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FOOT AND ANKLE KINEMATIC AND LOWER EXTREMITY MUSCLE ACTIVITY DURING DESCENT FROM VARYING STEP HEIGHTS

by

Emily E. Gerstle

A Thesis Submitted in

Partial Fulfillment of the

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ABSTRACT

FOOT AND ANKLE KINEMATIC AND LOWER EXTREMITY MUSCLE ACTIVITY DURING DESCENT FROM VARYING STEP HEIGHTS

by

Emily E. Gerstle

The University of Wisconsin-Milwaukee, 2014
Under the Supervision of Professor Stephen C. Cobb

Ankle injuries are common during activities of daily living, particularly in negotiation of steps. Previous studies examining steps have generally focused on the ankle, knee, or hip and descent of multiple steps. Joint motion within the foot, utilizing a multi-segment foot model, during step descent has not been extensively studied. Although peroneal muscle activity differences have been identified between participants with healthy and unstable ankles during static activities, little is known about peroneal activity during activities of daily living. A better understanding of the foot kinematics and muscle activity in persons with uninjured ankles may help future studies elucidate the problems encountered by individuals with chronic ankle instability during step descent. Therefore, the purpose of this study was to identify the foot and ankle kinematics and lower extremity muscle activity of uninjured individuals during descent from varying step heights. Twenty-two participants (12 female/ 10 male, 25.68± 5.5 years) walked on a level walkway, stepped down a single step of varying heights (5 cm, 10 cm, 15 cm, 20 cm and 25 cm) and continued walking on level ground. Data acquisition included walking gait kinematics, utilizing a six-segment foot model, and peroneal muscle activity recorded with surface electromyography. Three-dimensional kinematics (initial contact angle, range of motion) across the five step heights from initial contact to the end of weight acceptance were analyzed via RM MANOVAs. Paired t-tests were used to compare muscle activity during the 200 ms prior to initial contact between each step height.

Results demonstrated a greater percentage of participants preferred to switch initial contact from a heel strike to a forefoot strike as step height increased. The calcaneonavicular complex had significant differences in initial contact angle in the transverse plane between the 5-cm step and steps of 20 and 25 cm. Range of motion differences were not significantly different across any of the step heights. Integrated electromyography differences were significant between the 5-cm step and the 15, 20 and 25-cm step heights; between the 10-cm step height and the two highest steps; as well as between the 15-cm and 20-cm steps. These results indicate stability of the medial midfoot and medial longitudinal arch may become more dependent upon dynamic stabilizers as step-down height increases and/or landing strategy transitions from heel to forefoot.

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Chapter 1: Introduction

An estimated five to ten million ankle injuries occur each year in the United States. In addition to being one of the most common injuries in the United States, the annual cost of caring for ankle injuries has been estimated to exceed two billion dollars (Birrer, Fani-Salek, Totten, Herman, & Politi, 1999). Due to the complexity of the ankle joint and the large forces to which the joint is exposed during walking and running, the injury frequency is not surprising. Although athletic activities account for a large portion of these injuries, there are still a significant percentage of individuals that are affected by ankle sprains through the course of their activities of daily living. Specifically, in a study of 100 hospital emergency room visits over a four year period it was found that over 25% of the ankle sprains requiring hospital care were the result of a fall from stairs (Waterman, Owens, Davey, Zacchilli, & Belmont, 2010).

Not only are stairs a common mechanism of acute ankle injury, it has been demonstrated those with chronic ankle instability also have difficulty negotiating steps. For example, the Cumberland Ankle Instability Tool, a questionnaire used to assess those with and without functional ankle instability, includes a question regarding how the ankle feels when going down stairs (Hiller, Refshauge, Bundy, Herbert, & Kilbreath, 2006). Furthermore, difficulty on stairs is also included as one of the characterizations of foot health within the Foot and Ankle Outcomes Questionnaire ("American Academy of Orthopedic Surgeons Lower Limb Outcomes Assessment: Foot and Ankle Module (AAOS-FAM)," 2005) and the revised Foot Function Index form (Budiman-Mak, Conrad, Mazza, & Stuck, 2013). The inclusion of descending steps on these foot and ankle assessments indicate that the daily task of negotiating steps is an important factor in determining foot and ankle health.

Several previous studies have examined ankle, knee and hip kinematic differences during step down activities (Bosse et al., 2012; Giguère & Marchand, 2005; Karamanidis & Arampatzis, 2011; McFadyen & Winter, 1988; Protopapadaki, Drechsler, Cramp, Coutts, & Scott, 2007). To date, however, few studies have examined the joint motion of the articulations distal to the ankle during a step down activity (Rao, Baumhauer, Tome, & Nawoczenski, 2009).

The ankle and foot consist of 16 articulations. Often the talocrural joint (the articulation between tibia, fibula and talus) is grouped with the subtalar joint (the articulation between the talus and calcaneus) and defined as the ankle complex. Beyond the ankle complex are the midfoot and forefoot, which are made up of the cuboid, navicular and cuneiforms and the metatarsals and phalanges, respectively. Traditionally biomechanical studies have modeled the foot as a single rigid segment; however, recent studies of the distal articulations during gait have demonstrated significant motion in the previously overlooked joints. Several invasive in-vivo (bone-pin) and in-vitro (cadaver) studies have reported significant movement in the mid- and forefoot during walking gait (Lundgren et al., 2008; Nester et al., 2007; Wolf et al., 2008). Recently, surface based multi-segment foot models have been developed that track motion of the distal foot articulations (Carson, Harrington, Thompson, O'Connor, & Theologis, 2001; Cobb et al., 2009; Jenkyn & Nicol, 2007; Kidder, Abuzzahab, Harris, & Johnson, 1996; Leardini, Benedetti, Catani, Simoncini, & Giannini, 1999; MacWilliams, Cowley, & Nicholson, 2003; Tome, Nawoczenski, Flemister, & Houck, 2006). Studies utilizing these models have helped shed light on previously unrecognized intricacies of distal foot movement during walking gait. Application of a multi-segment foot model to establish typical distal foot motion over a range of step heights in uninjured persons may help identify those heights, which are most likely to cause instability or injury to those with foot or ankle pathologies.

In addition to joint kinematics, lower extremity muscle activity during multiple step descent has also been studied. Andriacchi, Andersson, Fermier, Stern, and Galante (1980) found delayed gastrocnemius activity and decreased soleus and tibialis anterior activity during the last step down of multiple stair decent that transitioned into level walking. Furthermore, a fatigue study of a single step descent found average muscle activity decreased in the gastrocnemius and tibialis anterior muscles (Barbieri, Lee, Gobbi, Pijnappels, & Van Dieen, 2012). In level walking, functionally unstable ankles were found to have increased tibialis anterior and peroneal activation compared to controls (Hopkins, Coglianese, Glasgow, Reese, & Seeley, 2012). Those with chronic ankle instability have also been found to have delayed peroneal reaction time during static standing activities (Kavanagh, Bisset, & Tsao, 2012; Konradsen & Ravn, 1990). Delayed muscle response during or near initial ground contact may contribute to the feeling of instability or cause re-injury in those with chronic ankle instability. Furthermore, muscle activity across varying single step heights in uninjured or injured populations has not been extensively examined (Freedman & Kent, 1987; Freedman, Wannstedt, & Herman, 1976), particularly in the course of continuous walking gait. Therefore, a better understanding of the variations of lower extremity muscle activity patterns and distal foot kinematics during step negotiation of different heights during continuous ambulation in persons with uninjured ankles may help future studies elucidate the problems encountered in individuals with chronic ankle instability during step descent.

Purpose

The purpose of this study was to identify the foot and ankle kinematics and lower extremity muscle activity of uninjured individuals during descent from varying step heights. It was hypothesized that participants would make initial ground contact with the heel at low step heights and with the forefoot at higher step heights (Freedman & Kent, 1987). These changes in initial contact position were anticipated to be accomplished through significant kinematic differences in the rear-, mid- and forefoot joint complexes. Additionally we postulated that the rearfoot complex and medial forefoot ranges of motion from initial contact to foot flat would be increased when initial contact was made with the forefoot. Furthermore, we proposed that the medial midfoot and lateral forefoot range of motion would decrease when initial contact was made with the forefoot, to facilitate foot stability. With respect to lower extremity muscle activity and timing, it was anticipated that as step height increases and initial contact shifts from the heel to the forefoot, evertor (peroneal) activation would occur earlier and be more active in the step-down cycle to stabilize the medial longitudinal arch.

Delimitations

1. Data were collected on healthy participants without injury or perceived ankle instability walking at a self-selected pace, therefore any generalizations made are limited to this population.

2. The step heights ranged from 5 cm to 25 cm in 5 cm increments. This range was chosen to encompass common step heights within public building codes as well as one step height beyond current codes. Step heights in private buildings may exceed the public building codes.
3. Participants walked in a sandal during the task to permit use of the multi-segment foot model. Therefore, generalizations of this study are limited to individuals in a shod condition.

Assumptions

1. All questions were answered honestly during the initial phone screening.
2. Lower extremity segments are rigid bodies.
3. Bone motion can be represented by surface based markers.

Limitations

1. Surface markers, placed on the skin have some error due to soft tissue artifact. Through attachment of markers with adhesives, marker clusters of four markers, and rigid body reconstruction optimization procedures these artifacts were minimized.
2. Electromyographical data is variable between persons and influenced by electrode placement. Electrode placement was made following accepted standards. Muscle activity timing was based upon comparison to individual participants’ static calibration recorded just prior to step down trials.

Significance

The results of this study advance the understanding of the lower extremity muscle activity, timing, and distal foot kinematics during step negotiation in persons with uninjured ankles. Previous studies have not examined the effect of step height variation during walking down a single step. Additionally, only one study to date has made use of a multi-segment foot model to capture distal foot motion during step negotiation. Timing of muscle activation and magnitude of activity in regards to differences across step heights during ambulation has also been limited. A better understanding of healthy ankle and foot kinematics and muscle activity may help future studies elucidate the problems encountered in individuals with chronic ankle instability during step descent.

Chapter 2: Review of Literature

Introduction

As indicated by the Waterman et al. (2010) study of ankle injuries requiring hospital care, falls on steps are a common cause of injury among apparently healthy individuals. Additionally, the inclusion of questions related to step negotiation on several foot and ankle health measures indicate step negotiation is an important component of foot and ankle health in injured populations ("American Academy of Orthopedic Surgeons Lower Limb Outcomes Assessment: Foot and Ankle Module (AAOS-FAM)," 2005; Budiman-Mak et al., 2013; Hiller et al., 2006). Despite the importance of step negotiation, the majority of gait research to date has been focused on foot and ankle movements during walking on level ground. Little is known regarding foot function during stair descent. The studies that have been conducted have attempted to define normal parameters of the hip, knee and ankle mechanics and muscle patterns during continuous step negotiation. There has been limited research examining distal foot segment function during single step negotiation (Rao et al., 2009).

This review of literature begins with an overview of step studies, examining consistencies, differences and influences of step height on stepping mechanics. Multi-segment foot models are then reviewed, following the progression of the models from two to eight segments. Movement pattern consistencies for each model will also be discussed. The section concludes with a discussion of the actual bone motion found in in-vivo and in-vitro multi-segment foot model studies. Finally, studies examining electromyographical data of the lower limb are reviewed, exploring the differences between level walking, single step, and continuous stairs as well as varying conditions of vision and height. These sections provide the necessary background information to demonstrate the need for this study, to identify the foot and ankle kinematics and lower extremity muscle activity of uninjured individuals during descent from varying step heights.

Steps

Level walking compared to stair ascent/descent. Several studies have investigated mechanical differences between level walking and walking on stairs. Similar to level walking, step descent may also be broken into stance and swing phases, which can be further divided into sub-phases. The first stance sub-phase, weight acceptance, begins with initial contact and continues until the start of single limb support or foot flat position. Forward continuance, the second sub-phase, begins at single limb support and continues until double support. Finally, controlled lowering, the last sub-phase of stance, begins at double support and continues until toe-off. The swing sub-phases consist of leg pull through and foot placement (McFadyen & Winter, 1988; Zachazewski, Riley, & Krebs, 1993).

A three step descent study (10 healthy men, 25.5 cm step height) by Andriacchi et al. (1980), found ankle flexion moments were not significantly different from walking on level ground, however knee joint flexion moments were greater during step descent. Freedman and Kent (1987) compared level walking to a 20 cm step down and found increased ankle flexion during descent due to a plantarflexed position at initial contact. McFadyen and Winter (1988) examined ascending and descending kinematics, kinetics, and gait parameters during step negotiation of five steps (22 cm high) and compared results to the level walking data presented by Andriacchi et al. (1980). Most recently, in agreement with the previous studies, a study of 10 subjects on a three step staircase (20 cm high) by Beaulieu, Pelland, and Robertson (2008) found that compared to level walking, the stance to swing ratio during descent was increased. In addition, the study found an extra instance of negative ankle power at initial contact of stair descent that was not present during level walking. This additional eccentric control was attributed to the initial forefoot contact position during stair descent as opposed to the initial heel contact associated with level walking. A study of the reproducibility of kinematics and kinetics on stairs and level ground (10 healthy subjects, 4 steps 18 cm in height) found the transition step between level ground and ascending or descending steps was the least reproducible in comparison to continuous ascent, descent, or level ground walking (Yu, Kienbacher, Growney, Johnson, & An, 1997). These studies have demonstrated walking on level ground employs different mechanics than those needed during stair negotiation. Due to these differences and the incidence of injury on steps, a better understanding of step mechanics is warranted.

Stair ascent versus stair descent. Stair negotiation can be divided into stair ascent or descent. Although the obstacle does not change, the mechanics of negotiating stairs are dependent upon the direction of travel (McFadyen & Winter, 1988). In ascent the first phase, weight acceptance, begins with contact in the middle of the forefoot progressing to the pull-up phase in which the knee generates energy to progress the body upward. It is at this point in ascent when the greatest instability is present as the forward limb holds all of the body weight with the hip, knee and ankle all in flexion. As the cycle continues to forward continuance, the ankle generates the most energy. In contrast, during descent weight acceptance, foot contact is initiated on the lateral portion of the foot, with energy absorbed by the ankle. In late stance, the ankle generates energy, although not to the same extent as during ascent. A study of 11 healthy adults negotiating four 18 cm steps examined differences in the relationship of center of mass to center of pressure between ascending and descending steps (Zachazewski et al., 1993). When the center of mass location corresponds to the location of the center of pressure, the individual was said to be in a more stable position. When these positions diverged, the stance was considered unstable. The study by Zachazewski et al. (1993) found that during step descent, a larger difference between center of pressure and center of mass position was present compared to step ascent. Additionally, time in double support, the most stable portion of the gait cycle, was decreased while descending stairs. The authors hypothesized that both differences could contribute to increased injury or falls during step descent versus step ascent. From these studies, differences in step ascent and descent have been demonstrated during initial contact foot placement, timing of ankle energy generation and absorption, timing of double support and instability. Mechanics of the hip, knee and ankle are relatively well documented during stair negotiation; however, a more detailed look at mechanics of joints distal to the ankle has not yet been explored.

Single Step. Although there have been many studies examining multiple step descent, only a few have examined a single or transition step, which Yu et al. (1997) has demonstrated to vary from continuous gait. A study of 23 healthy young adults examined the timing and step adjustment strategies of walking and safely ascending or descending a curb (15 cm high) in an outdoor environment (Crosbie & Ko, 2000). Crosbie and Ko

(2000) found no consistent patterns between step length adjustment strategy and speed aside from adjustments occurring in a range of time between one half second and five seconds prior to negotiating the curb. A control and midfoot arthritis study of 50 subjects during level walking with a single step descent (19.7 cm high) utilized a five segment foot model to analyze the kinematics of the distal foot (Rao et al., 2009). The foot function of the groups was assessed via the revised foot function index (Budiman-Mak, Conrad, Stuck, & Matters, 2006), with the arthritis group scoring significantly higher (less function) than the control group. Both groups had increased dorsiflexion of the first metatarsophalangeal joint, calcaneal eversion, forefoot abduction, and ankle dorsiflexion range of motion during the step down compared to level walking. The arthritis group also showed significantly more calcaneus eversion range of motion than the control group during the step down. A more complete understanding of healthy stepping kinematics across varied step heights will be beneficial to elucidate differences within injured populations.

Influences of stepping mechanics. In descending steps initial contact has been noted to occur with the forefoot, as opposed to level walking where the heel contacts the ground first (McFadyen & Winter, 1988). However, the switch from initially contacting the ground with the heel to a forefoot or toe strike seems to be dependent upon step height, the velocity the steps are encountered, and the ability to see the step.

Step height. In a study of 11 healthy participants stepping down a single step of varying heights (0, 2.5, 5, 10, and 20 cm) Freedman and Kent (1987) demonstrated that the likelihood of an initial forefoot strike increased as step height increased. At a step height of two and a half centimeters there was only one trial in which a participant made initial contact with the forefoot. At a step height of 20 cm almost two-thirds of participants switched to exclusively a forefoot strike. The percentage increased to almost 75 percent when participants' vision was impaired. A study of 10 subjects descending a five step staircase with step heights of 13.8 cm, 17 cm or 22.5 cm found forefoot contact was used for all heights (Riener, Rabuffetti, & Frigo, 2002). Additionally a study by Spanjaard, Reeves, van Dieen, Baltzopoulos, and Maganaris (2008) examined 10 subjects descending a four step staircase of 17 cm, 8.5 cm, 25.5 cm or 29.75 cm and also found forefoot contact for all step heights. Although the Spanjaard et al. (2008) and Riener et al. (2002) studies found consistent forefoot strike despite differences in step heights, they examined step descent during continuous descent as opposed to a single step.

Gait velocity. A study examining a 10 cm high step focused on the influence of foot strike pattern on gait velocity and impact forces. van Dieen, Spanjaard, Konemann, Bron, and Pijnappels (2008) found that gait velocity and impact force of 10 participants with a forefoot initial contact position was significantly decreased compared to participants with a heel strike initial contact position. With initial forefoot contact, the ankle was in a plantarflexed position that enabled significantly more negative work or energy absorption compared to initial contact position with the heel. During a heel contact landing, the energy absorption role is transferred to more proximal and larger muscle groups. In a follow-up study van Dieen and Pijnappels (2009) examined eight younger (23$\pm$1 years) and 17 older (73$\pm$5 years) participants encountering a single step (5 cm, 10 cm or 15 cm high) while walking at three, four or five kilometers per hour. The authors reported an increased incidence of heel strike landings at increased speeds; however initial heel strike contact incidence decreased at greater step heights.

Step sight. In a study of an expected and earlier than expected step down (10 cm high), participants wore glasses blocking their lower visual field with a flag in their field of view used to indicate the location of the expected step. During an unexpected step down, the 10 healthy participants exhibited no double support phase; this caused greater initial impact ground reaction forces and was compensated by a quick follow-up step by the trail leg. The authors also reported that the ankle joint contributed more to kinetic energy absorption during an unexpected step down (van Dieen, Spanjaard, Konemann, Bron, & Pijnappels, 2007).

Summary. Stepping mechanics, as demonstrated by the literature reviewed, is dependent upon the height of the step, the velocity at which the step is encountered, and the ability to see or anticipate the step down. As most step studies have focused on continuous descent, the mechanics of the single step are yet to be fully explored. Through previous studies, general understanding of mechanics have been outlined and are in agreement. However, further study to better define the transition between heel and forefoot contact, as well as establishing healthy movement patterns of the distal foot will provide a baseline of comparison for future studies of injured populations.

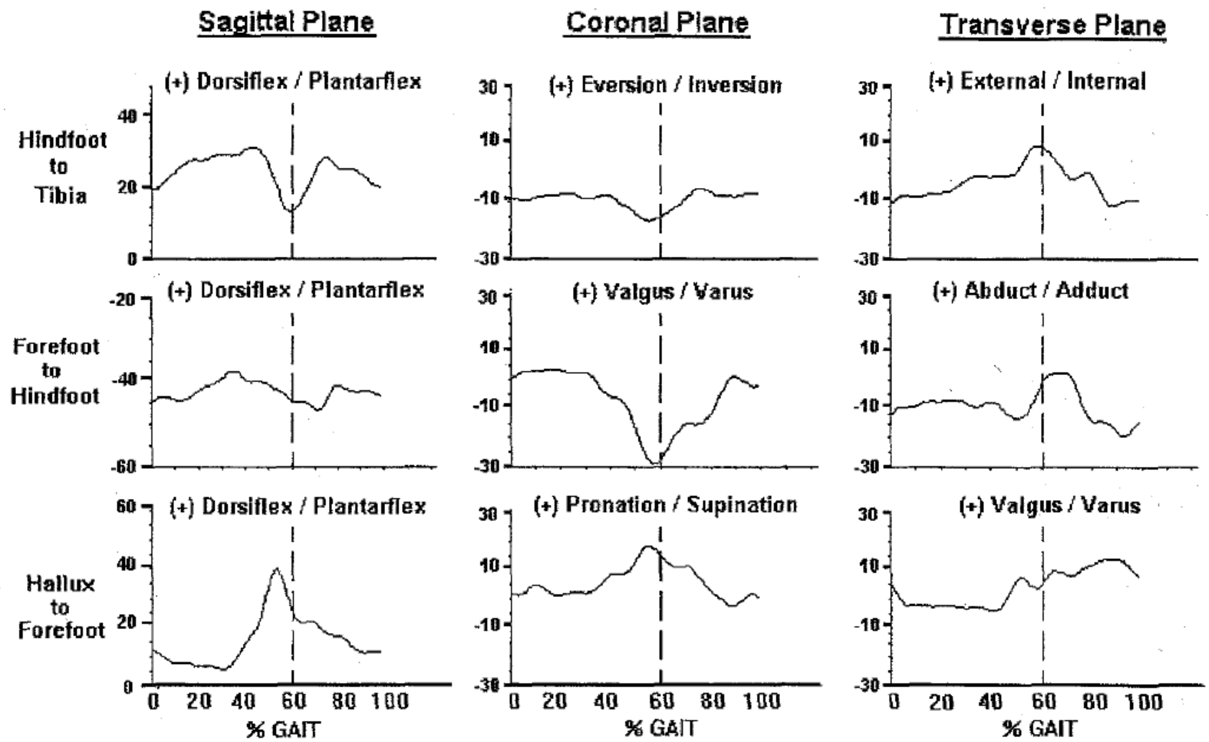
Multi-Segment Foot Models

Until recently foot motion has been tracked using single segment rigid body models. For studies primarily examining more proximal body segment movements, a single rigid body may be adequate. However, in attempting to better understand pathologies of the foot and ankle a more detailed look is warranted. As previously mentioned, in-vitro and invasive in-vivo studies have demonstrated significant motion within the foot. To study these motions many surface based multi-segment foot models have been developed.

Two and three segment foot models. The number of segments defined by the surface based multi-segment foot models have ranged from two (Pohl, Messenger, & Buckley, 2007) to eight (MacWilliams et al., 2003) foot segments. Pohl et al. (2007) used a two segment model of the rearfoot and forefoot to investigate barefoot walking and running in 12 healthy subjects. The repeatability of the model kinematics were high, and most motions were not significantly changed across varying running speeds. There were however, differences in peak motion timing and excursion between walking and running. A two segment foot model study of the hindfoot and forefoot on 24 healthy adults (16 female, 8 male) examined the kinematic differences of walking barefoot on various slopes. Their findings showed significant differences in the sagittal plane for both foot mechanics (more dorsiflexed position early in stance, more plantarflexed position at toe-off and during early swing) and timing of foot motion (peaks were reached earlier) as slope incline increased (Tulchin, Orendurff, & Karol, 2010).

Kitaoka et al. (2006) were the first investigators to utilize a multi-segment foot model (rearfoot and forefoot) in healthy participants (n = 20) walking in a shod condition. Markers were located on the foot segments by cutting holes in the shoes to accommodate the markers. The ankle-hindfoot complex (neutral at initial contact, plantarflexion through 25% of stance, dorsiflexion from 25-90%, plantarflexion through toe-off; total range of motion 18.3 ± 4.5°) and metatarsal-calcaneal complex (neutral at heel strike, dorsiflexion throughout stance, plantarflexion at toe-off; total range of motion 12.0± 3.0°) findings were similar in pattern to previous multi-segment foot models in the sagittal plane with differences attributed to coordinate systems and/or the footwear condition.

One of the first surface based multi-segment foot models was a three segment model of the foot (calcaneus, talus & navicular; cuneiforms, cuboid & metatarsals; proximal phalanx of hallux) and leg developed by Kidder et al. (1996). The study tracked a single male participant walking barefoot and found significant hindfoot, forefoot and hallux motion in all three planes (Figure 1). Carson et al. (2001) conducted a repeatability study of two individuals walking barefoot, utilizing a three segment model (hindfoot, forefoot and hallux). The study found good between trial repeatability and expected motion patterns of dorsiflexion of the hind- and forefoot at midstance and dorsiflexion of the hallux at heel off in the sagittal plane. During terminal stance, the study reported forefoot adduction and a neutral hallux in the transverse plane. Frontal plane motion in the forefoot also demonstrated good between trial reliability.



Adapted from Kidder, 1996

Figure 1. Kinematic motion during gait.

Four segment models. Leardini et al. (1999) utilized a four segment foot model (calcaneus, midfoot, first metatarsal and proximal phalanx of hallux) to investigate barefoot walking kinematics in nine healthy subjects (5 male, 4 female; 25-45 years). The study reported high repeatability within subjects’ however between subject, repeatability was only high for the tibia-calcaneus, metatarsal-phalanx, and midfoot-metatarsal in the sagittal plane and the metatarsal-phalanx in the transverse plane. Overall the tibial-calcaneus and metatarsal-phalanx motion maintained the same patterns of movement reported by Kidder et al. (1996) and Tulchin et al. (2010). This study was the first to report the motion in the calcaneus-midfoot, with small ranges of motion in all planes, and midfoot-metatarsal joints (slight dorsiflexion and pronation at heel strike, supination prior to terminal stance, and pronation at toe-off).

Jenkyn and Nicol (2007) studied 12 asymptomatic subjects walking barefoot using a four segment foot model which included calcaneus, tarsals, medial and lateral forefoot segments. They found the forefoot motion was in agreement with Kidder et al. (1996). Regarding the forefoot, this study was unique in that it included medial longitudinal arch height to length ratio. The ratio dropped after initial contact until late stance and then rose through toe-off.

In addition to investigating kinematics of healthy participants, multi-segment foot models have also been utilized to explore differences between healthy and injured feet. A barefoot walking study comparing 10 healthy subjects to 14 participants with posterior tibialis tendon dysfunction used a multi-segment foot model that defined four foot segments (rearfoot, medial forefoot, lateral forefoot and hallux). Although four segments were defined, only the rearfoot, midfoot and the forefoot as a single segment were discussed. Results of the study indicated that the injured group had greater rearfoot eversion, forefoot abduction and a significantly lower arch throughout stance (Tome et al., 2006). A study comparing walking kinematics during a shod (sandal) condition in participants with typical (11 subjects) and low-mobile (11 subjects) using a four segment foot model (rearfoot complex, calcaneonavicular complex, medial forefoot and first metatarsophalangeal complex) found that participants with low-mobile feet had significant differences in excursion of the calcaneonavicular complex as well as the rearfoot complex (Cobb et al., 2009). The calcaneonavicular complex had decreased abduction during midstance while the rearfoot complex had increased inversion in the pre-swing phase. The study also reported high reliability in all planes for all functional articulations. In a two week orthotic intervention study in 16 low-mobile foot posture subjects utilizing the Cobb et al. (2009) model, Cobb, Tis, Johnson, Wang, and Geil (2011) found greater rearfoot complex dorsiflexion displacement during the orthotic condition. The displacements during the orthotic intervention were similar to the typical group reported in the Cobb et al. (2009) study.

A comparison of foot kinematics during level walking and stepping down a single 19.7 cm step between 20 healthy controls and 30 participants with midfoot arthritis made use of a four segment foot model (Rao et al., 2009). Differences in peak angles and ranges of motion between level walking and stepping down were found at the first metatarsophalangeal joint, the rearfoot, and the forefoot in both groups. Step descent had increased peak dorsiflexion of the metatarsophalangeal and rearfoot joints and increased calcaneal eversion, as well as forefoot abduction. Greater ranges of motion were found in the sagittal plane of the metatarsophalangeal and rearfoot joints and abduction of the forefoot. The study also reported a significant Group x Activity interaction effect for ranges of motion of both calcaneal eversion and the first metatarsal plantarflexion. In walking, both groups demonstrated similar calcaneus eversion excursion. During the step task, however, the arthritis group exhibited significantly more excursion. The first metatarsal joint had similar plantarflexion range of motion in the step task of both groups, but the arthritis group had significantly decreased range of motion during walking (Rao et al., 2009).

Eight segment models. An eight segment model comprised of: the hallux, medial phalanges, lateral phalanges, medial forefoot, lateral forefoot, calcaneus, cuboid, and medial midfoot (talus, navicular & cuneiform) was used to study the barefoot walking kinematics of 18 adolescents (MacWilliams et al., 2003). The ankle complex, calcaneocuboid, and medial phalanges-hallux all had motions of similar shape and magnitude in all three planes as those reported by Leardini et al. (1999). The greatest differences between the models was the medial tarsal-metatarsal flexion of the MacWilliams model and the midfoot-metatarsal flexion of Leardini et al. (1999) model. The difference may be partially attributed to local coordinate system definitions. Furthermore, in the MacWilliams model the joints articulated by the talus, navicular and cuneiforms were all grouped into a single rigid segment. As a result, movements could be attributed to either the medial tarsometatarsal segment or the ankle. The study also reported high variability between subjects at the first metatarsophalangeal joint. Results from another eight segment foot model (talus, cuboid, calcaneus, medial and lateral forefoot, hallux, medial toes and lateral toes) developed by Hwang, Choi, and Kim (2004) was utilized to investigate barefoot walking kinematics of five healthy males. The primary difference between the Hwang et al. (2004) and MacWilliams et al. (2003) studies was in the definition of the medial midfoot. The Hwang et al. (2004) study only tracked the navicular rather than the talus, navicular and cuneiforms. Comparing the two studies, there are differences in all three planes between the medial and lateral midfoot segments; most noticeably in the frontal plane with the lateral midfoot inverted close to 15 degrees more than the MacWilliams et al. (2003) study. Additionally, the three joints of the hindfoot also showed differences. Although the frontal plane calcaneocuboid movement patterns were similar between the two studies, the MacWilliams et al. (2003) study showed greater inversion throughout. In the transverse plane both talocrural and subtalar joints were different with the Hwang et al. (2004) study showing greater internal rotation of the talocrural joint and a more externally rotated subtalar joint. In addition to the differences in the functional articulations, the differences between the studies may also have been attributed to the differences in subject population. The Hwang et al. (2004) study examined five adult males while the MacWilliams et al. (2003) study tested 18 adolescents (7-16 years), eight males and 10 females.

Model Overview. The wide variety of foot segment models has demonstrated unique motions for various foot segments. Based on the various models, there appears to be general agreement that the foot should be segmented into three components from proximal to distal. However, the medial and lateral segmentation does not appear to be widely agreed upon. Although the two segment models capture more motion than the previous single segment models, Pohl et al. (2007) recognized the difficulty of their model in identifying movements taking place at the midtarsal or tarsometatarsal joints. Additionally the Tulchin et al. (2010) two segment sloped walking study suggested the need for future studies to include among other things, electromyography and adaptations to stairs for better assessment of motion in activities of daily living. Although the Jenkyn and Nicol (2007) study collected the forefoot as medial and lateral segments, all analyses grouped the forefoot segments together. This was despite the authors stated importance of separating the medial and lateral forefoot. A follow-up study making use of the same four segment foot model found the midfoot segment motion of their model matched that of other studies modeling the foot as a single rigid segment, indicating single segment studies may have captured midfoot movement, but not the added details of rearfoot or forefoot movement. (Jenkyn, Anas, & Nichol, 2009).

Bone motion. In an attempt to identify the most clinically relevant segments for surface based multi-segment foot models, studies using in-vitro and invasive in-vivo techniques have attempted to identify functional units of the foot. A 13 foot segment cadaver study examining the kinematics of the tibia, talus, calcaneus, navicular, cuboid, three cuneiforms, five metatarsals and the proximal phalanx of the hallux utilized arrays of reflective markers on bone pins that were inserted into each segment of interest (Nester et al., 2007). The leg was placed in a walking simulator with motorized loads applied to the extrinsic muscles of the foot. Although movements were adjusted until results met previous studies' qualitative overall patterns of kinematic findings, there were limitations for several areas of movement. The authors determined results "describe what the foot was capable of rather than how it performed in-vivo." With that limitation in mind, movements of the metatarsals did not support grouping all five metatarsals as a single rigid segment. In addition, while the navicular and cuboid moved independently, the motion direction and timing were always matched, which the authors determined could provide rationale for grouping the midfoot together as a single functional unit. Additionally motion between the cuneiforms and navicular was found to be larger than either the navicular and talus or the cuboid and calcaneus supporting the authors' claim that many previous multi-segment models may still not have identified the articulations with the greatest motion (Nester et al., 2007).

In a follow-up study, the same group conducted a study of six healthy males utilizing bone pins in-vivo to examine the bones that could be grouped as functional units or rigid segments within the foot. It was determined that the medial bones from the navicular distal to the first metatarsal could not be considered a functional unit, however the medial cuneiform and first metatarsal could be considered rigidly linked. Additionally in agreement with Nester et al. (2007), it was suggested that the navicular and cuboid could be modeled as a functional unit. The authors' general recommendations included having markers denoting the following segments: calcaneus, navicular-cuboid, medial cuneiform-first metatarsal, and the fifth metatarsal (Wolf et al., 2008). Another related study utilizing the same data collected from Wolf et al. (2008), described the kinematics of joint motion during walking. The findings agree the cuboid and navicular could be considered a functional unit but could not be considered a rigid unit. The authors also cited greater than expected navicular-medial cuneiform motion as rationale for not defining the segments from the navicular distal to the first metatarsal as single rigid segment. Additionally, it was confirmed the medial and longitudinal arches individually have unique motion (Lundgren et al., 2008).

Electromyography (EMG)

Lower extremity muscle activity patterns during step down have been studied, although the emphasis has typically been on muscle activity amplitude (Barbieri et al., 2012; van der Linden, Marigold, Gabreels, & Duysens, 2007), rather than muscle activation initiation. EMG differences have been found across studies with varying step number, visual circumstances, and healthy and injured subjects. With varying protocols, finding consistent patterns and timing across studies is difficult, thus a more complete understanding of healthy muscle activity timing in varying conditions will be beneficial for future studies.

Level walking and single step descent. In comparing muscle activity during level walking and stair descent the lower leg muscles monitored most often have included the tibialis anterior, medial gastrocnemius and soleus. A study of 12 young adults by van der Linden et al. (2007) examined surface EMG of the tibialis anterior and gastrocnemius when encountering an expected or unexpected single step down (5 cm high). The unexpected step protocol used glasses that blocked the lower visual field to allow study of the unexpected step down. Increased muscle activity of both muscles was found during the unexpected step down, with muscle activity onset determined via visual inspection of the EMG graphs. The foot was also in a more plantarflexed position in the unexpected step down condition (van der Linden et al., 2007). Another study of 10 subjects walking, encountering a step (10 cm high) and continuing on level ground before and after completing a fatigue protocol found no significant changes in the amplitude of muscle activity in the tibialis anterior or lateral gastrocnemius (Barbieri et al., 2012).

Multiple step descent. Studies examining stair negotiation (multiple step descent) have included assessment of the tibialis anterior, gastrocnemius, and the soleus. Mann and Inman (1964) assessed the activity of six intrinsic foot muscles (abductor hallucis, abductor digiti minimi, dorsal interosseous, extensor digitorum brevis, flexor digitorum brevis, and flexor hallucis brevis) along with the tibialis anterior and gastrocnemius via intramuscular electrodes in eight subjects (5 normal, 3 flat-footed) walking on level and sloped surfaces and during ascent and descent of six steps (15.24 cm high). During level walking it was found that the intrinsic muscles in the subjects with normal feet were active only during mid- to late stance phase with individual muscles becoming active between 20 to 40 percent of the gait cycle. In subjects with flat-feet, intrinsic muscle activation varied in that most muscles were activated earlier, however, there were no differences in timing of muscle activity of the lower leg. During step descent, no intrinsic muscles or lower leg timing differences were found between normal and flat foot subjects. The gastrocnemius activated at 65 percent of the gait cycle and remained through 55 percent of the following step while the tibialis anterior did not show consistent activity.

A study of six males descending four steps (16.5 cm high) examined surface EMG of the tibialis anterior, soleus, and thigh muscles to examine if consistent muscle patterns were present during stair negotiation. The tibialis anterior was consistently active during the swing phase as well as from the end of swing to the start of support phase. The soleus, however, was only active during the support phase (Joseph & Watson, 1967). In a more recent study, Andriacchi et al. (1980) examined 10 healthy participants descending three steps (21 cm high). Their results demonstrated a difference in muscle activity between continual step descent and the last step onto level ground. The soleus, tibialis anterior and gastrocnemius were active for a longer period when stepping onto level ground. The soleus activated at the same time while the tibialis anterior activated earlier, and the gastrocnemius was delayed when stepping onto a level surface.

Effects of step height and vision conditions. As seen in the Andriacchi et al. (1980) study, continuous verses single step descent impacts muscle activation. Additionally knowledge of the step and when or if it will be encountered influences the timing of muscle activation. With these factors influencing muscle timing, it would seem that step height might also be in important factor. Freedman et al. (1976) examined EMG patterns of 12 healthy adults during varying visual conditions during a single step descent (8 cm, 22 cm, 33 cm, 43 cm) standing static before and after the step. The vision conditions consisted of normal, blindfolded with known step height, blindfolded with unknown step height, normal with Achilles tendon vibration, and blindfolded unknown step height with Achilles tendon vibration. Their findings indicated the muscle activity patterns of the triceps surae, tibialis anterior, vastus medialis and medial hamstrings were consistent across all step heights under two conditions; normal vision and blindfolded with knowledge of step height. Unknown step height while blindfolded or with Achilles vibration decreased amplitude and delayed timing of EMG activity during either visual condition.

A more recent study of 14 healthy subjects descending three steps of different height conditions (20 cm, 30 cm or 40 cm) under varying visual conditions (normal vision, blindfolded, and blocked close vision with manufactured visual cues) demonstrated EMG activity changes. The blocked close vision condition caused the EMG activity to be greatest and start the earliest. The blindfolded trials had the least anticipatory EMG activity. The overall conclusion drawn was that optimal step performance requires full vision and any changes to peripheral sensory information will change muscle activity (Craik, Cozzens, & Freedman, 1982). Comparison with the Freedman et al. (1976) study is difficult as only the overall pattern of all muscle activity was stated to be consistent, specific comparisons of similar conditions across step heights were not mentioned. Craik et al. (1982) demonstrated amplitude of EMG signal increased with step height but timing was not examined.

Summary. The timing of muscle activity seems to be dependent upon vision, knowledge of the step location and height, and whether a single step or multiple steps are encountered. Additionally there are conflicting claims between the Mann and Inman (1964) and Joseph and Watson (1967) studies regarding the tibialis anterior activity. The footwear worn by the participants may have contributed to the inconsistent results. Mann and Inman (1964) specifically mention rubber diving socks were worn and that initial contact occurred with the forefoot. In the Joseph and Watson (1967) study on-the-other hand, shoes were worn by the participants. The study did not mention however, the contact position of the foot. If contact was made with the heel, the different foot positions at contact may have caused the inconsistency in tibialis anterior activity. Further study examining muscle activation timing during continual movement descending a single step of varying heights while shod will help clarify these inconsistencies.

Chapter 3: Methods

Participants

Twenty-two apparently healthy participants (12 female, 10 male; 25.7±5.6 years) were recruited for the study from the UWM and surrounding community via UWM classroom announcements and posted flyers.

Inclusion/Exclusion Criteria

To participate, subjects must have been between the ages of 18 to 40 years, not wear bifocals, have weight bearing dorsiflexion range of motion of greater than 25 degrees, have had no history of surgery to the lower extremity and no lower extremity injury within the past six months. Ankle health was assessed via the Cumberland Ankle Instability Tool (CAIT) (Hiller, Kilbreath, & Refshauge, 2011; Hiller et al., 2006). To qualify, individuals had to score 28 or greater on the CAIT (29.5±0.67). A score of 28 to 30 on the CAIT indicates the individual is unlikely to have perceived instability, while a score of 27 or below indicates the presence of perceived instability. Prior to participation, all subjects were informed of the study procedures and asked to sign an informed consent form approved by the University's Institutional Review Board.

A previous study of continuous walking while encountering a single 10 cm step down found a large effect size when comparing ankle range of motion between fatigued and non-fatigued subjects (Barbieri et al., 2012). Based on this study to reach a power of 0.8 with alpha = 0.05 and to reach a moderate effect size (Effect size= 0.25) in variables between step heights a minimum of 21 subjects were needed.

Study Protocol

Initial Phone Screening. To confirm eligibility, participants completed a 15-minute phone screening assessment that included questions regarding the inclusion criteria. The CAIT questionnaire was also delivered during the phone screen (Appendix A).

Laboratory Visit.

Gait Analysis. Participants walked at a self-selected pace, along a 5 m runway, stepped down a single step and continued walking another 3 m. At each step height, three practice trials were given to familiarize them with the height and establish the participants' self-selected speed. During the subsequent step trials, the participant's approach speed was within a 10 percent range of their practice trial speed to ensure consistency across the trials. Participants completed trials at 5 cm, 10 cm, 15 cm, 20 cm, and 25 cm step heights, which encompass standardized building code stair heights and a step that is 5 cm above code (StairwayManufacturers'Association, 2006). Step heights on public property must be within code; however, private property where 47.9 percent of ankle injuries occur (Waterman et al., 2010) does not fall under these regulations and particularly in older homes transition steps may exceed the maximum code height. Ten trials were collected with the participants' preferred step down contact (heel or forefoot) at each step height. The increasing or decreasing progression of the step heights was counterbalanced by participant to control for any learning or fatigue effects.

Multi-Segment Foot Model. Foot kinematics during step descent were assessed by clusters of four retro-reflective markers placed on the participants' foot and leg on the following segments: calcaneus, navicular, cuboid, hallux, medial rays (first and second metatarsals), lateral rays (fourth and fifth metatarsals) (Figures 2 & 3). The six segment model that was used in the study has been shown by Cobb, Joshi, Bauer, and Klinkner (2011) to be very reliable during walking gait. The functional articulations that were defined from the six foot segments are identified in Table 1. Prior to the step descent trials, a static calibration procedure was performed. The calibration procedure involves capturing the position of retro-reflective markers located on several additional anatomical landmarks (Table 2). The additional markers were then removed prior to completion of the step descent trials. Additionally anatomical landmarks were also recorded during static calibration with a pointer. The use of a pointer was introduced by Leardini et al. (1999), to avoid difficult marker placement such as landmarks at the edge of the foot along the soft tissue.

Three dimensional marker position data was collected with a 10 camera Eagle system (Motion Analysis Inc., Santa Rosa, CA) sampling at 200 Hz. Initial contact and toe-off events of the stair descent were recorded with a force plate (AMTI, Inc., Watertown, MA) sampling at 1000 Hz.

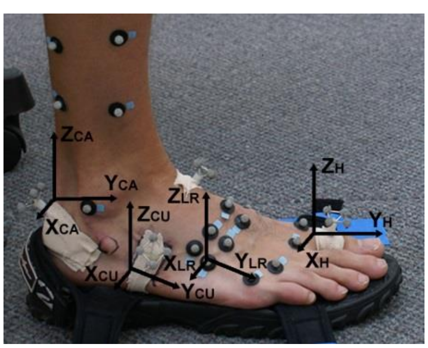


Figure 2. Lateral markers.

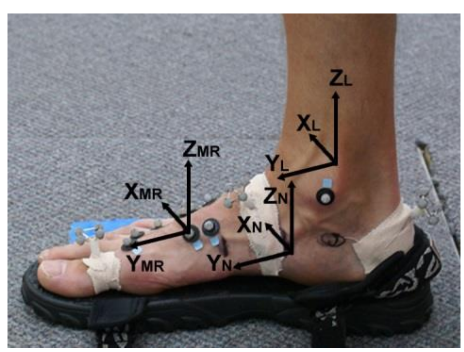


Figure 3. Medial markers.

Table 1. Functional Articulations.

Functional Articulation	Proximal Segment	Distal Segment
Rearfoot complex	Leg	Calcaneus
Calcaneonavicular complex	Calcaneus	Navicular
Calcaneocuboid	Calcaneus	Cuboid
Medial forefoot	Navicular	Medial rays
Lateral forefoot	Cuboid	Lateral rays
1st Metatarsophalangeal complex	Medial rays	Hallux

Table 2.
Anatomical Landmarks & Marker Locations

Segment	
Leg	Tibial tuberosity#
	Lateral malleolus#
	Medial malleolus#
	Marker wand cluster*
Calcaneus	Marker wand cluster* Sustentaculum tali† Peroneal tubercle† Posterior calcaneus†
Navicular	Marker wand cluster* Proximal dorsal† Proximal plantar† Distal plantar†
Cuboid	Marker wand cluster* Dorsal proximal† Plantar proximal† Plantar distal†
Medial rays	Head of 1st metatarsal#
	Head of 2nd metatarsal#
	Base of 1st metatarsal*
	Distal 1st metatarsal*
	Proximal 2nd metatarsal*
	Distal 2nd metatarsal*
Lateral rays	Head of 5th metatarsal#
	Head of 4th metatarsal#
	Base of 5th metatarsal*
	Distal 5th metatarsal*
	Proximal 4th metatarsal*
	Distal 4th metatarsal*
Hallux	Marker wand cluster* Base first proximal phalanx† Head first distal phalanx† Medial surface first distal phalanx†

*Technical markers
#Anatomical calibration markers identified with 6.4 mm marker
†Anatomical calibration marker identified with Davis Pointer

Electromyography. A Noraxon Telemyo Electromyographical (EMG) system (Scottsdale, AZ, USA) and Noraxon Ag/AgCl surface electrodes (Scottsdale, AZ, USA) were used to assess peroneal muscle activity during the step descent trials. The surface EMG electrodes were placed on the peroneus longus muscle. Skin preparation of shaving and cleansing with alcohol and electrode placement followed standardized techniques of Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM)(Hermens HJ, 1999) and the modifications proposed by Sacco, Gomes, Otuzi, Pripas, and Onodera (2009). The electrodes were placed at 25 percent of the muscle body length along the line of the fibula head to the lateral malleolus, the inter-electrode distance was at 2 cm, and the ground was located on the seventh cervical vertebra. Prior to performing the step trials, a static standing EMG calibration was taken to establish baseline muscle activity. During the trials muscle activity was monitored beginning 200 ms prior to initial ground contact through the weight bearing phase of the step (Delahunt, Monaghan, & Caulfield, 2006b; Gutierrez et al., 2012).

Data Processing.

Kinematic Data. Tracking of the kinematic data was performed using Cortex software (Motion Analysis Inc., Santa Rosa, CA). The tracked data were then exported to MATLAB (MathWorks, Natick, MA) to be filtered (fourth order zero-lag Butterworth filter; cutoff frequency of 5 Hz) and to perform rigid body transformation procedures using the calibrated anatomical systems technique (Cappozzo, 1984). The program also defined local joint coordinate systems within each foot segment and calculated joint angles of the six functional articulations using the joint coordinate system technique (Cole, Nigg, Ronsky, & Yeadon, 1993; Grood & Suntay, 1983). For analysis, the variables of interest from five successful trials of similar landing pattern for each subject at each step height were averaged.

Initial contact was considered when the vertical ground reaction force exceeded 10 N and single limb support was assumed to coincide with the first vertical ground reaction force peak which is also when the foot is flat (Zachazewski et al., 1993) (Figure 4). Three-dimensional initial contact angles and the ranges of motion for all of the functional articulations between initial foot contact and the foot flat position were determined.

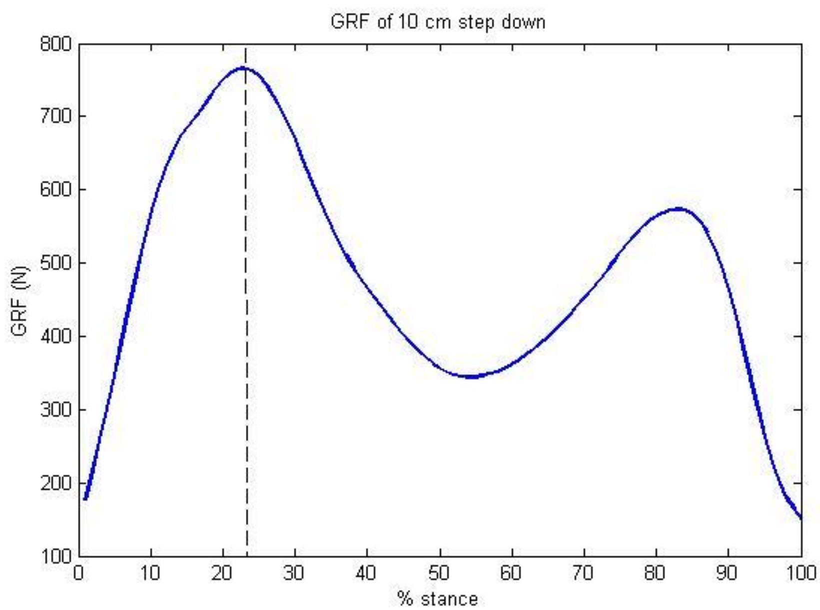


Figure 4. Dashed line indicates end of weight acceptance phase.

Electromyograpical Data. EMG data were filtered with a Butterworth high pass filter (20 Hz) and low pass filter (490 Hz), fully rectified and smoothed using custom written software (MATLAB) prior to determining amount of muscle activity and timing onset. To quantify muscle activity, the area 200 ms prior to contact was integrated for each step height. Muscle activity onset was defined as EMG signal that exceeds three standard deviations beyond the mean amplitude of the static resting calibration and checked visually. Muscle activation timing was expressed as time (ms) prior to contact.

Data Analysis. Muscle activation was analyzed via dependent t-tests with a Bonferroni correction utilizing alpha = 0.005 (0.05/10 tests).

Three dimensional foot position at initial contact were analyzed with RM MANOVAs with three dependent variables (sagittal, frontal, transverse plane initial contact position) and five within subject factors (step height). Univariate repeated measures ANOVAs were performed to investigate significant omnibus RM MANOVA F-ratios and dependent t-tests with a Bonferroni adjustment were used to investigate significant RM ANOVA omnibus F-ratios.

Finally, range of motion data from initial ground contact until foot flat were analyzed using RM MANOVAs with three dependent variables (sagittal, frontal, transverse plane range of motion) and five within subject factors (step height) for each of the five joint complexes. Univariate repeated measures ANOVAs were performed to investigate significant omnibus RM MANOVA F-ratios and dependent t-tests with a Bonferroni adjustment used to investigate significant RM ANOVA omnibus F-ratios. All kinematic tests were performed with a significance level of $\alpha = 0.05$, all statistics were run using IBM SPSS statistics package (v. 22.0).

Chapter 4: Results

Step Landing Strategy - Visual Assessment

The distribution of preferred landing strategy at each step height changed as step height increased. For each step height participants repeated step trials until ten of the same landing strategy were recorded, the distribution of preferred strategy of the first ten trials regardless of landing strategy at each step height was then analyzed (Figure 5). Of the 220 step trials for the 5-cm step height, 218 (99%) trials were heel landings. The 10-cm step had a heel strategy preference 82.7% of the time. At the 15-cm step, heel contact was the preferred landing strategy in 57.3% of the trials. For the 20-cm and 25-cm steps, heel contact was the preferred strategy during 45.5% and 30% of the trials respectively.

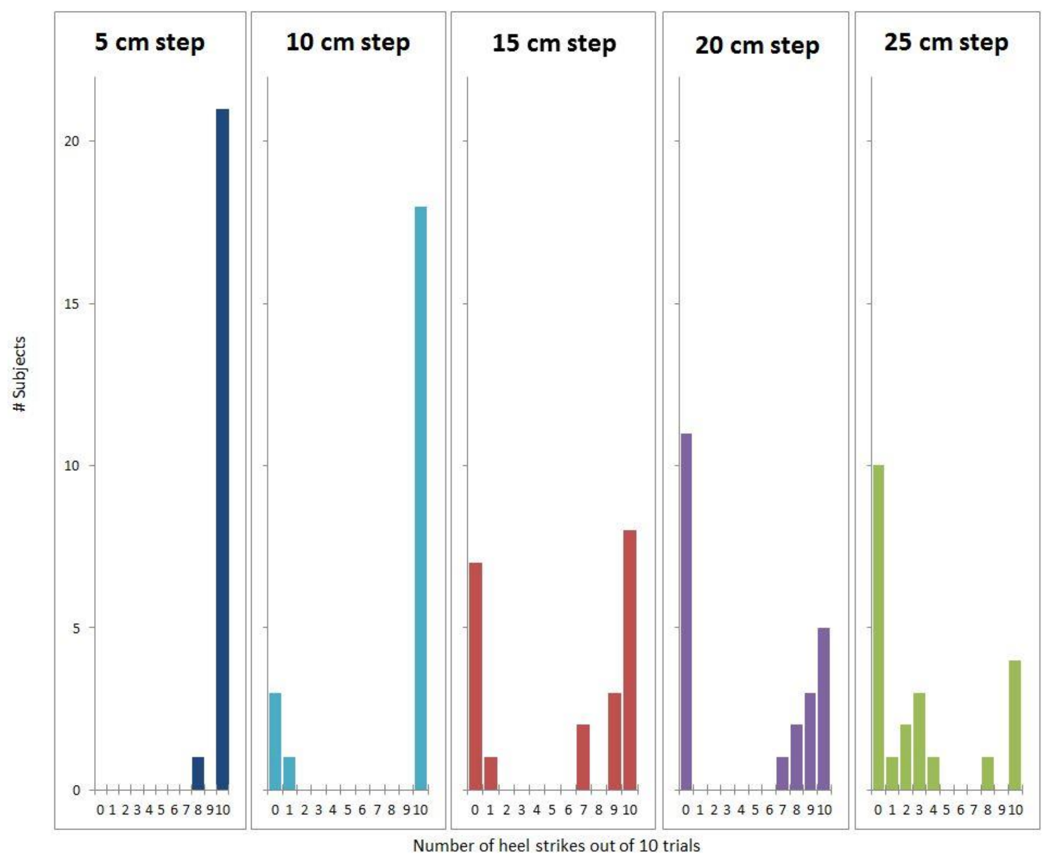


Figure 5. Preferred heel strike landing distribution

Kinematics

Due to difficulty with marker tracking of five participants, kinematic data from 17 participants' (8 male, 9 female) were used for analysis. Three participants had markers that did not clearly reflect during the gait trials and two participants had markers that could not be tracked during the static calibration.

Initial Contact Angles. RM MANOVA results for the calcaneonavicular complex revealed significant results between step heights ($F_{12,4}$=17.639, $p = 0.007$). Sphericity tests for the follow-up RM ANOVAs indicated that sphericity could be assumed in the sagittal plane ($p = 0.448$), but not in the frontal ($p < 0.001$) or transverse planes ($p = 0.045$). When sphericity could not be assumed the Greenhouse-Geisser correction was used. Follow-up univariate tests resulted in no significant differences in the sagittal plane ($F_{4,1} = 1.386$; $p = 0.25$) or frontal plane ($F_{2.18,1} = 0.699$; $p = 0.516$); however, the transverse plane differences were significant ($F_{2.54} = 7.0$; $p = 0.001$). Follow-up dependent t-tests with a Bonferroni adjustment ($\alpha = 0.005$) between each step height were then performed to determine between which step heights the significant differences occurred. Results of the follow-up dependent t-tests revealed that the calcaneonavicular complex was significantly more adducted at initial contact at the 5-cm step than at the 20 and 25-cm step heights. There were no significant differences between the other step heights (Table 3). RM MANOVA results for the rearfoot ($F_{12,4} = 0.914$; $p = 0.597$; $\eta^2 = 0.733$) , medial forefoot ($F_{12,5} = 2.493$; $p = 0.436$; $\eta^2 = 0. 857$), lateral forefoot ($F_{12,5} = 0.566$; $p = 0.805$; $\eta^2 = 0.576$) and calcaneocuboid ($F_{12,4} = 1.877$; $p = 0.286$; $\eta^2 = 0.849$) complexes did not reveal significant sagittal, frontal, or transverse plane differences between the step heights (Tables 4 & 5).

Table 3. Transverse Plane Calcaneonavicular Complex Dependent T-test Results Between Step Heights

Steps	t (df)	p
5-10 cm	1.761 (15)	0.099
5-15 cm	2.705 (15)	0.016
5-20 cm	3.351 (15)	0.004*
5-25 cm	4.473 (15)	< 0.001*
10-15 cm	2.069 (15)	0.056
10-20 cm	2.392 (15)	0.030
10-25 cm	2.493 (15)	0.025
15-20 cm	1.107 (15)	0.286
15-25 cm	0.822 (15)	0.424
20-25 cm	-0.141 (15)	0.89
*Significance with Bonferroni correction at α = 0.005		

Table 4. RM MANOVA Across Sagittal, Frontal and Transverse Plane Initial Contact Angles

	F (df)	p	Partial Eta Squared
Rearfoot complex	0.914 (12,4)	0.597	0.733
Medial forefoot	2.493 (12,5)	0.436	0.857
Lateral forefoot	0.566 (12,5)	0.805	0.576
Calcaneocuboid	1.877 (12,4)	0.286	0.849
Calcaneonavicular	17.639 (12,4)	0.007*	0.981
*Significance at α=.05			

Table 5. Initial Contact Angles Mean & SD (degrees)

		5 cm step		10 cm step		15 cm step		20 cm step		25 cm step	
		Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Rearfoot complex	Sagittal	4.12	5.10	2.88	8.55	-1.79	12.82	-3.37	14.14	-10.22	11.89
	Frontal	3.59	4.03	3.40	3.96	3.37	3.96	3.22	4.33	3.88	3.72
	Transverse	-2.18	4.36	-2.69	3.79	-1.96	4.81	-1.29	5.19	0.00	4.96
Calcaneo navicular complex	Sagittal	-8.76	8.41	-9.60	8.85	-8.61	9.67	-9.81	8.67	-8.51	8.13
	Frontal	-4.27	6.42	-3.34	6.34	-3.00	5.94	-3.33	6.05	-3.87	5.17
	Transverse	3.46	4.72	2.20	4.78	0.88	3.90	0.31	4.90	0.39	3.85
Medial forefoot	Sagittal	6.98	8.20	6.26	9.34	3.98	10.93	3.69	12.56	-1.38	10.49
	Frontal	1.57	7.81	0.60	7.32	0.30	5.79	-0.11	5.49	-0.36	4.97
	Transverse	-0.49	8.05	-0.33	6.44	0.69	6.65	1.09	7.21	1.35	5.05
Lateral forefoot	Sagittal	1.40	3.47	1.64	3.50	1.47	3.36	1.07	3.75	0.48	3.28
	Frontal	5.70	7.33	5.71	6.28	4.75	6.37	4.23	6.70	3.54	5.21
	Transverse	3.61	3.71	3.93	3.63	3.59	3.69	3.92	4.42	3.34	4.13
Calcaneo cuboid joint	Sagittal	-1.73	6.23	-2.75	6.78	-4.30	8.43	-5.27	8.53	-8.10	8.45
	Frontal	-4.59	5.46	-5.07	4.57	-4.36	3.67	-3.19	4.69	-3.23	5.02
	Transverse	-0.41	2.83	-0.84	2.71	-1.11	4.22	-1.38	5.42	-0.74	4.66

Range of Motion. RM MANOVA results, did not reveal any significant range of motion differences across step heights for any of the articulations of the foot in any plane (Tables 6 & 7).

Table 6. Sagittal, Frontal and Transverse Plane ROM RM MANOVA Results

	F (df)	p	Partial Eta Squared
Rearfoot complex	1.084 (12,4)	0.517	0.765
Medial forefoot	1.640 (12,5)	0.305	0.797
Lateral forefoot	1.251 (12,5)	0.429	0.750
Calcaneocuboid	1.928 (12,4)	0.276	0.853
Calcaneonavicular	3.130 (12,4)	0.140	0.904
*Significance at α = 0.05			

Table 7. Weight Acceptance Phase ROM Means & SD (degrees)

		5 cm step		10 cm step		15 cm step		20 cm step		25 cm step	
		Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Rearfoot complex	Sagittal	8.60	3.63	9.32	4.18	12.25	7.52	14.36	8.60	17.74	8.70
	Frontal	6.02	2.23	6.30	2.61	6.45	3.05	6.12	2.30	6.70	2.84
	Transverse	7.31	2.58	6.83	2.41	7.18	2.64	7.69	3.65	8.45	3.53
Calcaneo navicular complex	Sagittal	4.56	1.93	5.20	2.03	4.41	2.29	5.23	2.68	4.51	2.52
	Frontal	6.89	2.64	6.58	2.79	5.05	2.43	4.90	2.15	4.52	2.39
	Transverse	4.36	2.37	4.31	2.03	3.63	1.18	3.21	1.53	3.23	1.11
Medial forefoot	Sagittal	9.07	5.38	10.23	5.74	11.76	7.52	12.56	8.74	15.98	8.23
	Frontal	7.67	4.32	7.05	3.76	6.24	4.07	6.01	4.60	5.20	2.83
	Transverse	2.33	1.05	3.17	1.86	2.95	1.63	3.09	1.80	3.75	2.31
Lateral forefoot	Sagittal	2.88	1.38	2.91	1.79	3.14	2.00	3.83	2.82	4.52	2.85
	Frontal	8.13	2.59	7.83	3.03	6.98	3.72	6.78	3.27	6.06	3.41
	Transverse	7.99	2.93	8.45	3.06	8.00	2.54	8.45	3.17	7.93	2.70
Calcaneo cuboid joint	Sagittal	5.21	3.46	6.80	5.28	8.30	6.02	9.65	5.79	11.38	6.88
	Frontal	4.34	1.47	3.98	1.43	3.86	1.71	3.57	1.53	3.10	1.07
	Transverse	3.74	1.92	3.70	1.64	3.78	1.52	3.98	2.67	3.59	2.33

EMG

The timing of muscle activation was to be compared across step heights, however, inspection of the EMG data suggested the peroneals were constantly active prior to initial contact at all step heights even when utilizing a threshold as large as five standard deviations above static standing calibration activity levels. Determining onset visually was also unclear (Figure 6). Therefore, only magnitude of muscle activity was quantified during the step heights. Visual inspection of the EMG data for each participant indicated that 21 of the 22 subjects (12 female, 9 male) had clean EMG recordings (all data trials with similar shape and amplitudes) for at least eight out of the ten trials using their preferred step down strategy. Following inspection of the trials, each subject's clean EMG recordings from their preferred step down strategy were averaged and then normalized to peak amplitude at the 25-cm step height (Delahunt, Monaghan, & Caulfield, 2006a) (Figure 7). Dependent t-tests with Bonferroni corrections showed significant increases in integrated EMG in the 200 ms prior to initial contact between the: 5-cm step and the 15, 20 and 25-cm step heights; 10-cm step height and the 20 and 25-cm step heights; and 15-cm and 20-cm step heights (Table 8).

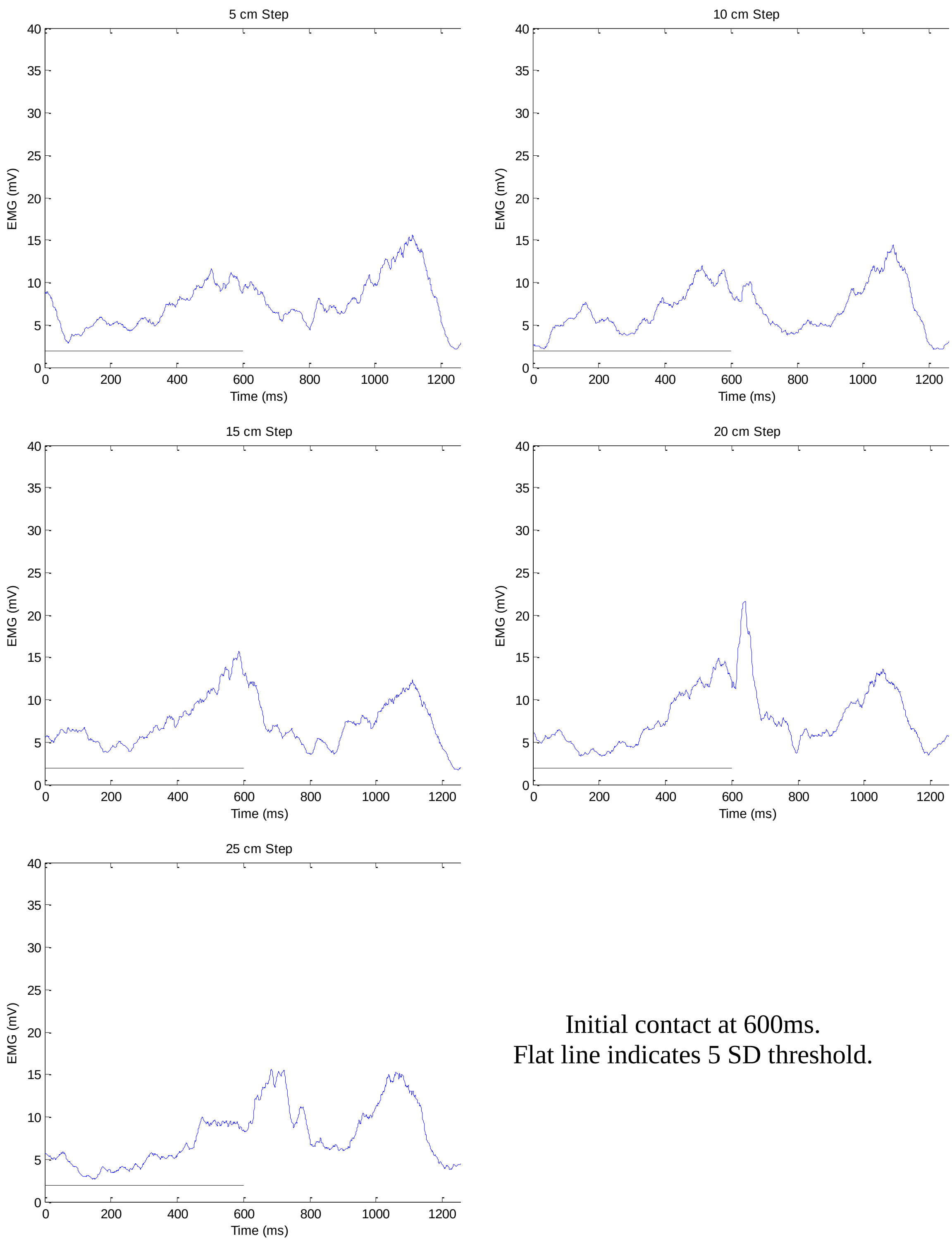


Figure 6. Representative data of one participant's averaged 10 trials at each step height.

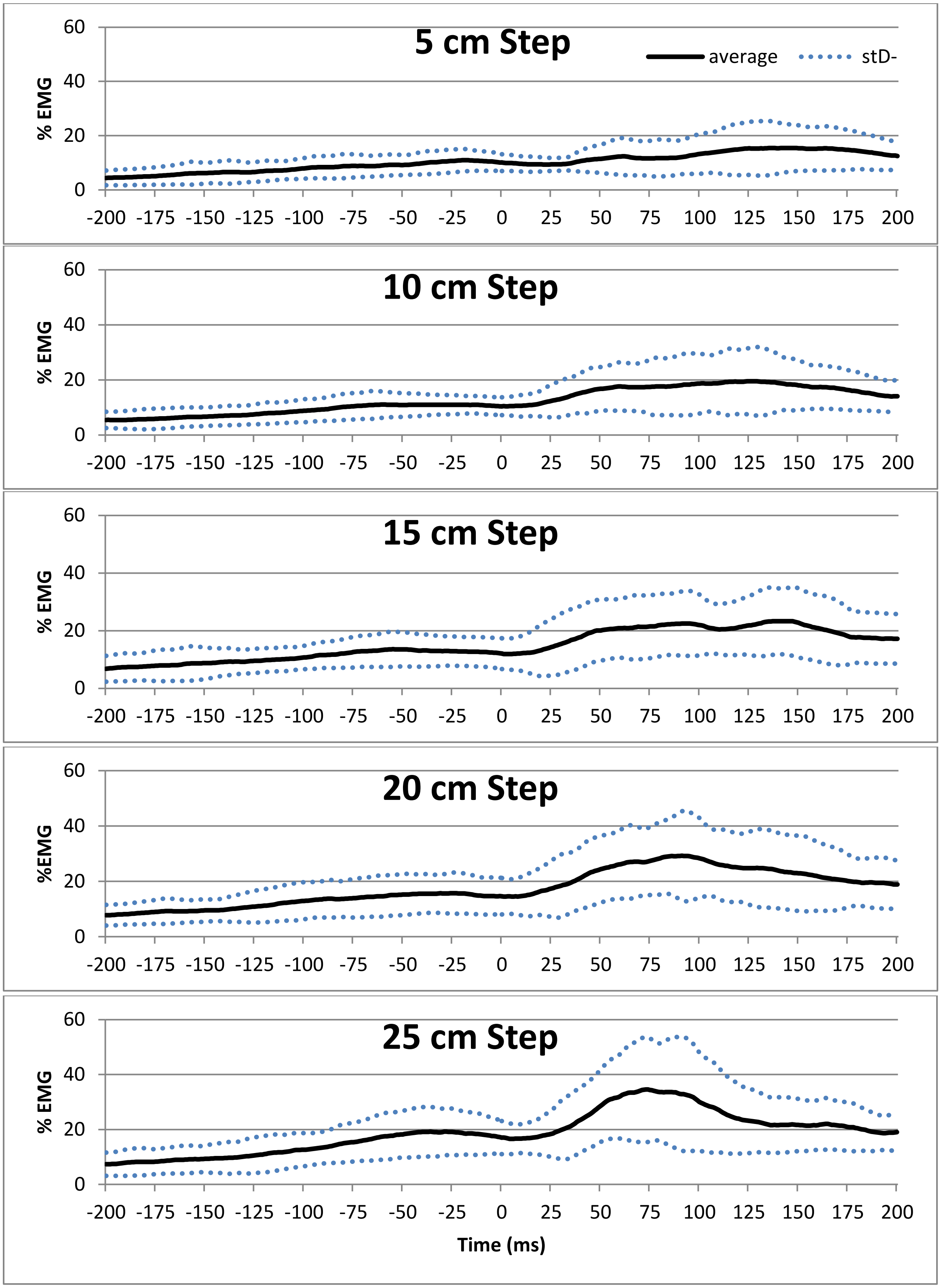


Figure 7. Normalized EMG 200 ms pre- and post- initial contact across subjects. Initial contact occurred at Time = 0 ms.

Table 8. EMG Dependent T-test Results.

	t	df	p
step 5 - step 10	-2.235	20	0.037
step 5 - step 15	-3.557	20	0.002*
step 5 - step 20	-4.229	20	< 0.001*
step 5 - step 25	-4.506	20	< 0.001*
step 10 - step 15	-2.754	20	0.012
step 10 - step 20	-3.797	20	0.001*
step 10 - step 25	-4.15	20	< 0.001*
step 15 - step 20	-3.242	20	0.004*
step 15 - step 25	-3.104	20	0.006
step 20 - step 25	-1.553	20	0.136
* Significant at α= 0.005			

Chapter 5: Discussion

The purpose of this study was to identify the foot and ankle kinematics and lower extremity muscle activity of uninjured individuals during descent from varying step heights. It was hypothesized initial ground contact would occur with the heel at low step heights and the forefoot at higher step heights. These changes in contact position were postulated to be accomplished through significant kinematic differences within the articulations of the distal foot at initial contact angle and during the weight acceptance period. Lower extremity muscle activity of the peroneals was anticipated to increase as step height increased and initial contact shifted from the heel to the forefoot.

Step Landing

As hypothesized the preferred step landing strategy during the lowest two step heights was heel strike (5 cm – 218 heel strike/220 total step downs; 10 cm – 182 heel/220 total), while the highest two steps preferred forefoot landings (20 cm – 120 forefoot strike/220 total step downs; 25 cm – 154 forefoot/220 total). A previous visual assessment study by Freedman and Kent (1987), examining preferred landing in five trials at three step heights, also reported that most subjects switched foot contact strategy between steps of 5-cm and 20-cm (5 cm – 53 heel/55 total; 10 cm – 49 heel/55 total; 20 cm – 43 forefoot/55 total). Examining the preferred landing strategy results in the current study by participant, only one of the 21 participants in the study landed with the forefoot at the 5-cm step height while all but five of the 21 participants preferred landing on the forefoot at the 25-cm step height. The change in landing strategy preference may be clinically relevant due to the fact that the plantarflexed position of the ankle joint during the forefoot landing strategy results in decreased bony structure joint stability thus placing greater dependency upon ligaments and muscles (dynamic stabilizers) of the ankle and foot. Knowledge of step heights at which persons typically transition from a heel to forefoot strike pattern may be beneficial for clinicians developing rehabilitation programs for patients with ankle pathologies.

Kinematics

As anticipated, the preferred step landing strategy transitioned from rearfoot to forefoot with an increase in step height; however, the initial contact and range of motion results did not demonstrate as many significant step height differences as anticipated.

Initial Contact Angles. It was hypothesized that as step height increased and landing strategy transitioned to the forefoot, the joints of the foot would become more plantarflexed, inverted, and internally rotated (supinated)/adducted to increase the stability of the medial longitudinal arch. Differences in initial contact angles were found only in the calcaneonavicular complex and only in the transverse plane. Furthermore, the change in initial contact position was contrary to our hypothesis. As the step height increased the calcaneonavicular complex became less adducted, which may be associated with decreased medial arch rigidity. This unexpected difference may be due to the need of the midfoot to be slightly pronated to position the forefoot for weight acceptance. This slight pronation would place greater reliance on ligaments and muscles rather than bony structure to stabilize the medial midfoot. Due to differences in multi-segment foot model definitions and lack of previous studies examining single step height differences, there are no transverse plane midfoot initial contact position data to which the results of the current study may be compared. However, although the differences were not significant across step heights in the current study, the sagittal plane initial contact angles of the rearfoot complex are generally consistent with the contact angles found in previous studies, given the differences in foot segmentation and footwear (Table 9) (Buckley, MacLellan, Tucker, Scally, & Bennett, 2008; Mian, Thom, Narici, & Baltzopoulos, 2007; Protopapadaki et al., 2007). Additionally although differences within the other articulations were not statistically significant, the effect sizes were large (η^2 >0.14). The consistency of the initial contact angles in the current study with previous studies reporting significant step descent differences and the large effect sizes associated with the present study may indicate the presence of other clinically meaningful differences warranting further investigation. This may be particularly true in the sagittal plane at the rearfoot complex, medial forefoot and calcaneocuboid which all had differences of at least six degrees between the two extreme step heights. Identification of changes in landing strategy and/or kinematic differences due to step height may help in isolating differences between injured and healthy persons.

Table 9. Initial Contact Sagittal Plane Ankle Angles Across Studies

Study	Step Height	Foot Segment	Contact Angle (±SD) degrees	Footwear
Buckley et al. (2008)	Averaged: 7.4, 14.5 & 21.8 cm	Single	~ -7 (20)	Shoes
Mian et al. (2007)	17 cm multi-step	Single	~ -18 (NA)	NA
Protopapadaki et al. (2007)	18 cm multi-step	Single	~ -25 (~7)	Barefoot
Current study	15 cm	Rearfoot	-1.79 (12.82)	Sandals
	20 cm		-3.37 (14.14)	

Range of Motion. In the rearfoot complex and medial forefoot it was hypothesized range of motion would increase with increased step heights. Both the medial and lateral forefoot ranges of motion were anticipated to decrease to facilitate foot stability in the plantarflexed position. Although not statistically different, the sagittal plane ROM across all but the calcaneonavicular complex increased as step height increased, additionally, effect sizes were large (> 0.14) for all joints of the foot (Tables 6 & 7). In addition, the sagittal plane rearfoot complex range of motion results in the current study are consistent with previous literature. Level or sloped walking generally resulted in 7-10 degrees of motion of the rearfoot in the sagittal plane (Bruening, Cooney, & Buczek, 2012a; Jenkyn et al., 2009; Lundgren et al., 2008; Tulchin et al., 2010), which is consistent with the magnitude of motion found with lowest step in this study. Prior studies examining step heights of 10 cm or more (Rao et al., 2009; Riener et al., 2002) found range of motion in the sagittal plane within the weight acceptance phase to be from 10 – 17 degrees, which also corresponds to this study's findings. The amount of change across step heights and large effect sizes suggest that further investigation is warranted.

EMG

As hypothesized, there was in increase in peroneal activity as the step height increased (Table 10), suggesting an increased need for muscle activation to provide frontal plane stability and medial longitudinal arch stability with a change in landing strategy. The results suggest that a height difference of 10 cm or greater on lower steps initiates some level of adjustment regardless of landing strategy. A significant difference was also found between the 15-cm and 20-cm steps, the point at which the majority of participants switched landing strategies. As seen with the increase in step height, the foot changes landing position from dorsiflexed on lower steps to plantarflexed at higher step heights. With this transition, the amount of stability available through the ankle mortise is decreased, thus requiring more ligamentous and muscular activity to maintain joint stability as well as support for the medial longitudinal arch as the weight is accepted on the distal foot. Previously differences have been seen in peroneal activity between injured and healthy populations in a single task (Delahunt et al., 2006a, 2006b; Konradsen & Ravn, 1990); this study demonstrates peroneal activity can also vary across magnitude of the task in healthy subjects.

Table 10. 200ms Pre-Contact Normalized and Integrated EMG mean & SD (%ms)

Area of Integrated EMG			
	Mean	Std. Deviation	Std. Error Mean
step 5	0.041	0.027	0.006
step 10	0.046	0.026	0.005
step 15	0.054	0.028	0.0060
step 20	0.061	0.031	0.007
step 25	0.065	0.028	0.0060

Limitations

There are some limitations to this study which should be considered prior to drawing conclusions from the results. The participants all wore the same style sandal provided by the lab, however the sandals had a flat footbed, which may influence both joint motion and muscle activity and therefore comparison to barefoot or footwear with arch support conditions may not be appropriate. Furthermore, the decision to investigate participant's preferred landing strategy resulted in inclusion of rearfoot and forefoot strike patterns at each of the different step heights. As a result, it is possible the varying step landing strategies across participants, demonstrated by the large standard deviations (Table 4), may have masked differences that may have occurred if landing strategies were controlled at each height. The decision to use each participants' preferred landing strategy in subsequent data analysis rather than require participants to adopt a specific landing strategy was made to mimic their step approach in daily activity rather than requiring intentional effort toward a required strategy. However, a cursory examination of the 25-cm step height sagittal plane kinematics of the rearfoot, medial forefoot and calcaneocuboid complex of participants that maintained heel strike and individuals that switched to a forefoot landing strategy, generally showed initial contact angle and ranges of motion differences of greater than ten degrees. Additionally participants ambulated at their preferred approach speed, as they would during daily activity; variation across subjects' speeds could also be a potential limitation. Finally, with respect to peroneal muscle activity, the inclusion of a maximal voluntary contraction may have provided a more consistent means of normalizing the muscle activity data.

Further Research

Additional study within the healthy population analyzing differences in step height by landing strategy may identify differences that were masked in the current study due the inclusion of both landing strategies at each step height. Additionally further study of peroneal muscle activity during different time points within the step down cycle, such as post-contact time or examining short and long latency reflexes may also elucidate clinically relevant differences associated with step descent height. Inclusion of activity in other lower extremity muscles would also be valuable in better understanding the synergetic relationship between joint motion and muscular activity during descent of a single step. A comparison between a single step down and level walking could also be beneficial in determining potential areas of dysfunction that may contribute to increased risk of ankle injuries during step descent.

Further study of patients with foot/ankle pathologies such as chronic instability or osteoarthritis utilizing a similar protocol would also be valuable in determining where differences exist between uninjured and injured populations. Identifying differences between populations would enable clinicians to better tailor treatment and rehabilitation programs to the specific pathology.

Summary

In examining the differences of healthy individuals while walking down a single step of varying step heights several points are of note. The preferred landing strategy typically changed, from initial contact with the heel at the lower step heights to forefoot contact at higher heights. Across this change in landing strategy kinematic differences were expected; however, the extent of changes found were not as great as hypothesized. Regarding the initial contact angles, the only statistically significant difference was in the calcaneonavicular complex in the transverse plane where at the 5-cm step height the complex was more adducted compared to the two highest steps. Changes in range of motion within the distal foot were not significant, although the effect size for both ranges of motion and initial contact angle of all articulations were large. The differences demonstrated in both kinematic variables along with the large effect size warrants further study.

Peroneal pre-contact activity, determined by integrating the 200 ms pre-contact EMG envelopes, demonstrated increased integrated area as step height increased. Significant differences were found when comparing the lowest step heights to any step 10 cm or greater in height. Additionally there was significant difference between the 15-cm and 20-cm step heights, which was the point at which the majority of participants switched landing strategy. A function of the increased peroneal activity during a forefoot landing strategy may be to provide dynamic stability to the medial midfoot and medial longitudinal arch. This is necessary to compensate for greater instability associated with the decreased calcaneonavicular complex adduction. The possible link between peroneal activity and step landing strategy would be valuable for clinicians working with populations with injury or dysfunction of the peroneals.

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Appendix A: Phone Screening Questionnaire Step Study

Subject #_____

General Health Questionnaire

Are you between the ages of 18-49?

Do you have a current lower extremity injury?

Have you ever had surgery on your lower extremity?

Have you sprained your ankle within the last 6 months?

THE CAIT QUESTIONNAIRE

Please tick the ONE statement in EACH question that BEST describes your ankles.

	LEFT	RIGHT	Score
1. I have pain in my ankle			
Never	☐	☐	5
During sport	☐	☐	4
Running on uneven surfaces	☐	☐	3
Running on level surfaces	☐	☐	2
Walking on uneven surfaces	☐	☐	1
Walking on level surfaces	☐	☐	0
2. My ankle feels UNSTABLE			
Never	☐	☐	4
Sometimes during sport (not every time)	☐	☐	3
Frequently during sport (every time)	☐	☐	2
Sometimes during daily activity	☐	☐	1
Frequently during daily activity	☐	☐	0
3. When I make SHARP turns, my ankle feels UNSTABLE			
Never	☐	☐	3
Sometimes when running	☐	☐	2
Often when running	☐	☐	1
When walking	☐	☐	0
4. When going down the stairs, my ankle feels UNSTABLE			
Never	☐	☐	3
If I go fast	☐	☐	2
Occasionally	☐	☐	1
Always	☐	☐	0
5. My ankle feels UNSTABLE when standing on ONE leg			
Never	☐	☐	2
On the ball of my foot	☐	☐	1
With my foot flat	☐	☐	0
6. My ankle feels UNSTABLE when			
Never	☐	☐	3
I hop from side to side	☐	☐	2
I hop on the spot	☐	☐	1
When I jump	☐	☐	0
7. My ankle feels UNSTABLE when			
Never	☐	☐	4
I run on uneven surfaces	☐	☐	3
I jog on uneven surfaces	☐	☐	2
I walk on uneven surfaces	☐	☐	1
I walk on a flat surface	☐	☐	0
8. TYPICALLY, when I start to roll over (or "twist") on my ankle, I can stop it			
Immediately	☐	☐	3
Often	☐	☐	2
Sometimes	☐	☐	1
Never	☐	☐	0
I have never rolled over on my ankle	☐	☐	3
9. After a TYPICAL incident of my ankle rolling over, my ankle returns to "normal"			
Almost immediately	☐	☐	3
Less than one day	☐	☐	2
1–2 days	☐	☐	1
More than 2 days	☐	☐	0
I have never rolled over on my ankle	☐	☐	3

Appendix B: Recruitment flyer

Do You have Healthy Ankles?

University of Wisconsin – Milwaukee
Musculoskeletal Injury Biomechanics Laboratory, END 132

Title: Biomechanical analysis of foot and ankle kinematic and lower extremity muscle activity during descent from varying step heights

Purpose: The purpose of this study is to identify the foot and ankle motions and timing of lower extremity muscle activity of uninjured individuals during descent from varying step heights.

Who can participate?

- Males and Females
- Ages 18 to 40
- No major surgery to the lower extremity
- No recent (previous 6 months) history of ankle sprain
- No current lower extremity injury
- Do not wear bifocals

What will I do? (~2.25 hours):

- Preliminary procedures Phone screening (~15 min)
  -General Health and Ankle stability assessment
- Walking step down gait analysis (~ 2 h)
  -Ankle range of motion will be measured
  -Walk, step down a single step and continue walking for 5 different step heights (5 cm, 10 cm, 15 cm, 20 cm & 25cm)

Do I get paid??

- **YES!** Participants that complete the gait trials will receive **$20.00 in gift cards**

Questions?

Principal Investigator:
Stephen C. Cobb, Ph.D., LAT
Associate Professor
Department of Kinesiology
414.229.3369

Co-Investigator:
Emily Gerstle, BS
Graduate Student
Department of Kinesiology
414.229.5147

This research project has been approved by the University of Wisconsin-Milwaukee Institutional Review Board for the Protection of Human Subjects (IRB Protocol Number 14.235)

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Appendix C: Informed Consent

Informed Consent IRB Protocol Number: 14-235

Version: IRB Approval Date: 2/24/2014

UNIVERSITY OF WISCONSIN – MILWAUKEE CONSENT TO PARTICIPATE IN RESEARCH

THIS CONSENT FORM HAS BEEN APPROVED BY THE IRB FOR A ONE YEAR PERIOD

1. General Information

Study title:

- Biomechanical analysis of foot and ankle kinematic and lower extremity muscle activity during decent from varying step heights

Person in Charge of Study (Principal Investigator):

- Stephen Cobb, PhD, ATC, CSCS is the Principal Investigator for this study. Dr. Cobb is a faculty member in the Department of Kinesiology. Emily Gerstle, BS is the co-principal investigator for the study. Emily is a graduate student in the Department of Kinesiology.

2. Study Description

You are being asked to participate in a research study. Your participation is completely voluntary. You do not have to participate if you do not want to.

Study description:
The purpose of this study is to identify the foot and ankle motions and timing of lower extremity muscle activity of uninjured individuals during descent (step down) from varying step heights. A better understanding of the effect of step height on muscle activity and foot and ankle motion will help to clarify differences in persons with foot or ankle injuries. Little is known about motion in the joints of the foot during step down, establishing healthy baseline factors for comparison to injured populations is important.

Participant screening, data collection, and storage will be done in the Musculoskeletal Biomechanics Injury Laboratory (Enderis 132). 22 individuals will participate at UWM.

As a participant in this study, you will be asked to participate in a phone screening (~15 minutes) and attend 1 step gait analysis session (~2 h)(total 2.25 h)

3. Study Procedures

What will I be asked to do if I participate in the study?
If you agree to participate you will be asked to perform the below tasks
Initial screening will be by phone interview performed by the co-PI, which includes:

- General Health Questionnaire
    - To participate in this study, you must meet the following:
        - Age 18-40
        - No major surgery to the lower extremity
        - No current lower extremity injury
        - No recent ankle sprain (previous 6 months)
        - Must have a shoe size between a women's size 6 - men's size 13.
        - Not wear bifocals
- Cumberland Ankle Instability Test
    - This questionnaire includes questions about previous lower body injuries and possible problems with your feet or ankles to classify your ankles as stable or unstable.

You will be asked to report to the Musculoskeletal Biomechanics Injury Laboratory (END 132) for testing. All procedures will be performed by the PI or co-PI.

- Step Gait Assessment
    - Ankle range of motion will be measured with a hand held goniometer (protractor for measuring joint angles).
    - The step gait assessment will consist of practice trials and 10 successful recorded trials descending 5 different step heights (5 cm, 10 cm, 15 cm, 20 cm, 25 cm). You will walk along a runway with a step and force plate (used to measure forces between the ground and your foot) below the step. Trials will be at your comfortable walking speed while wearing sandals.
    - During the trials, there will be groups of small reflective markers placed on your legs and feet. The markers will be placed directly on your skin with double sided adhesive tape, liquid adhesive and secured with elastic tape. The marker positions will be recorded using a 10 camera Motion Analysis system. The camera system only records the marker reflections not your image, additionally a reference camera will record only the legs and feet to ensure acceptable foot placement. If you choose not to be recorded during the trials you will not be eligible for the study. To record muscle activity, adhesive electrodes will be placed over muscles on shaved and cleaned skin of the lower leg, which will be attached to a battery pack worn on a belt, to hold the electrodes in place they will be secured with elastic tape. The electrodes and EMG system will record electrical signals produced by your muscles as they are active.
      Prior to the walking trials, additional reflective markers will be placed on your feet and legs. The position of these markers will be recorded while you are sitting and standing still. The extra markers will be removed before the walking trials begin. A sitting recording of EMG activity will also be taken.

4. Risks and Minimizing Risks

What risks will I face by participating in this study?

The potential risks for your participation in this research study are minimal.
Physical Risks:
Less Likely

- Trip or fall from the step
- Allergic reaction to the liquid adhesive used to secure the markers (<2%)

Protection of Physical Risks:

Initial first aid and/or emergency care will be provided by the investigators. In the event of irritation or allergic reaction to the liquid adhesive, please inform the investigators as soon as possible. Follow-up care for UWM students will be referred to the Norris Health Center. Non-students will be referred to their primary care physician.

Risks to Privacy and Confidentiality:
Less Likely

- As private information will be collected, there is the possibility of breach of confidentiality (less than 1%)

Protection of Risks to Privacy and Confidentiality:
All data will be stored in a locked filing cabinet in a locked room, electronic data will be stored in a password protected computer and network drive accessible to the PI, co-PI and limited Musculoskeletal Biomechanics Injury Laboratory personnel. All data will be given a code unique to you, which will not contain any identifiers to your person. The key to the code will be stored separately with access only accessible to those actively involved in this study. Once data collection is complete the code will be destroyed. All appropriate measure to protect your information will be taken.

5. Benefits

Will I receive any benefit from my participation in this study?

- There are no benefits other than to further research

6. Study Costs and Compensation

Will I be charged anything for participating in this study?

- You will not be responsible for any of the costs from taking part in this research study.

Are subjects paid or given anything for being in the study?

- Participants completing the study will receive a $20 gift card.

7. Confidentiality

What happens to the information collected?
All information collected about you during the course of this study will be kept confidential to the extent permitted by law. We may decide to present what we find to others, or publish our results in scientific journals or at scientific conferences Information that identifies you personally will not be released without your written permission. Only the PI, co-PI and limited Musculoskeletal Biomechanics Injury Laboratory personnel will have access to the information. However, the Institutional Review Board at UW-Milwaukee or appropriate federal agencies like the Office for Human Research Protections may review this study's records.

As described under the section "Risks & Minimizing Risks" the confidentiality of your information and data will be secured.

8. Alternatives

Are there alternatives to participating in the study?

- There are no alternatives available to you other than not taking part in this study.

9. Voluntary Participation and Withdrawal

What happens if I decide not to be in this study?

Your participation in this study is entirely voluntary. You may choose not to take part in this study. If you decide to take part, you can change your mind later and withdraw from the study. You are free to not answer any questions or withdraw at any time. Your decision will not change any present or future relationships with the University of Wisconsin Milwaukee.

- If you withdraw from this study prior to completing all step trials, we will destroy all information we collect about you. Your decision not to participate or to withdraw early will not affect their grade or class standing.

10. Questions

Who do I contact for questions about this study?

For more information about the study or the study procedures or treatments, or to withdraw from the study, contact:

Stephen Cobb, PhD, LAT
Department of Kinesiology
PO Box 413
Milwaukee, WI 53201
(414) 229-3369

Who do I contact for questions about my rights or complaints towards my treatment as a research subject?

The Institutional Review Board may ask your name, but all complaints are kept in confidence.

Institutional Review Board
Human Research Protection Program
Department of University Safety and Assurances
University of Wisconsin – Milwaukee
P.O. Box 413
Milwaukee, WI 53201
(414) 229-3173

11. Signatures

Research Subject's Consent to Participate in Research:

To voluntarily agree to take part in this study, you must sign on the line below. If you choose to take part in this study, you may withdraw at any time. You are not giving up any of your legal rights by signing this form. Your signature below indicates that you have read or had read to you this entire consent form, including the risks and benefits, and have had all of your questions answered, and that you are 18 years of age or older.

____
Printed Name of Subject/ Legally Authorized Representative

____ _______________________
Signature of Subject/Legally Authorized Representative Date

Research Subject's Consent to Audio/Video/Photo Recording:

- INSERT IF you are audiotaping, videotaping or photographing individual subjects:

It is okay to videotape my lower extremity while I am in this study and use my videotaped data in the research.

Please initial: ____Yes ____No

Principal Investigator (or Designee)
I have given this research subject information on the study that is accurate and sufficient for the subject to fully understand the nature, risks and benefits of the study.

____ _______________________
Printed Name of Person Obtaining Consent Study Role

____ _______________________
Signature of Person Obtaining Consent Date