strike runners ran with a heel strike pattern and with a mid-foot strike pattern. An increase of muscle activity in the lateral gastrocnemius was identified in the mid-foot strike pattern during pre-activation phase. In this study, muscular pains in the plantarflexors were reported by most of the participants. Giandolini also suggested perceivable problems of adopting a mid-foot strike pattern such as metatarsal stress injuries, shin splints, and muscular and tendon injuries (Giandolini 2013).

Kulmala and colleagues identified an equation to calculate an Achilles tendon force. They utilized a subject pool of 19 forefoot strike female runners carefully paired up with 19 heel strike female runners. Through the results, the data calculated a statistically significant increase of Achilles tendon force in the forefoot strike runners compared to the heel strike runners. This increase suggests a risk of ankle injuries associated with a forefoot strike pattern (Kulmala 2013).

Additionally, during a forefoot strike, the individual hits the ground in a plantarflexed state. As soon as the foot hits the ground, the ankle transitions from plantarflexion to dorsiflexion as the heel lowers to the ground. Throughout this transition, the heel is lowered by the triceps surae muscles as well as the Achilles tendon. The triceps surae muscle is also known as the calf muscle containing the lateral gastrocnemius, the medial gastrocnemius, and the soleus muscles. Therefore, as the heel lowers in a forefoot strike, additional work is needed from the calf and Achilles tendon (Lieberman 2010).

## E. Methodology

There have been a variety of methods used to research the effects of foot strike patterns. A 3D motion capture analysis system has been used to identify foot strike patterns as mentioned above as well as various joint angles. Force plates have been utilized to identify ground reaction forces as well as foot strike identification. EMG sensors have been utilized to assess muscle activity patterns. Surveys have been used to identify correlations between various factors typically with a focus on injury.

### 1. <u>3D Motion System Analysis</u>

A motion capture analysis system is a common method in obtaining kinematic data. There are a variety of components involved in the method including placement areas and joints of interest.

**a. <u>Marker Placement</u>.** Ahn and colleagues placed markers unilaterally on the greater trochanter, lateral knee, lateral ankle, 1$^{st}$ and 5$^{th}$ metatarsal, and the calcaneus (Ahn 2014). Similarly, Lieberman placed markers unilaterally on the greater trochanter, center of the knee, lateral ankle, 5$^{th}$ metatarsal, and lateral calcaneus (Lieberman 2010). Thompson and colleagues placed markers bilaterally on the anterior superior iliac spines (ASIS), posterior superior iliac spine, lateral mid- thigh, lateral femoral epicondyle (knee), lateral mid-shank, lateral malleolus, second metatarsal head, and the calcaneus (Thompson 2014).

**b. <u>Static Data.</u>** Static data is necessary in gaining a baseline set of data. During this trial, additional markers are placed on the medial knees and the medial ankles (Williams 2012).

**c. Joint Angles of Interest.** Blaise Williams and colleagues computed joint angles at the ankle, hip and knee (Williams 2012). Lieberman and colleagues were interested in the knee, ankle, and forefoot joint angles (Lieberman 2010). In another study, Lieberman was additionally interested in the trunk angle (Lieberman 2014). Ahn and colleagues were interested in the foot strike angle, ankle angle, shank angle, and knee angle (Ahn 2014).

**d. Computing Joint Angles.** Lieberman and colleagues identified foot strike angles in relation to a horizontal line stemming from the heel marker (Lieberman 2014). They further identified the ankle angle by computing the angle between a vector stretching from the knee to the lateral ankle and the vector stretching from the lateral ankle to the lateral metatarsal. The knee angle was computed as the angle between a vector stretching from the knee marker to the hip marker and the vector from the knee marker to the ankle marker. The hip angle was computed by the vector stretching from the knee to the hip relative to the horizontal. The trunk angle was measured by computing a vector from the greater trochanter (hip) to the neck in relation to a horizontal line (ground) (Lieberman 2014). Blaise Williams and colleagues computed joint angles utilizing the cardan sequence method (Williams 2012). Ahn calculated the foot strike angle by creating a vector from the heel marker to the 5$^{th}$ metatarsal marker. This vector was computed with respect to a horizontal line or ground. The ankle angle was computed in regards to a 90 degree angle (a straight line in the z direction). The shank angle is determined in regards to a vertical (z) line. The knee angle is calculated relative to a straight leg (an extension of the shank vector). A picture of this method is seen in Figure 3 above (Ahn 2014).

**e. Stride Assessment.** To determine the correct stride assigned a range is calculated. The range found will confirm a difference in the angles of interest between the various strides. This will assist in confirmation that the stride assigned was performed in the correct manner. A range in FSA will be computed between the HS and FF conditions. Additionally, a range in BLA will be computed between the FFU and FFL conditions.

## 2. Electromyography (EMG)

Electromyography detects electrical activity. A surface electrode sensor consists of two sensors where a voltage change is detected across them. When a muscle fires, an action potential travels down the muscle as ions are rapidly being transported. This ion exchange is detected by the surface electrodes as a voltage change. Given that the electrode is placed on the surface of the skin, the signal is presented with a large amount of noise. To combat unwanted noise, an active surface electrode can be used. The Delsys active EMG sensors are based off a differential amplifier concept. Therefore, there are two bars one centimeter apart on the sensor detecting a signal. The difference between the two signals is calculated and then amplified. Therefore, if a signal is consistent at both sites, the differential being amplified will be zero and remain negligent. Noise tends to present itself consistently at each site and therefore is negligent helping to keep a clean signal (Technical Note 101).

**a. Muscles of Interest.** The calf muscles are of high importance. The anterior tibialis is a calf muscle on the front part of the shin. When activated, this muscle constricts and lifts the toe upward toward the ceiling placing the foot into a dorsiflex position. This muscle has previously been investigated in running studies using EMG sensors (Chumanov 2012). The medial gastrocnemius muscle is on the back part of the calf towards the inside of the lower shank. The lateral gastrocnemius muscle is also on the back part of the leg but towards the outside of the lower shank. Both of these muscles contract and place the foot in a plantarflexed position where the toe is pointed more towards the ground. These two muscles have also been studied previously (Chumanov 2012), (Ahn 2014). It is also important to note that these two muscles connect to the heel via the Achilles tendon. Therefore, if problems occur with these two muscles, the Achilles tendon could experience harm as well.

The glut muscles are an important muscle during running. The gluteus medius muscle stops the hip from flexing. Chumnaov and colleagues were interested in the gluteus medius and gluteus maximis throughout their study. Their study was interested in increasing the step rate during a stride and tracking the muscle activity associated. For the glut muscles in particular, they noticed an increase in muscle activity at a higher step rate during the late swing phase of the running cycle. Therefore, the opposite glut could be assisting in the pull through of the foot increasing the step rate (Chumanov 2012).

The quadriceps muscles are also utilized throughout running. The vastus lateralis and rectus femoris assist in pulling the leg up as well as extending the leg. These two muscles have been studied previously as muscles of interest. A previous study done recorded a lack of change noticed in the vastus lateralis when an increased step rate is applied (Chumanov 2012).

The hamstring muscle group is an additional group vital to running. The hamstring muscles act opposite to the quadriceps. Therefore, the hamstrings assist in preventing extension of the leg and would be expected to be most used during pre-swing and mid swing when the leg is bending to pull through to the next foot strike. The hamstring muscles have also been studied before (Chumanov 2012).

**b. <u>Placement.</u>** The surface electrode placement contributes to the accuracy of the signal output. According to De Luca, the electrode should be placed either between two motor points or between a motor point and a tendon insertion. The electrodes should be placed on the belly of the muscle where the longitudinal axis of the electrode is parallel to the muscle as seen below (De Luca 2003). Technical Note 101 presents pictures illustrating the appropriate surface electrode placement.

Furthermore, Delagi and colleagues detailed appropriate places to apply the electrode given the individual muscle (Delagi 1981).

**c. <u>Filtering.</u>** The raw EMG signal goes through the filtering process to rid unwanted noise and obtain a cleaner output. This process contains multiple steps and has been done a variety of ways. Ahn and colleagues filtered their EMG signal first with a second order Butterworth band-pass filter ranging from 20 to 400 Hz and included a 60 Hz notch filter. A band pass filter allows everything inside the low and high frequency to pass through. Therefore, everything in the frequency range of 20 to 400 Hz remained as part of the signal. By applying the notch filter at 60 Hz, everything containing a frequency of 60 Hz was removed from the signal. They then rectified the signal and smoothed their data with a moving average filter. There are two ways to rectify the signal and they were not clear with which method they used. Rectifying the signal in general means having a completely positive signal. A full wave rectification is performed by taking the absolute value of each value. A half wave rectification takes each negative data value and makes it 0 while the positive values remain the exact same. The final piece of data processing performed in the study was a moving window average. A moving window average is performed by finding the average of x amount of data points prior to the data point and x amount of data points following the data point. For example, a moving window average of 100 data points would be calculated by adding the sum of 50 points prior to the data point and 50 points following the data point and then dividing that by 100. In performing a moving window average, the data is smoothed (Ahn 2014).

Nawoczenski and colleagues filtered their EMG signal using a root mean square with a 25 msec moving window average. Processing the data with a root mean square function is another method used for smoothing the data. Similar to the moving window average, a root mean square function with a moving window of 25 msec would be calculated using 12.5 msec before the data point and 12.5 msec after the data point. The data points are squared, added together, and then divided by the number of data points used. The square root of this value is then computed giving the root mean square value for the intended data point. Bartlett and colleagues also used a root mean square method to process their signal (Nawoczenski 1999).

Chumanov and colleagues filtered EMG signals with a full- wave rectification and a low pass filter with a bidirecrional, 6$^{th}$ order Butterworth filter with a cutoff frequency of 50 Hz. The low pass filter allowed only data under 50 Hz frequency to remain as part of the signal. The absolute value of the data points were computed to create the full wave rectification (Chumanov 2012).

**d. <u>Mean EMG activity.</u>** Chumanov and colleagues evaluated EMG activity by means of finding a mean value. They included a baseline (preferred step rate) stride to use for normalization. Therefore, the mean value of the stride of interest was then normalized (divided by) the baseline mean value. The mean value is computed by finding the area under the curve and dividing it by the amount of data points present (Chumanov 2012).

**e. <u>Temporal Characteristics.</u>** Chumanov and colleagues identified mean EMG values in temporal time periods (phases) throughout the gait cycle. The first phase of interest was referenced as the “loading phase.” This phase began at foot contact and lasted until peak knee flexion (0% of the gait cycle to about 15%). The second phase was coined as “preswing/early swing.” This phase represented the 30-50% stage of the gait cycle. 70-80% stage of the gait cycle was declared as the “mid-late swing” phase. 80-90% stage of the gait cycle was identified as “late swing/pre-activation” phase as well as the 90-100% stage of the gait cycle.

Ahn and colleagues also assessed EMG activity in temporal means. Two muscles were being assessed in the study. The information gathered was related to the percent of time in the cycle at which the muscles became active and when they deactivated as well as the amplitude at which they fired at. Therefore, the duration at which each muscle was activated for was determined as well as the firing rate during the activation period. It also assessed where along the gait cycle each of these muscles would become active and inactive (Ahn 2014).

Lohman and colleagues broke the running gait cycle into various phases. The first phase is the stance phase which consists about 35% of the overall cycle. This phase is marked by the event of right foot strike (RFS) through right foot toe off (RTO). This first phase was further broken down into 5 stages. The first stage is the initial contact which occurs immediately when the foot contacts the ground. The second stage is the loading response which occurs as the foot continues to make full contact with the ground. The third stage is the midstance stage which occurs when all the body weight is placed on that foot. The fourth stage is the terminal stance leading into the fifth stage of preswing as the foot gets ready to lift off the ground. The second phase is termed "early float" and occurs when both feet are off the ground leaving the foot that just came off the ground in an initial swing stage. This phase accounts for approximately 15% of the cycle and is marked by RTO through left foot strike (LFS). The third phase is claimed to be the swing phase. The duration of this phase is while the opposite foot is in contact with the ground leaving the foot of interest in a midswing stage. This duration is marked by the event of LFS through left foot toe off (LTO). This phase is about 35% of the entire cycle. The final phase is the late float phase leaving the foot of interest in a terminal swing stage as it prepares for ground contact. This float phase is marked by the event of LTO through RFS as both feet are off the ground. This phase makes up the remaining 15% of the cycle (Lohman 2011).

## F. Current Findings

Giandolini and colleagues looked at the difference between strike pattern and muscle activity of four various muscles. Their results identified a decrease in the tibialis anterior during a midfoot strike compared to a heel strike. This decrease was observed while looking at the entire stride cycle instead of splitting the cycle into various phases. An increase in lateral gastrocnemius activity was observed with a midfoot strike verses a rearfoot strike. This increase was observed only in the pre-activation phase (Giandolini 2013).

Ahn and colleagues measured the activity present at the lateral and medial gastrocnemius with a chronically forefoot strike population, chronically heel strike population, and a shifter population. The data collected signified an increase in calf activation prior to foot strike in the forefoot strike population. The data also presented a similar trend between the lateral and medial gastrocnemius. The deactivation time of the calf muscles did not depend on the stride. Therefore, when running with a forefoot strike, the calf muscle is activated for a longer duration in comparison to a heel strike (Ahn 2014).

## G. Purpose

Although all the research mentioned above is leading the way to further understanding, there are still holes missing. As mentioned previously there are many benefits to running without a heel strike pattern. The ground reaction force graphs identified two major force spikes when the heel hits first compared to the gradual force increase seen in a forefoot strike. There was also a greater force displayed on the knee when a heel strike pattern is utilized (Kulmala 2013). However, taking on a forefoot strike has been seen to have problems as well. One of the main issues seen is linked to the Achilles tendon and calf muscle overuse. The recent study by Ahn and colleagues utilized surface electromyography sensors to study muscle activity patterns. The only muscles studied here were the lateral and medial gastrocnemius muscles (Ahn 2014). To gain further understanding of muscle activity occurring during various strike patterns, muscle activity from a more complete muscle set needs to be studied.

Ahn and colleagues also simply presented information related to muscle activity for a heel strike pattern and a forefoot strike pattern without mention of body lean (Ahn 2014). It is hypothesized that adding a forward tilt in conjunction with a forefoot strike will engage different muscles and reduce the muscle activity going on in the calf. Transferring the muscle activity into larger muscle groups will reduce the onset of fatigue allowing runners to run longer and reduce the chance of calf strain. In order to declare that the forward tilt really transfers the muscle activity, an EMG system needs to be utilized and record the activity of the muscles of interest. Therefore, 7 muscles unilaterally (right side) will be observed to gain further understanding of their activity patterns during different stride patterns. The first part of this study is used to investigate the different muscle pattern activities used in a heel strike (HS) versus a forefoot strike (FF). After compiling this data, a forefoot strike with a forward body lean (FFL) versus a forefoot strike with an upright torso stance (FFU) is assessed in hopes that adding a forward body tilt will alter the muscle activity pattern to preserve the calf muscles.

## H. Specific Aims

The first aim of this study was created to identify the difference in muscle activity occurring during a heel strike pattern and a forefoot strike pattern. The second aim is designed to identify if adding a forward body tilt while performing a forefoot strike will alter the muscles activated in hopes of reducing the amount of activation at the calf.

1. To determine the lower extremity muscle activity patterns during a heel strike running pattern and a forefoot strike running pattern.

   H1: A forefoot strike pattern will result in a different muscle activity pattern.

   H2: A forefoot strike pattern will result in altered muscle activity magnitudes.

2. To determine the effects of forward body tilt on lower extremity muscle activity patterns while running with a forefoot strike pattern.

   H1: A forward tilt will result in a different muscle activity pattern.

   H2: A forward tilt will result in altered muscle activity magnitudes.

# ll. METHODS

## A. Subject Recruitment

### 1. Process

Subjects were recruited via word of mouth. The subjects were runners who were associated with the University of Louisville or from a local running store's running group. The participants in this study had to be within a 15 year age range which ranges from 20 to 35 years old. They had to run at least 8 miles a week (Ahn 2014) for at least 3 months prior to data capture (Chumanov 2012). All participants needed to be injury free for the past 3 months. No one who had undergone hip, knee, or ankle surgery were allowed to participate in this study (Chumanov 2012) due to the likelihood of compensation. A survey was given to verify similar subjects.

From the subject pool, ten subjects were used for the analysis. This subject population was validated by running a Paired t test power analysis. Ahn and colleagues presented averages and standard deviations for the overall stride cycle (magnitude) for their four different conditions (different speeds). Using the standard deviations calculated by Ahn and colleagues as well as the differences calculated in the preliminary results for the conditions in this study, values were obtained to run a power analysis. The Paired t test power analysis was run with a maximum difference of 30 (found in the preliminary results for this study), a power of 0.8, a standard deviation of 30 (Ahn 2014) and an alpha value of 0.05. Based on these values, the population size was calculated to be 10.

**2. Survey**

The participant survey can be located in Appendix I. The questions presented in this survey were used to gain understanding of the subject and to group subjects together. The similarities between subjects will make the data gained stronger.

## B. Session

At the start of the session, reflective markers were placed on the subject in anatomical places which remained in place for the duration of the session. The markers were utilized in conjunction with a motion capture analysis system for kinematic data capture. To assess the muscle activity pattern, a Delsys EMG system was used along with the motion capture analysis system. Electrodes were placed on the participant in specific locations to obtain muscle activity data.

The subject was assigned a total of four different running styles. To assess the first aim mentioned previously, the strike the participant stated as their preferred strike was performed first. For example, if a participant claimed to run with a heel strike pattern (FIGURE 1), then a heel strike pattern was assigned first. Secondly, a forefoot strike pattern (FIGURE 2) was assigned if a participant claimed a heel strike pattern to be their preferred strike. Initially, the instructions given to the subjects upon using the forefoot strike pattern was simply to land on the forefoot first without mention of a body lean. During the final two running styles, the subject was asked to use a forefoot strike pattern.

One pattern included directions of maintaining a vertical torso position (i.e. no forward body tilt, FIGURE 3). The other pattern included directions to consciously maintain a forward body tilt (FIGURE 4).

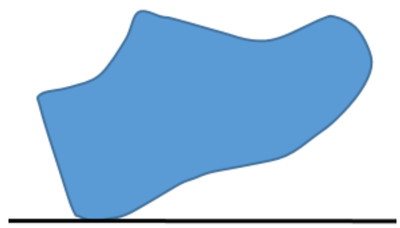

FIGURE 1: Foot Orientation at Initial Ground Contact for a Heel Strike (HS) Pattern

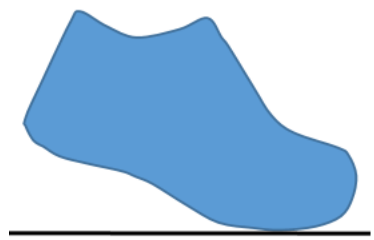

FIGURE 2: Foot Orientation at Initial Ground Contact for a Forefoot Strike (FF) Pattern

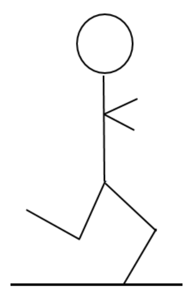

FIGURE 3: Body Posture for a Forefoot Strike with Upright Stance (FFU)

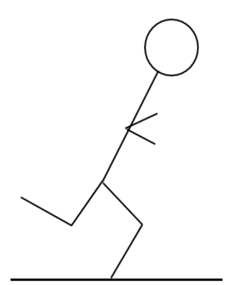

FIGURE 4: Body Posture for a Forefoot Strike with Forward Body Lean (FFL)

Each running style was assigned one at a time. Once the subject was assigned the specific running style, the participant was allotted time to get used to the strike pattern by running up and down a hall for 3 minutes (Pohl 2010). Following this, they were asked to enter the lab and begin the kinematic and kinetic data collection on a treadmill. The kinematic data was used as a surrogate measure to confirm the runner is indeed using the foot strike pattern assigned (FF or HS) by utilizing the results of a study by Ahn and colleagues (Ahn 2014). The EMG data was used to gain muscle activity data.

Data collection occurred on the treadmill. The treadmill speed was chosen by the individual runner as a “comfortable” pace. To ensure consistency, the speed per individual was the same for each running style. The subject ran on the treadmill for a maximum of 3 minutes. The sole purpose of this time span was for the subject to further accustom to running on the treadmill with the intended strike pattern followed by data collection (2 minutes of running followed by 15 seconds of data collection). Additionally, while the participant was on the treadmill, stride encouragement was given every 15 seconds to reinforce the assigned stride.

Once data collection was complete for the first running style, the subject was assigned to the second one and repeated the steps of becoming accustomed to the stride followed by data capture. This cycle repeated itself for all four running styles.

The end of the treadmill data capture for the fourth running pattern concluded the session. At this point, the reflective markers as well as the EMG electrodes were removed and the subject was dismissed.

## C. 3D Motion Analysis System

Kinematic data was gathered by means of a 3D motion analysis system. This system consists of 8 different cameras surrounding the capture field. This system is the Hawk 8 camera motion capture system from MotionAnalysis.

### 1. Markers

The markers used in the motion capture analysis portion are spherical. There are 8 cameras used all identifying the markers from different perspectives. By using a spherical marker, each camera identified the marker as a circle from every angle. The centroid of the circle identified is the centroid of the sphere itself. Therefore, the centroid calculated by one cameras perspective of the marker was uniform throughout all eight cameras.

The subject was placed with markers upon the beginning of the session. These reflective markers were secured to the individual via their clothing or, when necessary (due to the markers placement), placed on their skin with hypoallergenic tape and rubber bands. The anatomical markers were placed at the following locations: bilaterally on the lateral mid-shank, lateral mid-thigh, shoulder, lateral knee, lateral ankle (malleolus), second metatarsal head, calcaneus, lateral pelvis (posterior superior iliac spine), and ASIS. An additional marker was placed on the sacral bone midline on the body and one offset marker was placed near the left shoulder. This marker placement is similar to the marker placement used by Thompson and colleagues; however, the shoulder markers, offset marker, and sacral marker are additional (Thompson 2014). These anatomical markers are for the dynamic data collection.

## 2. <u>System Calibration</u>

The motion capture analysis system was calibrated at the beginning of the session. The calibration process involved two main steps. The first calibration step involved an angled frame. There are four markers present on this frame with known distances apart. The frame was leveled with the floor so that nothing was present in the z- axis. This is a static calibration as the frame is simply left in center of the collection floor to change pixels into accurate numeric distances. Once this calibration step was complete, the angled frame was removed from the capture area and the second step began. The wand calibration is a dynamic calibration. The wand has three markers on it all in a line. During this 60 second calibration, the wand was moved around in the capture area gaining information in the x, y, and z-planes. The end of this calibration step completed the system calibration.

## 3. <u>Static Calibration</u>

The subject was placed with markers upon the beginning of the session. The static calibration was performed (Williams 2012), (Pohl 2010) at the start of each session. During this calibration, markers were placed upon the subject in all the locations previously mentioned as well as on the medial knee and the medial ankle. The subject was then directed as to where to place their feet as they stood as tall and still as possible. Therefore, all angles were claimed to be 0: the knees are fully extended, the hip angle is fully extended, the arms are fully extended and the ankles are in their normal position. This allows a segment coordinate system to be defined and used throughout this session maintaining consistency of the results. For example, if the body lean angle was calculated to be -2 degrees in the static trial, then every value calculated for results would subtract -2 from the result value.

#### 4. Surrogate Data

A study by Ahn et al identified a kinematic pattern consistent with different strike patterns. The strike pattern was verified by utilizing force plate data while simultaneously recording kinematic data using a motion capture analysis system. The kinematic patterns (joint angles) were significantly different between a heel strike pattern and a forefoot strike pattern. The difference identified allows the kinematic data to be a respectable means to verify the correct strike pattern is being performed (Ahn 2014). Therefore, the kinematic data was used as surrogate data in confirming the strike pattern performed. Visual review of the kinematic data was also done to further validate the correct strike pattern occurred.

### D. Electromyography (EMG)

#### 1. Muscle Standardization

The EMG placement was done at the beginning of the session. The sensors were left on for the entire duration of the session in which all four conditions were performed. The motor units that are sensed for activity depend on the exact location of the EMG sensors on the muscle. Therefore, it is vital to standardize the EMG activity each time the sensors are placed on the subject. There are various ways to standardize the EMG sensors; however, using a baseline stride to standardize is ideal for this study.

It has been suggested that a heel strike is detrimental to the knees but utilizes larger muscle groups. The purpose of this study was to identify the muscle activity differences between a heel strike and a forefoot strike. Taking it a step further, a forward lean is then added to the forefoot strike to identify if there is a muscle activity change with the possibility of utilizing larger muscle groups similar to the muscle activity in the heel strike. Therefore, the heel strike was the baseline stride.

Chumanov and colleagues performed a study where they utilized this standardization method. They used a baseline stride and proceeded to calculate the mean muscle activity during the baseline stride. The average of the EMG activity over a full stride cycle was averaged resulting in a single value. They proceeded by dividing all the EMG activity measures throughout the entire stride cycle by this value for each condition (Chumanov 2012).

Utilizing this method, the procedure of data analysis was similar. Upon data collection, the subject ran with a baseline stride being the heel strike. The filtered EMG value calculated for the assigned stride was then divided by the filtered EMG value calculated for the baseline stride as a means of normalization.

## 2. <u>Muscles of Interest</u>

The lateral and medial gastrocnemius muscles as well as the tibialis anterior were the three calf muscles being researched in this study. The soleus is an additional calf muscle in between the lateral and medial gastrocnemius. The soleus is a deeper muscle in comparison to the gastrocnemius muscles which would have hindered the accuracy of the surface EMG output; therefore, the soleus was not a muscle of interest. The anterior tibialis muscle activity was recorded using channels 1 and 8 giving two recordings. In doing so, consistency between channels ensured accurate readings.

The vastus lateralis and rectus femoris were the quadriceps muscles of interest. The feasibility of obtaining surface EMG output for these two muscles has previously been identified (Chumanov 2012). The final two muscles of interest were associated with the hamstring muscles, the semitendinosus and the biceps femoris.

As mentioned previously, an activation of the gluteus muscles were seen to be greater when the step rate frequency increased (Chumanov 2012). Therefore, the gluteus muscles could contribute to the pull through of the opposite leg which would be transferring muscle activity. However, accessing this area violates privacy. For means of this study, the gluteus muscles will not be specifically researched. However, it is important to know the gluteus muscles were seen to be more active in case of possible decreases in other muscle groups activation throughout the study. For example, if a body lean is added resulting in a decrease in gastrocnemius activity, the decrease could potentially be made up by the gluteus muscles not being directly researched (all hypothetical).

### 3. <u>EMG muscle placement</u>

EMG placement varies greatly from subject to subject. To make this process more precise, details were needed as to where the EMGs were to be placed to maintain consistency. Doctors Delagi, Perotto, Lazzetti, and Morrison define where adequate placement of EMGs were to be placed in their book “Anatomic Guide for the Electromyographer” (Delagi 1981).

Anterior Tibialis: Have the subject lying on their back with their legs outstretched. From this position, the placement should be “four fingerbreadths below the tibial tuberosity and one fingerbreadth lateral to the tibial crest.” (Delagi 1981, 159)

Lateral Gastonemii: Have the subject plantar flex the foot while maintaining extension in the knee. From this position, the placement should be “one handbreadth below the popliteal crease on the lateral mass of the calf.” (Delagi 1981, 145)

Medial Gastronemii: Have the subject plantar flex the foot while maintaining extension in the knee. From this position, the placement should be “one handbreadth below the popliteal crease on the medial mass of the calf.” (Delagi 1981, 147)

Vastus lateralis: Have the subject lying on their back with their knee extended. Have them then lift their heel while maintaining the extension in their knee. From this position, the placement should be “over the lateral aspect of the thigh, one handbreadth above the patella.” (Delagi 1981, 193)

Rectus Femoris: Have the patient lying on their back. From this position, the placement should be "on the anterior aspect of the thigh, midway between the superior border of the patella and the anterior superior iliac spine (ASIS)." (Delagi 1981, 181)

Semitendinosus: Have the subject lying on their stomach. From this position, the placement should be "midway on a line between the medial epicondyle of the femur and the ischial tuberosity." (Delagi 1981, 186)

Biceps Femoris (Long head): Have the subject lying on their stomach. From this position, the placement should be "at the midpoint of a line between the fibula head and the ischial tuberosity." (Delagi 1981, 170)

### 4. <u>Muscle Voltage Range</u>

Once the EMG sensors are placed on the subject, it was important to identify appropriate voltage ranges. To do this, the individual was instructed to flex each targeted muscle group individually to identify signal output. If clipping occured, the voltage range was increased to prevent a loss of signal. If the signal was barely identifiable, the voltage range was decreased. This was done prior to data collection for each time the EMG sensors were applied.

## E. <u>Analysis</u>

The analysis component of data is extremely important. The way in which the data was analyzed can potentially alter the results if not understood thoroughly.

Therefore, it was important to lay out how the data was analyzed to give credibility to the results presented.

**1. <u>Kinematic Data</u>**

The kinematic data was assessed for two components. The foot strike angle in reference to the ground was the first component of interest and was used to accurately identify foot strike. The second component was the body tilt angle. This was important in the second half of the protocol where a forward lean and a non-lean were instructed to be performed. Again, this piece of data was used simply to validate the assigned stride was being performed as instructed.

**a. <u>Foot Strike Angle.</u>** The foot strike angle was computed using the right heel and right toe 3D coordinates retrieved from the 3D motion analysis system. The only point of interest was at right foot strike. To calculate the angle, a vector parallel to the z direction was used. Therefore, the vector was as follows: v1=0i+0j+1k or v1=k. The following vector was created as a vector starting at the heel and stretching out to the toe. This was done using equations 1, 2 and 3 to achieve the unit vector for the foot strike angle (U(FSA)). The angle was then created using these two vectors. The dot product of the unit vectors is seen in Equation 4. The unit vectors equal 1. Therefore, the left side of the equation goes to one and is rewritten in Equation 5 to separate out the angle. This angle is actually ϴ(F) in Equation 5 and 6. When the two vectors are crossed, Equation 5 shows the various components that are multiplied. Given that V1 = 1k, V(x) and V(y) are equal to 0 and V(z) is equal to one. That reduces Equation 5 down to Equation 6 where

ϴ(F) is calculated. In order to represent the angle in comparison to ground, 90 minus the angle calculated is the final equation to retrieve the foot strike angle which is seen in Equation 7. Below is a list of equations used to accomplish what is described (Robertson 2004). FIGURE 5 displays this in a diagram.

$$U_{H/T} = (x_T - x_H)\hat{\imath} + (y_T - y_H)\hat{\jmath} + (z_T - z_H)\hat{k} \tag{1}$$

$$\left|U_{H/T}\right| = \sqrt{\hat{\imath}^2_{U(H/T)} + \hat{\jmath}^2_{U(H/T)} + \hat{k}^2_{U(H/T)}} \tag{2}$$

$$\hat{U}_{FSA} = \frac{U_{H/T}}{\left|U_{H/T}\right|} \tag{3}$$

$$\hat{V}_1 \cdot \hat{U}_{FSA} = \left|\hat{V}_1\right|\left|\hat{U}_{FSA}\right| cos\theta \tag{4}$$

$$\theta = cos^{-1}(V_x U_x + V_y U_y + V_z U_z) \tag{5}$$

$$\theta = cos^{-1}(U_z) \tag{6}$$

$$\theta_{FSA} = 90^0 - \theta \tag{7}$$

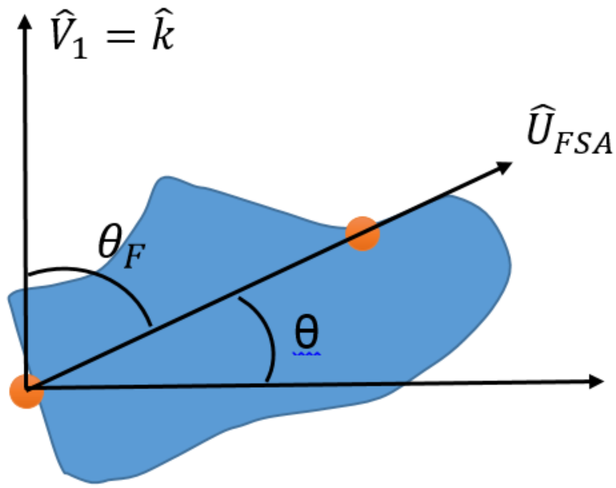


FIGURE 5: Diagram of FSA Calculation

A static measure was also calculated. During the static recording, the subject stood perfectly still. Twenty frames were taken into consideration and are averaged. The average value retrieved identified the angle created due to how the markers were placed on the subject. This static angle value was subtracted from the condition angle found to gain accuracy of the angle seen in Equation 8.

$$\theta = \theta_{FSA} - \theta_s \tag{8}$$

**b. <u>Body Lean Angle.</u>** The body lean angle was computed using the right shoulder, left shoulder, right ASIS, left ASIS, and sacral marker coordinates. To calculate the body lean angle, a vector was created perpendicular to the ground. This vector was again, v1= k. Vector 2 was computed next. First, a midpoint coordinate was created between the right and left shoulder. This new point was simply referenced to as the shoulder midpoint. Next, a midpoint was calculated between the right and left asis markers. This was called the asis midpoint. The midpoint between the new asis point and sacral marker was then calculated giving a virtual point inside the body. This new point was referenced as the body midpoint. The vector(2) was computed going from the body midpoint through the shoulder midpoint. Therefore, V(2) was created using Equations 1, 2, and 3 except the heel marker is the body midpoint and the toe marker is the shoulder midpoint. The dot product of V(1) and V(2) was utilized as seen in Equation 4 (V(2) was used instead of U(FSA)). Once again, the vector components equal 1 and ϴ was separated out in order to calculate it as seen in Equation 5. Equation 6 was then used as the simplified version of Equation 5 due to the fact the V(1) x and y components equal 0 and the z component equals 1. Therefore, the angle was computed as the angle between vectors 1 and 2 and a diagram of this is seen below in FIGURE 6.

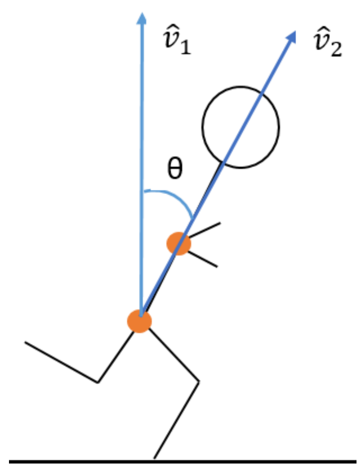


FIGURE 6: Diagram of BLA Calculation

Throughout the entire cycle, the average body lean value was calculated and was the value of interest. Like the FSA, a static recording was done to identify the body lean angle while standing tall and still. This angle took into account 20 frames and is the average over these frames. The average static angle calculated is subtracted from the average cycle angle calculated for each condition seen in Equation 8. Again, this helps the accuracy of the values.

**c. Stride Assesment.** To determine the stride assigned was the stride performed, the FSA and BLA were used. The FSA was used to determine the correct foot placement on the ground. In order to do this, a range was calculated using the minimum HS FSA minus the maximum FF FSA for any of the FF conditions (FF, FFU, FFL). This range was calculated per subject and uses the minimum HS FSA and maximum FF, FFU, and FFL FSA for the five strides assessed for the subject. For this component of the stride to be correctly performed, the range value had to be positive signifying that the largest FF FSA is smaller than the smallest HS FSA.

The BLA was used to determine the correct body positioning during the FFU and FFL conditions. Similarly to the FSA, a range was calculated to assess the subject indeed performed the stride assigned. For this range calculation, the minimum FFL BLA minus the maximum FFU BLA was computed. The range calculation was performed per subject using the minimum FFL BLA and maximum FFU BLA for the five strides assessed for the subject. Again, this range value had to be positive in order to state the stride assigned was correctly performed. This range calculation was used to prove that the BLA during a FFL condition is always greater than the BLA during a FFU condition.

## 2. Muscle Activation

The muscle activation patterns were identified using EMG data. The raw EMG data was retrieved from the myomonitor EMG system synced with the 3d motion analysis system. Based on prerecordings, the EMG channels were set with independent limits to enhance the reading for each channel. For example, the EMG channel for the right lateral gastrocnemius used a range from -0.5 to 0.5 to enhance the reading whereas the channel for the right tibialis anterior used a range from -2.5 to 2.5 to enhance the reading. The EMG signal was obtained as values ranging from 0 to 4096. From these values, the first step was to scale the value back into voltages with their associated range used. For example, the value x that was given by the motion analysis system for the lateral gastrocnemius was multiplied by (0.5-(-0.5))/4096. Therefore, the resulting value was in accurate voltage form based on the channels individual settings.

Next, the average value of all the voltage readings was calculated per channel (muscle). The data was then centered and rectified by taking the absolute value of the voltage value minus the average value calculated (Sarver), (Robertson 2004). The EMG processing chosen for the data was a root mean square filter with a 50 msec interval. This is represented below in Equation 9 (Robertson 2004).

$$RMS = \left(\frac{1}{51}\sum_{t-25}^{t+25} EMG^2\right)^{\frac{1}{2}} \tag{9}$$

From here, the start and end of the foot strike cycle was identified. Therefore, all the root mean square values would be present throughout the entire cycle instead of missing the first and last 25 msec of data. All of the root mean square values were added together to obtain a total amount of voltage per channel throughout the cycle. To represent this data in terms of cycle time, a percentage of cycle was calculated as following: % of cycle = (0.001/(end time- start time))*100. This % of cycle was then multiplied by the total amount of voltage during the cycle. This is also known as an integrated signal (Robertson 2004). Now, the data was able to be interpreted in units of voltage*(% of cycle).

Heel strike pattern EMG values found during the heel strike data capture was utilized as a reference stride. It was calculated as mentioned above. After obtaining the cycle in terms of voltage*(% of cycle), this reference value was used to normalize the other conditions. Therefore, each condition value found was divided by the reference value found. The final value for each condition was in units of percent of reference.

Furthermore, the entire cycle was broken down into five different phases (Lohman 2011). The first phase was early stance (RFS-MS). The second phase was late stance (MS-RTO). The third phase was initial swing (RTO-LFS). The fourth phase was midswing (LFS-LTO). The fifth phase was terminal swing (LTO-RFS). The phase EMG value was the summation of the EMG values in that phase using the root mean square values. The summation of the root mean square values were then multiplied by the % of cycle value found over the entire cycle. Therefore, each phase had its own total. These totals were then normalized to be a percent of the full cycle. For example, the overall summated EMG value found for the entire cycle of a forefoot stride in units of voltage*(% of cycle) is x and the amount of EMG in units of voltage*(% of cycle) the forefoot stride during initial swing is y. The normalized value was calculated as (y/x)*100%, and was the normalized phase value for a forefoot strike during midswing in units of percent.

### 3. <u>Statistical Analysis</u>

For all the results, the statistics were processed using an ANOVA. An ANOVA was performed for each muscle as well as each phase in the cycle and the overall cycle. All ANOVAs were run using a 95% confidence interval. In each ANOVA, the response was set to be the EMG activity values calculated while the factors were designated to be the subject (1-10) as well as the stride (HS, FF, FFT, and FFL). The subject factor was assigned as a random factor while the stride was assigned as a fixed factor and was the main factor being assessed. Subsequently, a Tukey test was utilized to determine which strides are statistically different given each ANOVA.

For any muscle activity where the standard deviation exceeded the mean, non-parametric analysis was necessary due to the non-normality implied. The non-parametric analysis was performed using a Friedman test where the specific phase where non-normality was identified was the response, the four different strides were the treatment, and the subjects were the block. For each Friedman test where a p-value less than 0.05 was calculated, a Wilcoxon sign rank test was utilized to identify the significant differences between the strides.

# lll. RESULTS

## A. <u>Subjects</u>

Thirteen subjects participated in this study. Data from 3 subjects were discarded and 1 data collection was repeated. For two of the subjects, perspiration created a loss of contact with the surface EMGs. By reducing the speed and using a more secure attachment method, one subject was able to repeat the study. The second subject was not able to be rescheduled to repeat the study. Two subjects were not able to produce the correct strides assigned to them. Therefore, data was analyzed for 10 subjects (5 male and 5 female). Of the 10 subjects, 3 claimed to be forefoot runners, 1 claimed to run with a heel strike and 6 claimed to be unknown or midfoot.

## B. <u>Kinematic Data</u>

As mentioned above, two subjects were not able to perform a FF strike pattern in all three FF conditions. Instead, during the FFU and FFL patterns, they reverted back to a HS pattern which both self-identified as their stride of preference. All of the kinematic data was used to assess correct stride performance. Visual assessment also took place and confirmed the correct stride occurred with all 10 subjects.

### 1. <u>Foot Strike Angle (FSA)</u>

For each subject, the range of the minimum HS FSA minus any FF FSA was always larger than 10.2 degrees. Additionally, each FF strike FSA is less than 8 degrees. Each HS angle is greater than 8 degrees.

TABLE I

FOOT STRIKE FSA DIFFERENCES BETWEEN HS AND FF PATTERNS

| Range | | | |
|---|---|---|---|
| Subject | min HS- max FF | min HS- max FFU | min HS- max FFL |
| 1 | 21.58 | 23.87 | 22.47 |
| 2 | 26.80 | 27.16 | 24.98 |
| 3 | 29.78 | 28.36 | 28.96 |
| 4 | 22.38 | 23.87 | 24.41 |
| 5 | 19.13 | 22.66 | 21.72 |
| 6 | 15.66 | 20.91 | 16.61 |
| 7 | 22.11 | 24.57 | 21.44 |
| 8 | 26.29 | 26.81 | 26.18 |
| 9 | 22.65 | 29.02 | 23.25 |
| 10 | 10.22 | 15.04 | 14.71 |

**2. Body Lean Angle (BLA)**

For each subject, the range of the minimum FFL BLA minus the maximum FFU BLA was always greater than 4.15 degrees as seen in Table 14.

TABLE II

MEAN BLA DIFFERENCES BETWEEN FFU AND FFL PATTERNS

| Range | |
|---|---|
| Subject | min FFL- max FFU |
| 1 | 13.28 |
| 2 | 6.96 |
| 3 | 9.59 |
| 4 | 4.15 |
| 5 | 6.44 |
| 6 | 8.85 |
| 7 | 10.32 |
| 8 | 14.70 |
| 9 | 9.15 |
| 10 | 7.87 |

**3. <u>Stride Cycle Figures</u>**

FIGURE 7: Body Posture at RFS

FIGURE 8: Body Posture at MS

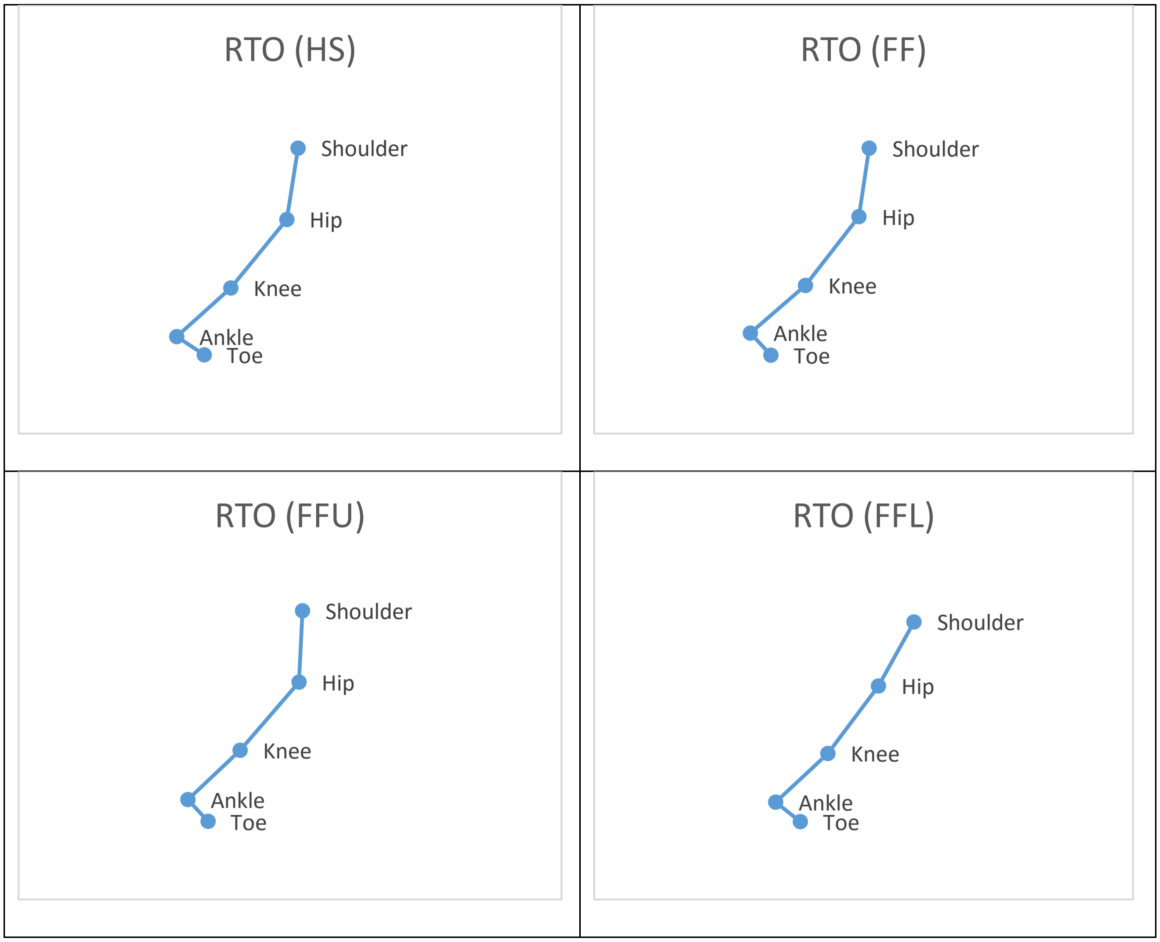

FIGURE 9: Body Posture at RTO

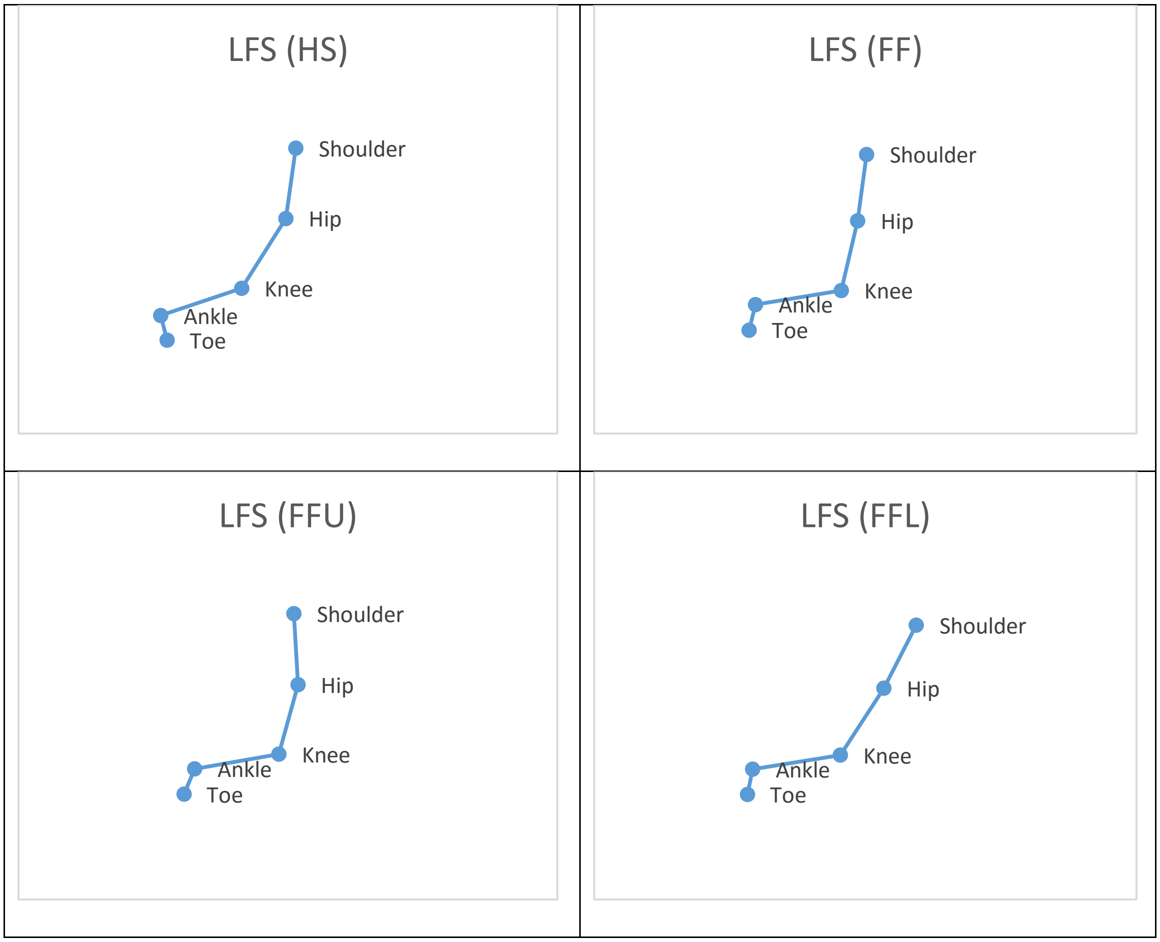

FIGURE 10: Body Posture at LFS

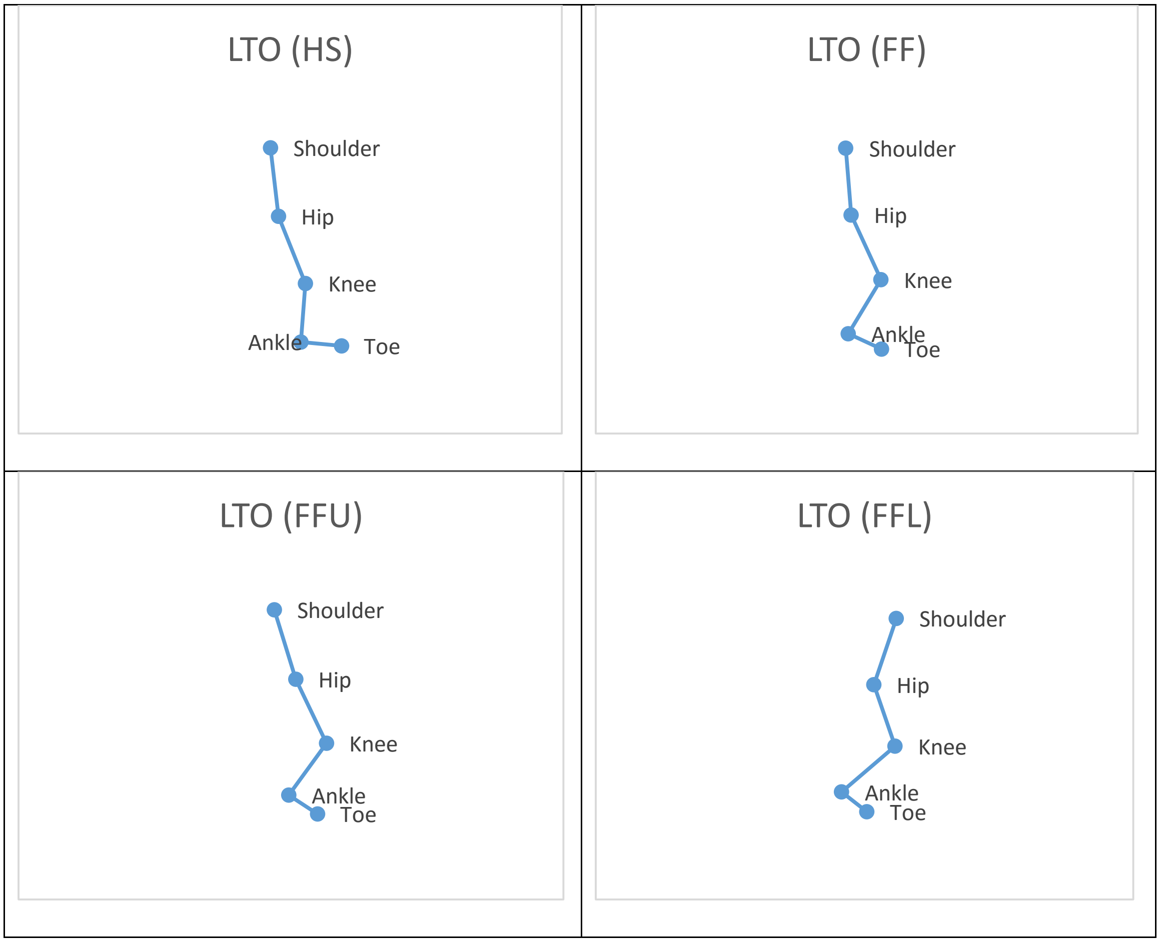


FIGURE 11: Body Posture at LTO

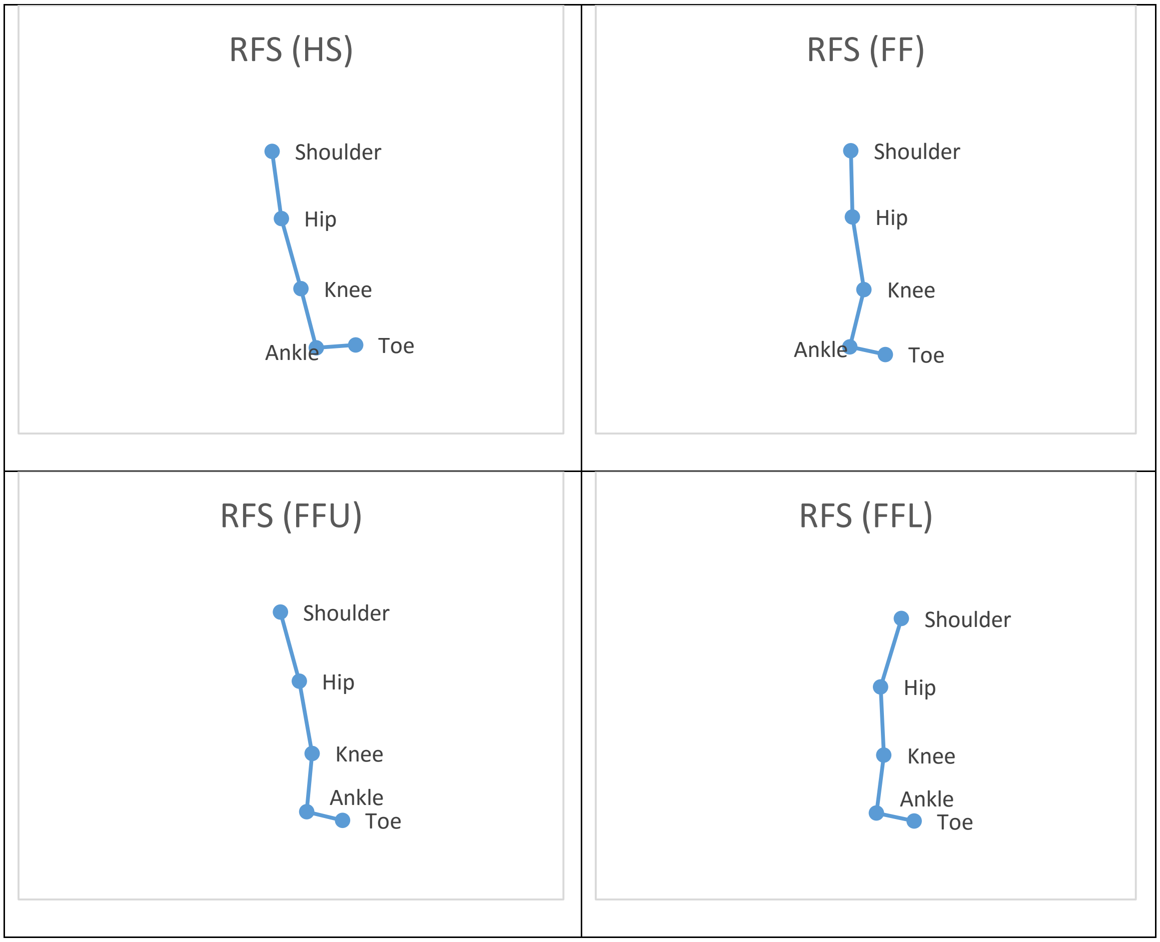


FIGURE 12: Body Posture at RFS, Completion of Cycle

### C. <u>Muscle Activation</u>

All EMG data describes muscle activity in terms of % of reference (HS pattern). Each muscle is analyzed for four phases of a complete stride cycle as well as for the cycle as a whole. Each individual muscle had an n=10 except for the tibialis anterior (2) which had an n=9 due to a lack of signal in this specific channel during testing for one individual.

1. **<u>Tibialis Anterior (TA)</u>**

The HS pattern had significantly ($p<0.001$) higher TA activity compared to the other three stride patterns during early stance, midswing, and the overall cycle. During initial swing, no significant difference was found between the four strides ($p>0.05$). During terminal swing, the HS pattern had a significantly ($p=0.014$) higher activity level compared to the FF condition and the FFL condition. FIGURE 13 provides visual representation of the normalized, integrated EMG in bar graph form while Table III provides the normalized, integrated values.

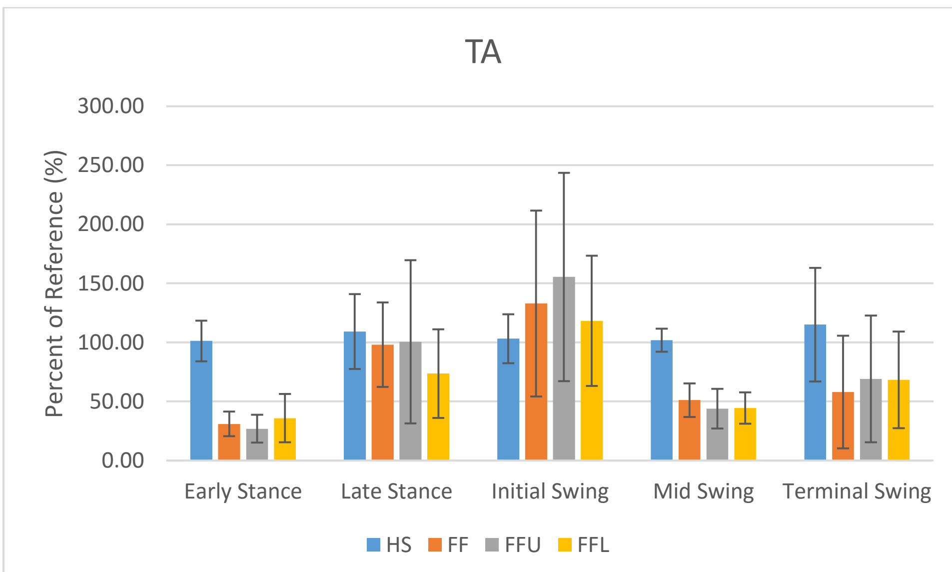


FIGURE 13: Normalized, Integrated TA Activity for Defined Running Phases

TABLE III

NORMALIZED, INTEGRATED TA ACTIVITY FOR DEFINED RUNNING PHASES

| Tib. Ant. | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 101.20 | 109.24 | 103.11 | 101.76 | 115.03 |

| | | | | | |
|---|---|---|---|---|---|
| FF | 30.97 | 97.99 | 132.95 | 51.05 | 57.88 |
| FFU | 26.83 | 100.50 | 155.48 | 43.83 | 69.15 |
| FFL | 35.85 | 73.59 | 118.16 | 44.42 | 68.26 |

## 2. **Tibialis Anterior (2)**

The HS pattern had a significantly ($p<0.001$) higher activity level compared to the other three forefoot conditions (FF, FFU, FFL) in early stance, midswing, and over the entire cycle. No significant difference ($p>0.05$) was found between the strides during initial swing. The HS condition was significantly ($p<0.001$) higher in TA activity compared to the FF stride during terminal swing. In late stance the HS condition was significantly ($p=0.034$) higher in TA activity compared to the FFU condition. FIGURE 14 provides visual representation of the normalized, integrated EMG in bar graph form while Table IV provides the normalized, integrated values.

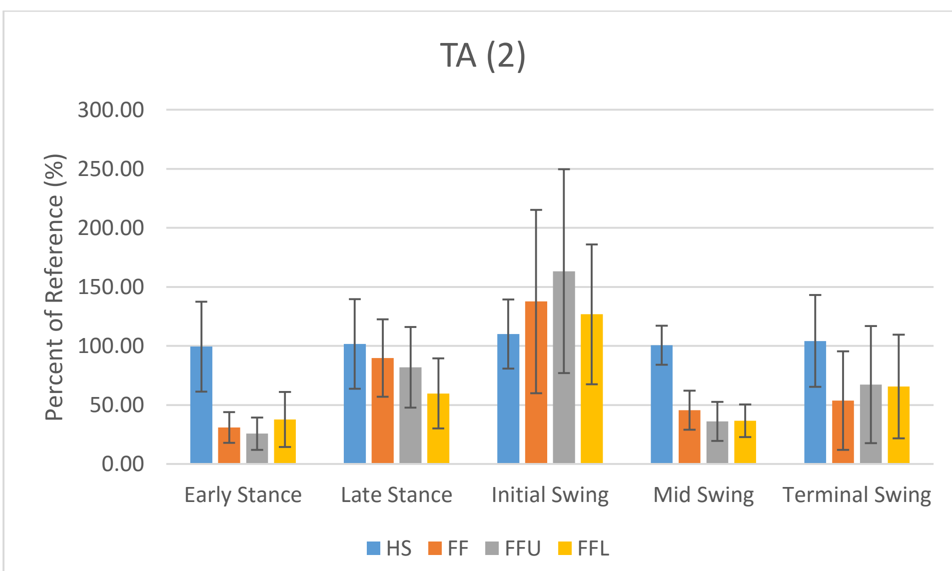

FIGURE 14: Normalized, Integrated TA (2) Activity for Defined Running Phases

TABLE IV

NORMALIZED, INTEGRATED TA (2) ACTIVITY FOR DEFINED RUNNING PHASES

| Tib. Ant (2) | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 99.45 | 101.71 | 110.06 | 100.64 | 104.27 |
| FF | 30.89 | 89.88 | 137.62 | 45.65 | 53.78 |
| FFU | 25.74 | 81.96 | 163.25 | 36.17 | 67.26 |
| FFL | 37.83 | 59.82 | 126.85 | 36.68 | 65.75 |

### 3. <u>Lateral Gastrocnemius (LG)</u>

The three FF conditions had significantly (p=0.001) higher LG activity compared to the HS pattern during terminal swing. During late stance, the HS, FF, and FFU conditions had significantly (p<0.001) higher LG muscle activity compared to the FFL condition. Throughout initial swing, the FF and FFU conditions had significantly (p=0.012) higher LG activity in comparison to the HS condition. No significant difference between conditions was identified for early stance, midswing, or the overall cycle. FIGURE 15 provides visual representation of the normalized, integrated EMG in bar graph form while Table V provides the normalized, integrated values.

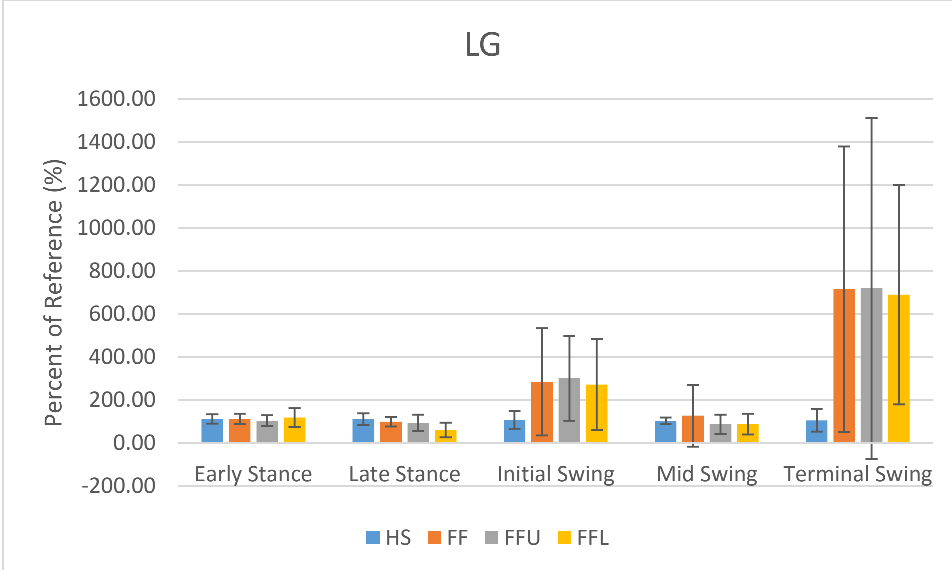


FIGURE 15: Normalized, Integrated LG Activity for Defined Running Phases

TABLE V

NORMALIZED, INTEGRATED LG ACTIVITY FOR DEFINED RUNNING PHASES

| Lat Gastroc | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 111.96 | 110.80 | 107.47 | 102.59 | 105.34 |
| FF | 112.46 | 99.14 | 284.32 | 126.66 | 715.31 |
| FFU | 104.05 | 93.11 | 301.00 | 86.64 | 719.33 |
| FFL | 118.35 | 60.00 | 271.59 | 87.79 | 689.85 |

### 4. <u>Medial Gastrocnemius (MG)</u>

The HS, FF, and FFU conditions had significantly ($p<0.001$) higher MG activity compared to the FFL condition during late stance. During initial swing, the FFU condition had significantly (p=0.022) higher MG activity compared to the HS condition. Throughout midswing and early stance, no significant difference (p>0.05) was identified. Throughout terminal swing, the three FF conditions (FF, FFU, FFL) had significantly (p=0.005) higher activity over the HS condition. Finally, the FF and FFU conditions throughout the entire cycle had significantly (p=0.004) higher MG activity compared to the HS condition. FIGURE 16 provides visual representation of the normalized, integrated EMG in bar graph form while Table VI provides the normalized, integrated values.

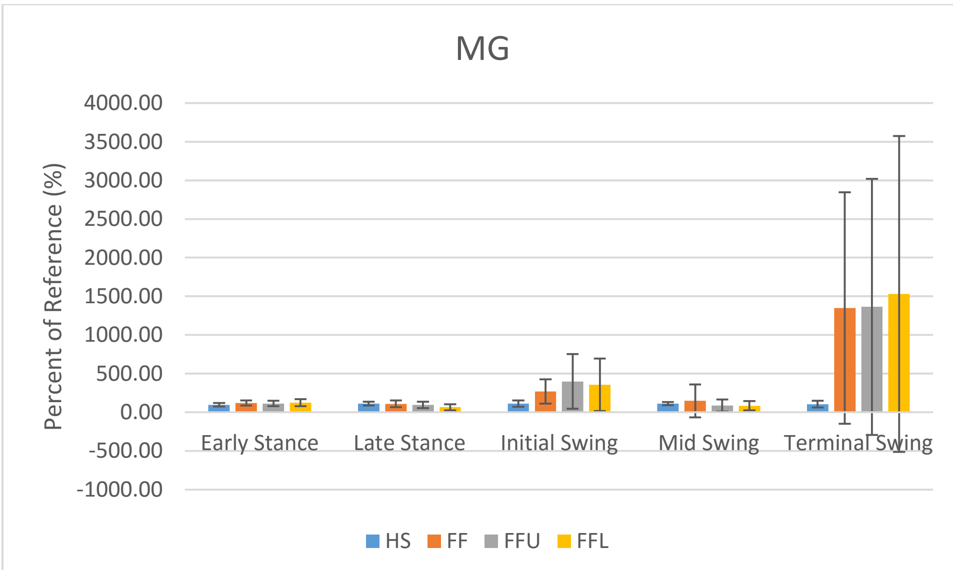


FIGURE 16: Normalized, Integrated MG Activity for Defined Running Phases

TABLE VI

NORMALIZED, INTEGRATED MG ACTIVITY FOR DEFINED RUNNING PHASES

| Med. Gastroc | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 95.45 | 109.36 | 111.68 | 110.24 | 103.78 |
| FF | 119.89 | 108.90 | 269.55 | 146.76 | 1347.13 |

| | | | | | |
|---|---|---|---|---|---|
| FFU | 112.01 | 92.89 | 398.41 | 87.13 | 1362.63 |
| FFL | 123.29 | 63.55 | 354.92 | 83.30 | 1529.97 |

5. **Vastus Lateralis (VL)**

During late stance, the FF condition had significantly (p=0.035) higher VL activity compared to the FFL condition. The three FF conditions had significantly (p=0.028) VL activity compared to the HS condition during initial swing. During terminal swing, the VL activity is significantly (p=0.016) higher in a FF strike compared to a HS condition. The other phases as well as the muscle activity for the overall cycle did not display any significant differences between the four stride conditions. FIGURE 17 provides visual representation of the normalized, integrated EMG in bar graph form while Table VII provides the normalized, integrated values.

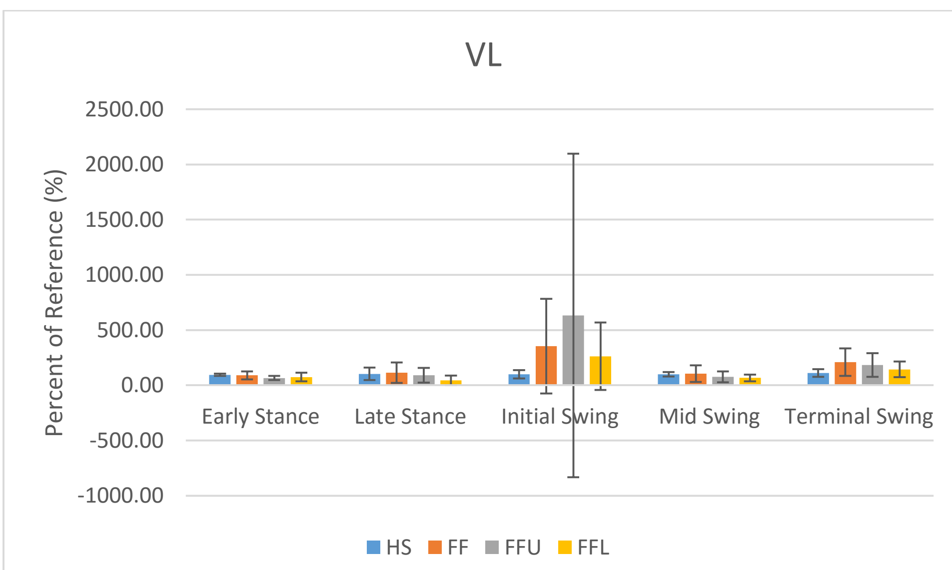


FIGURE 17: Normalized, Integrated VL Activity for Defined Running Phases

TABLE VII

NORMALIZED, INTEGRATED VL ACTIVITY FOR DEFINED RUNNING PHASES

| V. Lat | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 95.37 | 103.97 | 99.88 | 100.36 | 111.80 |
| FF | 89.70 | 113.84 | 353.53 | 104.97 | 210.33 |
| FFU | 66.43 | 90.49 | 632.49 | 76.48 | 184.50 |
| FFL | 75.10 | 45.75 | 262.28 | 66.91 | 144.08 |

### 6. Rectus Femoris (RF)

During late stance, the RF activity is significantly ($p=0.008$) higher in the FF and HS condition compared to the FFL condition. Throughout terminal swing, the three FF conditions had significantly ($p=0.004$) higher RF activity compared to the HS condition. The other phases as well as the overall cycle did not contain any significant ($p>0.05$) differences between the four strides. FIGURE 18 provides visual representation of the normalized, integrated EMG in bar graph form while Table VIII provides the normalized, integrated values.

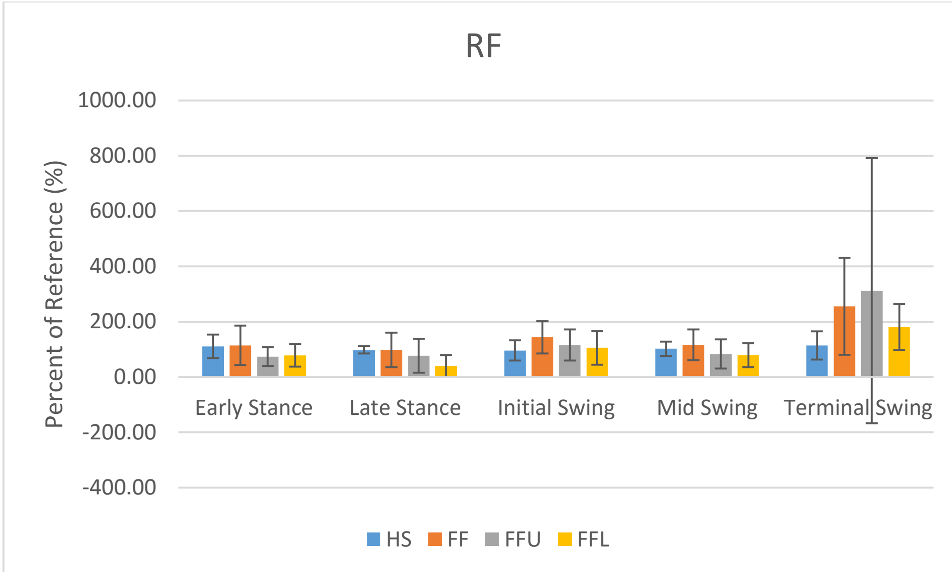


FIGURE 18: Normalized, Integrated RF Activity for Defined Running Phases

TABLE VIII

NORMALIZED, INTEGRATED RF ACTIVITY FOR DEFINED RUNNING PHASES

| Rect Fem | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 110.09 | 98.17 | 95.96 | 101.81 | 114.38 |
| FF | 114.38 | 97.70 | 143.92 | 116.00 | 255.48 |
| FFU | 73.78 | 77.21 | 115.59 | 83.27 | 311.74 |
| FFL | 78.40 | 39.95 | 105.47 | 79.03 | 181.08 |

### 7. **Semitendinosus (ST)**

During early stance, the ST activity is significantly (p=0.007) higher in the FFL and HS conditions over the FFU condition. During initial swing, the FFU condition has significantly (p=0.028) higher ST activity compared to the FF and HS conditions; additionally, all of the FF conditions have significantly (p=0.028) higher ST activity compared to the HS condition. Throughout midswing, the HS condition has significantly (p=0.016) higher ST activity compared to the FFL condition. During terminal swing, all three FF conditions have significantly (p=0.002) higher ST activity compared to the HS condition. The late stance and overall cycle do not contain significant differences between the four conditions in regards to ST activity.  FIGURE 19 provides visual representation of the normalized, integrated EMG in bar graph form while Table IX provides the normalized, integrated values.

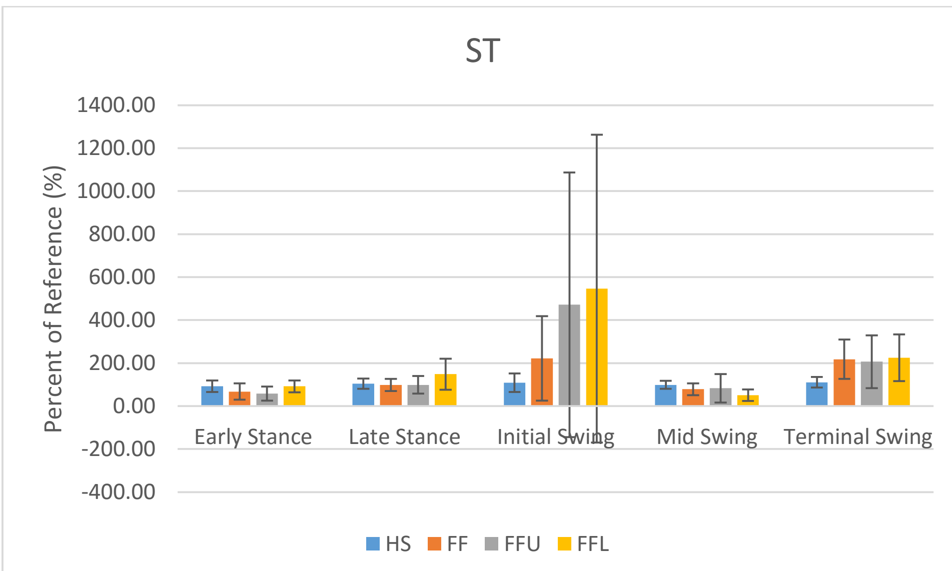


FIGURE 19: Normalized, Integrated ST Activity for Defined Running Phases

TABLE IX

NORMALIZED, INTEGRATED ST ACTIVITY FOR DEFINED RUNNING PHASES

| Semitend | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 92.25 | 104.75 | 108.67 | 98.96 | 110.52 |

| | | | | | |
|---|---|---|---|---|---|
| FF | 67.05 | 98.64 | 221.73 | 78.66 | 218.09 |
| FFU | 57.49 | 98.75 | 471.53 | 83.10 | 206.41 |
| FFL | 91.82 | 148.20 | 546.83 | 49.86 | 224.57 |

**8. Biceps Femoris (BF)**

In early stance, the BF activity is significantly ($p<0.001$) higher in the FFL condition compared to the FFU and FF conditions. Throughout initial swing, the FFL condition had significantly ($p=0.011$) higher BF activity compared to the FF and HS conditions; additionally, the three FF conditions had significantly ($p=0.011$) higher BF activity compared to the HS condition. During midswing, the HS pattern had a significantly ($p=0.004$) greater amount of BF activity compared to the FFU and FFL conditions. The BF activity is significantly ($p=0.002$) higher in the FFL condition compared to the HS condition during terminal swing. For the overall cycle, the BF activity is significantly ($p=0.019$) higher in the FFL condition compared to the HS and FFU conditions. FIGURE 20 provides visual representation of the normalized, integrated EMG in bar graph form while Table X provides the normalized, integrated values.

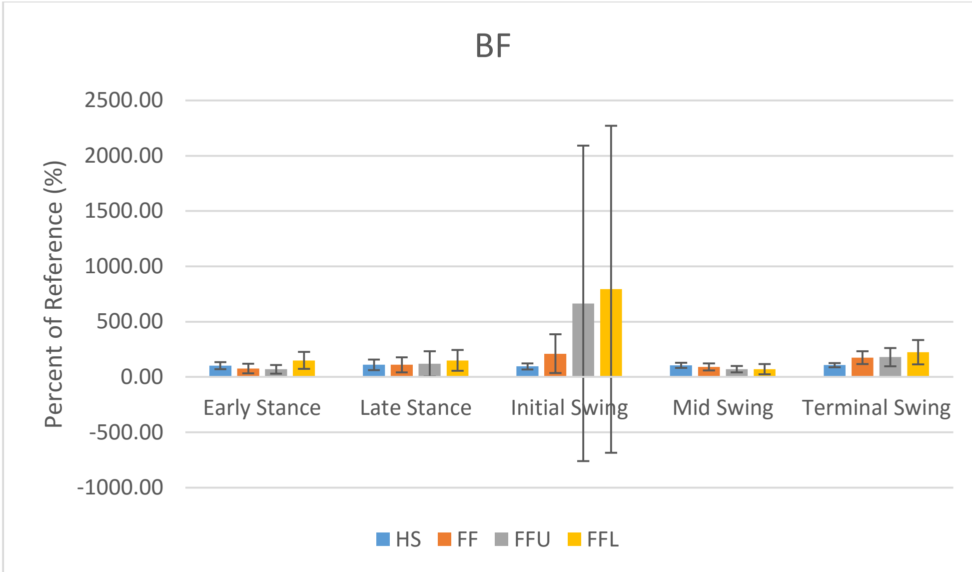


FIGURE 20: Normalized, Integrated BF Activity for Defined Running Phases

Table X

NORMALIZED, INTEGRATED BF ACTIVITY FOR DEFINED RUNNING PHASES

| Biceps Fem | Early Stance | Late Stance | Initial Swing | Mid Swing | Terminal Swing |
|---|---|---|---|---|---|
| HS | 101.80 | 110.34 | 95.37 | 105.13 | 106.94 |
| FF | 75.62 | 110.24 | 210.82 | 89.87 | 173.74 |
| FFU | 69.91 | 119.65 | 665.16 | 69.90 | 179.46 |
| FFL | 149.95 | 150.14 | 794.00 | 70.17 | 223.68 |

**9. <u>Overall Stride Cycle</u>**

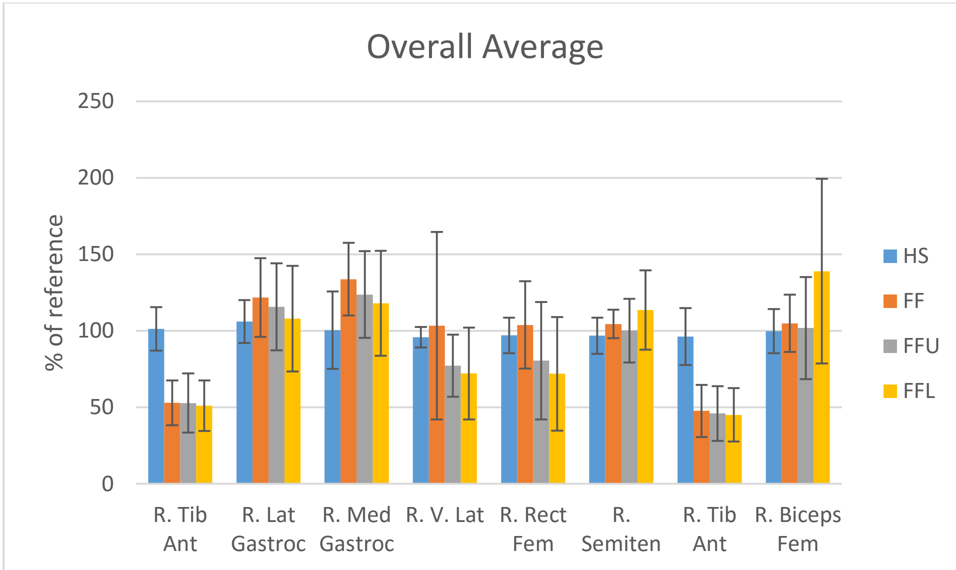


FIGURE 21: Normalized, Integrated Muscle Activity for Complete Stride Cycle

Table XI

NORMALIZED, INTEGRATED MUSCLE ACTIVITY FOR COMPLETE STRIDE CYCLE

| | TA | LG | MG | VL | RF | ST | TA (2) | BF |
|---|---|---|---|---|---|---|---|---|
| HS | 101.32 | 106.09 | 100.41 | 95.96 | 97.11 | 96.87 | 96.34 | 99.81 |
| FF | 52.99 | 121.74 | 133.82 | 103.46 | 103.83 | 104.54 | 47.70 | 104.94 |
| FFU | 52.81 | 115.68 | 123.78 | 77.20 | 80.50 | 100.20 | 46.02 | 101.83 |
| FFL | 51.09 | 107.97 | 118.01 | 72.16 | 71.96 | 113.67 | 45.12 | 139.01 |

# IV. DISCUSSION

The positive range values calculated by taking the minimum HS FSA minus the maximum FF FSA identify a difference in how the foot made contact with the ground. Additionally, the FF FSA and the HS FSA met the FSA degree requirements found by Ahn and colleagues (Ahn 2014). This additionally supports the findings of Lieberman and colleagues (2014) who found a difference in the kinematic angles produced by a forefoot strike and heel strike at foot strike. Therefore, each subject successfully made foot contact with the ground utilizing the stride assigned (FF or HS). Verification of the difference in FSA at ground contact can be found in TABLE I.

The positive range values calculated per subject by taking the minimum FFL BLA minus the maximum FFU BLA demonstrates each FFL stride was done with a greater BLA in comparison to the FFU stride. This positive range demonstrates that each body position assigned was successfully performed by each subject. Verification of the difference in body postures can be found in TABLE II.

The first aim involved determining differences in muscle activity patterns between HS and FF conditions. The second aim consisted of examining differences in muscle activity patterns between FFU and FFL conditions. Each aim had two hypotheses to address. The first hypothesis associated with each anticipated differences in muscle activity patterns with respect to separate phases within running cycles. The second hypothesis for each aim anticipated differences in the muscle activations over the entire cycle.

During terminal swing, a runner prepares for ground contact. While running with a FF pattern, individuals activated the gastrocnemius muscles to facilitate the initial forefoot ground contact. By activating the gastrocnemius muscles, the foot becomes plantarflexed as seen in FIGURE 11 for all three FF strides. Activation of the gastrocnemius during this phase for the FF patterns is additionally needed to prevent the heel from hitting the ground with excessive impact force following contact. These findings in gastrocnemius activity increases during this phase for the FF conditions is consistent with the findings of previous studies (Giandolini 2013) (Ahn 2014).

On the contrary, a runner with a HS pattern activates the TA to prevent “foot slap” immediately following ground contact. TA activation during this phase for the HS condition additionally causes dorsiflexion of the foot so that the runner can hit the ground heel first and fulfill the guidelines of the assigned stride. The dorsiflexion occurring during this phase is visually represented in FIGURE 11. The hip exhibits greater extension movement in the three FF conditions compared to the hip extension movement in the HS condition. Such increased hip extension movement is consistent with greater hamstring activity present in the FF conditions. While the hamstring muscles and gastrocnemius muscles have demonstrated increased hip extension movement and plantarflexion respectively in the FF conditions, due to their bi-articular nature, a tendency towards knee flexion consequently occurs as well. Therefore, greater RF and VL activity is present in the three FF conditions to create knee extension for ground contact despite the tendency for the bi-articular muscles to promote knee flexion.

In early stance, the gastrocnemius muscles continue to activate in the FF patterns to prevent the heel from slapping the ground. A lack of significance between FF conditions and HS condition in early stance is due to an increase in activation of the gastrocnemius muscles in the HS condition as well. The TA is activated in the HS condition to control the toe as it lowers to the ground after contact. Without TA activation here, the runners toe would immediately slap the ground at initial contact. The increase in activation of the MG, LG and TA for the HS condition leads to a possibility of co-contraction occurring. Co-contraction is utilized by the muscles to create joint stiffness which increases stability (Lee 2006). The need to increase stability could be occurring for a few different reasons. It is likely the runner is off balance when initially landing on the heel. Also, there could be a factor of overstriding occurring where the runner is landing on their heel further in front of their body center forcing the runner to have a need for increased stability.

During late stance, the runner creates a backwards force on the ground in order for the ground reaction force to push him in the forward direction. This backwards force can be made by plantarflexing the foot utilizing the calf muscles. An alternative way would be knee flexion or hip extension utilizing the hamstring muscles. Interestingly, the results identified a statistically significant increase in gastrocnemius activity for the HS, FF, and FFU condition during this phase. This finding would suggest an alternative mechanism for forward propulsion in the FFL condition. Examining the hamstring activity during late stance for all four conditions, no significant difference was found between conditions for the ST nor the BF hamstring muscles. This observation poses a theory which needs further testing to prove. When an individual makes initial ground contact, the foot exerts horizontal force on the ground in the anterior or posterior direction. If the foot exerts a force in the forward direction a deceleration phase occurs as the ground reaction force pushes the runner in the posterior direction. The runner then potentially has to compensate for the decelerating ground reaction force with increased subsequent acceleration to propel the runner forward. If the runner utilizing a FFL stride reduces the deceleration induced at foot contact, the acceleration needed would be reduced as well. If this is the case, it could explain the reduction of gastrocnemius activity in the FFL condition during late stance.

The gastrocnemius activity begins to decline as the runner goes into initial swing phase. The majority of the gastrocnemius activity calculated in this phase is due to the leftover result of the activity present from the previous phase. Additionally, as seen in FIGURE 9, the foot placement of the three FF conditions is more plantarflexed compared to the HS condition requiring the activation of the MG and LG. The hamstring muscles for the FF conditions must then activate to create more knee flexion for toe clearance while the foot is plantarflexed which is consistent with the increase in activation of the hamstring muscles for the FF conditions compared to the HS condition. To prevent knee flexion beyond the level needed for toe clearance, the VL activates to control the amount of knee flexion explaining the increase in VL activity for the three FF conditions compared to the HS condition. Once the runner hits peak hip extension, the hip goes into flexion utilizing the RF. By increasing the rate of hip flexion, the runner is able to increase forward propulsion.

During mid swing, gastrocnemius activation begins and then continues throughout terminal swing as previously discussed. Mid swing also involves toe clearance. The TA is activated to prevent toe drag as the foot passes closest to the ground during this portion of swing. All four conditions exhibit TA activity; however, the HS condition exhibits greater TA activity compared to the FF conditions. Elevated TA activation in the HS condition during midswing prepares the runner to make ground contact with the heel first.

The TA activity demonstrated lower variability in each phase per running pattern compared to the other muscles. The TA is the only muscle observed in this study that is able to achieve dorsiflexion explaining the consistency in TA activity between subjects. On the other hand, forward propulsion throughout a stride can be achieved in various mechanisms utilizing the MG, LG, VL, RF, ST and BF in different activation patterns.

Throughout this study, there were potential sources of error. One source of error could be a decrease in surface electrode/ subject conductivity due to sweat induced separation. One subject was disregarded and another subject came in for retesting. In these two cases, the issue of sweat induced electrode separation was magnified; however, it is possible that similar electrode separation could have been present but to a lesser degree in the remaining subjects.

A potential source of variance could stem from a lack of acclamation to each stride prior to testing using each stride. It is likely that each runner only ran naturally in one condition while the other three, possibly all four, conditions required concentration throughout the duration of the testing period. If the runners were able to perform each stride naturally, there may have been a difference in the results. There is evidence in the data suggesting this lack of adaption was present. The FF strike condition with no instruction of body alignment was performed with a BLA in between the FFL BLA and the FFU BLA in every case except one where the subject actually stood up taller without any instruction. However, in multiple instances the muscle activity output was reduced or higher in the FFL and FFU conditions and even lower or higher in the FF condition. For example, in the results section for the various BF stages, the average muscle activity was highest for the FFU condition followed by the FFL and then FF condition in the late stance, initial swing, and the terminal swing phases (Table 10). For midswing, the muscle activity was lowest in the FFL followed by the FFU and then the FF condition. In all four phases, it would have been expected that the FF muscle activity would have been in between the output of muscle activity for the FFL and FFU conditions.

This study identified varying muscle activity patterns when running with different strides. The biggest difference was seen when running with a heel strike condition verses a forefoot condition. Previous studies commented on force increases seen during a heel strike condition verses the forefoot condition. In fact Kulmala (2013) suggested a forefoot strike pattern “may decrease the risk of developing running- related knee injuries.” Kulmala’s theory was based on forces occurring at the knee. He calculated the forces with regards to joint angles, moment arms, ground reaction force, and center of mass data. This present study did not obtain extrinsic force data; however, the results displayed a greater amount of quadriceps, gastrocnemius, and hamstring muscle activation at various phases throughout the stride cycle in a forefoot condition compared to a heel strike condition. All of the muscles associated cross the knee joint suggesting there to be an increase in intrinsic force seen at the knee in a FFS verses a HS. Kulmala’s data is calculated while the foot is on the ground where as the muscle activity in the current study was monitored throughout the entire stride cycle. Given the differences in data collection time points between the two studies and the difference between intrinsic and extrinsic force calculations does not allow the current study to support or refute

Kulmala's study. However, these findings suggest the intrinsic forces at the knee should be considered while performing force studies at the knee.

Additionally, when the same runners ran with different strides, different muscle activity patterns were identified. Lohman (2011) identified a difference in stride when running in a traditional shoe compared to a minimalist shoe. Combining the results of the current study with Lohman's results, could explain McDougall's (2009) recount of the Tarahumara runners switching from brand new modern day shoes (elevated heel) to their usual flat sandals (minimalist) midrace due to pain. Once they started running in the new modern day shoes they're stride pattern most likely changed leading them to most likely use different muscle activity patterns. If this were the case, it could explain their discomfort and reason for switching back to their accustomed shoes midrace.

This study raises questions that could be addressed in future studies. Given the individuality of how a subject runs, altering the speed on the various conditions could magnify which methods and muscles the runner relies on for forward propulsion. Furthermore, identifying how the individual adapts could assist in modifying the individuals stride in a more efficient way. For example, if the subject relies mainly on calf activity, trying to modify the stride by standing more upright or adding a lean to identify where the individual will utilize different muscle groups as well could assist in a more efficient stride. Another future test needed to validate the data presented would include gait retraining for each of the four strides so the muscles become accustomed to the pattern in which they run naturally in each stride pattern.

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# APPENDIX I: PARTICIPANT SURVEY

Survey

Name:

Age:

Gender:

Height:

Weight:

What is your dominant foot?

How long have you been running?

Approximately how many miles do you run a week?

What is the brand and model of the shoe you run in?

What is your primary running surface (e.g. pavement, trail, etc.)?

What would you classify your strike pattern as: heel strike, forefoot strike, or unknown?

Have you ever had a running associated injury?

If so, what was it?

Do you do any sort of cross training?

If so, what kind?

# VITA

Nicole Knapp is originally from Shawnee, Kansas. She came to the University of Louisville as a member of the Swimming and Diving team as well as an undergraduate student in bioengineering. She received her Bachelors of Science in bioengineering from the University of Louisville in 2013. Her background in sports, passion for running, and education sparked the interest behind this study. She will receive her Masters of Engineering degree from the University of Louisville in 2016.