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BIOMECHANICAL STRATEGIES FOR HILL WALKING AS WELL AS THE TRANSITIONS BETWEEN LEVEL AND HILL SURFACES: COMPARISONS WITHIN FOOTWEAR TYPES AND AGE GROUPS

A Thesis in

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Abstract

Healthy adults navigate hills as well as the transitions between level and hill surfaces while maintaining a safe and stable walking pattern. It is important to understand how healthy adults safely navigate changing terrains and how this ability may be modified through rehabilitation interventions such as motor development or footwear type. Therefore, the purpose of this study was to characterize the temporal-spatial gait parameter and muscle activity changes that occur during uphill and downhill walking and the transitions between level and hill walking. We hypothesized that during hill walking and hill transitions, participants will alter gait parameters to achieve greater anterior-posterior and medial-lateral stability. Additionally, we hypothesized that these observed differences will be less in textured insole footwear worn during locomotion and that, compared to healthy adults, children age 3 – 6 will present greater differences in temporal-spatial gait parameters between the level and hill walking tasks. Results showed that the level to downhill transition poses the greatest challenge to both anterior-posterior and medial-lateral stability. Compared to level walking, speed and step length were significantly less during the L-DN stride and speed, step length, and stance time variability, step width, and base of support area significantly increased. During the same stride, wearing textured insoles resulted in less lower limb muscle activity and improved response to medial-lateral stability as measured by step width. Finally, during level, uphill, and downhill walking, gait variability is significantly greater in children as compared to adults and this variability significantly increases from level to downhill walking in the child group only. Combined, these results further the current knowledge of how humans adapt to terrain changes during locomotion.

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Chapter 1

Introduction

Healthy adults navigate hills as well as the transitions between level and hill surfaces while maintaining a safe and stable walking pattern. Complex walking tasks are continually encountered which can challenge anterior-posterior and medial-lateral stability. Even so, healthy adults are typically able to navigate these terrain changes while maintaining a safe and stable walking pattern. Individuals that can not easily adapt to these terrain changes reduce the amount of walking they do on a daily basis and increase their risk of falling when walking (Myers et al., 1996). In order to improve quality of life through increased mobility it is important to first understand how healthy adults safely navigate changing terrains and then how this ability may be modified through rehabilitation interventions such as footwear type or motor development.

Normal level walking has been studied in detail and can be characterized by several variables. For analysis, walking is divided into gait cycles where one gait cycle is defined as one complete stride, or the time between two equivalent events - for example, the time from one heel contact of the left foot to the next heel contact of the left foot (Perry, 1992). The gait cycle is further divided into the swing phase, when the foot is not in contact with the ground, and the stance phase, the time when the foot is in contact with the ground. During normal walking, these phases typically comprise 38% and 62% of the gait cycle respectively, and double support phase occurs during the overlap between stance phases of the left and right feet (Perry, 1992; Rose and Gamble, 1994).

During each gait cycle, temporal-spatial gait parameters are used describe the time and space dependent movements of walking. These gait parameters include speed, step length, step rate, step width, and stance time. Speed is the product of step length and step rate. In healthy adults, step length and rate range between 1.3 - 1.6 meters per stride and 0.92 - 1 stride per second, resulting in a speed of 1.2 - 1.6 meters per second (Kirtley, 2006). However, these traits vary in relation to the participant's leg length, and preferred walking speed, step length, and step frequency are chosen to optimize metabolic cost (Kuo, 2001). Preferred step width is also selected to minimize metabolic costs and maintain medial-lateral stability (Donelan et al., 2001; Donelan et al., 2004). Not only do these preferred values help describe normal, healthy gait, but deviations from these mean values also provides insight into when anterior-posterior and medial-lateral stability is compromised during walking (Said et al., 2008).

In order to measure levels of fall risk in the laboratory without eliciting a fall, quantifiable measures of stability will be utilized. Stability will be defined as a participant's resistance to perturbations during walking, experienced in either the anterior-posterior (forwards and backwards) or medial-lateral (left and right) directions. Anterior-posterior stability is quantified by mean values of speed, step length, stance time, and lower limb muscle activity. Medial-lateral stability is quantified by mean values of speed and stance time in addition to step width and base of support area. Reduced stability has been shown to be characterized by slower speeds, shorter and wider steps, and increased time in double support (Thies et al., 2005; Said et al., 2008). Slower speeds and shorter steps increase the time and ability to adapt to perturbations during walking, and wider steps with more time spent in double support increases the overall base of support during walking (Krebs et al., 2002). Although these changes in mean values represent a decrease in stability, it has been shown that changes in mean values correlate strongly to a fear of falling (Maki, 1997; Delbaere et al., 2009). Therefore, these changes are likely a proactive response to reduced stability tasks. In contrast, Maki (1997) showed that increases in stride to stride variability in speed, step length, double support time, and step width correlate strongly to future fall risk. Hence, stride to stride variability or within stride gait changes correlate to reactive responses to reduced stability. Due to these factors, it is important to measure both mean changes and stride to stride variability of gait parameters in order to quantify stability.

Although there is a certain level of stride to stride variability in normal, healthy walking, increases beyond this range often represent reactive, within stride gait adjustments to anterior-posterior stability requirements (Maki, 1997). Magnitude of stride to stride variability, as measured by coefficient of variance, provides insight into stability as controlled by the central nervous system and cognition (Hausdorff, 2007). Typically, increased within-subject stride to stride gait variability is a sign of compromised stability, yet variability of step width above and below the mean levels has been shown to predict future fall risk (Brach et al., 2005; Hausdorff, 2007). Therefore, slower walking speeds, shorter step lengths, increased support time, wider steps, and increased stride to stride variability will all be reviewed when studying walking stability.

Muscle activity plays an important role in walking and anterior-posterior stability control (Rose and Gamble, 1994). Muscles provide propulsive forces by controlling joint angles and improve stability by increasing joint stiffness (Winter, 1991). Referencing surface electromyography (EMG) data with kinematic data can provide insight into the functional control of the active muscle as there is a strong correlation between EMG peaks and gait cycle events (Yang and Winter, 1985; Winter and Yack, 1987; Rose and Gamble, 1994). Muscle activity of the tibialis anterior (TA) peaks during initial stance phase to control ankle extension and during mid to late swing phase to control toe clearance. Activity of the soleus (SOL) and lateral gastrocnemius (LG) peak during late stance phase to provide propulsive forces at the ankle and LG is also active during late swing to control foot placement (Winter, 1983). Activity of these muscles at other phases of the gait cycle relate to changes in walking task requirements and are often due to increased co-contraction and joint stiffness to aid in stability. Therefore, larger EMG amplitudes for the same participant during different walking tasks relate to increased propulsive or braking forces, or increased anterior-posterior stability requirements.

Variation in walking patterns, temporal-spatial parameters, and EMG activity are controlled by the central nervous system in response to afferent information from visual, vestibular, and proprioceptive feedback (Dietz, 2002; Hijmans et al., 2007). Although there is often a balance between these senses, compensatory affects take place as specific senses are prioritized for certain tasks. For instance, visual and vestibular senses are often not acute or immediate enough to detect small surface changes whereas proprioceptive feedback can result in a within step modulation to surface irregularities as small as 3 degrees (Klint et al., 2008). These proprioceptive cues, which create a conscious representation of the body's orientation in space, include the golgi tendon - sensing muscle force, muscle spindles - sensing muscle stretch, and mechanoreceptors in the muscle, tendon, joint, and skin - sensing pressure (Dietz, 2002). Cutaneous feedback at the foot is especially important for stability as receptors in the foot sense center of mass location during movement (Perry et al., 2001; Nurse et al., 2005; Hijmans et al., 2007). However, this ability is often limited in the elderly and individuals with Parkinson's disease or peripheral neuropathy (Dietz, 2002; Hijmans et al., 2007). Fortunately, it is also possible to increase or decrease cutaneous feedback by altering footwear to increase tactile responses (Collins et al., 2003; Palluel et al., 2008; Perry et al., 2008).

Limitations in muscle activity and proprioception are common among both older adults (Prince et al., 1997) and children (Michel et al., 2010). Compared to healthy young adults, older adults are at an increased risk of falling (Prince et al., 1997; CDC, 2007) and walk with slower speeds, shorter and wider steps, spend longer time in stance, and have greater stride to stride variability during normal level walking (Nigg and Skeleryk, 1988; Brach et al., 2010). Additionally, children between the ages of 3 and 6 also walk at slower speeds, with shorter steps, longer percent time in stance, and have greater stride to stride variability (Sutherland et al., 1980; Rose-Jacobs, 1983; Dusing and Thorpe, 2007). Although these measures are dependent upon body size, these results are due to both growth and motor development of children (Sutherland et al., 1980; Kingsnorth and Schmuckler, 2000).

Hill walking can pose a challenging stability demand for healthy young adults and children alike. During downhill walking at various slopes up to 12 degrees, speed and step length decreased, and during uphill walking speed and step frequency decreased (Kawamura et al., 1991; Sun et al., 1996). Similar reduction in step lengths have been found in children walking down hills (Gill et al., 2009) and during obstacle avoidance where step width variability is also modified (Vallis and McFadyen, 2005). Furthermore, hill walking elicits modifications in muscle activation patterns to meet changes in anterior-posterior stability and propulsion requirements (Tokuhiro et al., 1985; Earhart and Bastian, 2000). Activity of the lateral gastrocnemius and soleus begins earlier in stance phase and remains active for a longer period of time during uphill and downhill walking (Lay et al., 2007). It is evident that in addition to age and motor development, challenging walking tasks such as hill walking require changes in gait patterns in order to maintain anterior-posterior and medial-lateral stability.

The purpose of this thesis was to investigate the various changes in locomotion that occur during hill walking tasks. To facilitate this, a series of three studies were conducted. First to characterize the gait changes during hill walking among healthy young adults and then to examine two special cases: changes that occur during hill walking in modified footwear, and differences in hill walking among children. Each study examined temporal-spatial gait parameters comparing participant data during 15 degree uphill and downhill walking tasks to the same participant's data during level walking. Due to the complex walking task and the study specific conditions, we first hypothesized that during hill walking and hill transitions, participants will alter gait parameters to achieve greater anterior-posterior and medial-lateral stability. Second, we hypothesized that these observed differences between level and hill walking will be less while wearing textured insole footwear during locomotion. Lastly, we hypothesize that, compared to healthy adults, children age 3 – 6 will exhibit greater differences in temporal-spatial gait parameters between the level and hill walking tasks.

Chapter 2

Walking Strategies During the Transition Between Level and Hill Surfaces

2.1 Introduction

Healthy, young adults are typically able to maintain stability while walking in the complex, outdoor environment. These individuals continually transition between level and hill surfaces of various angles while walking at fluctuating speeds. Surface transitions have the potential to decrease stability in both the anterior-posterior and medial-lateral directions. Therefore, it is plausible that healthy, young adults adopt a distinct gait strategy during transition strides in order to maintain stability.

Past studies differentiate between gait strategies by evaluating spatial-temporal parameters such as speed, step length, stance time, and step width (Myers et al., 1996; Maki, 1997; Balash et al., 2007; Delbaere et al., 2009). For instance, in an effort to improve anterior-posterior stability, adults walk with a slower speed, shorter step length, and greater stance time. In addition, in order to enhance medial-lateral stability, adults walk with a larger step width by maintaining a greater distance between the heels or toes. Together, changes in the mean magnitude of these parameters illustrate that individuals modify their gait patterns as a precaution to a future decrease in stability. Moreover, changes in stride-to-stride variability of speed, step length, and stance time illustrate that individuals modify their patterns as a reaction to a current decrease in stability (Maki, 1997). Thus, when comparing spatial-temporal gait strategies between strides of level, hill, and transition walking, it is important to examine both *mean* differences as well as *variability* differences.

Previous investigators have examined walking kinematics on hill surfaces but not during the transition between level and hill surfaces. Kawamura et al. (Kawamura et al., 1991) measured speed, step length, stance time, and step width during walking on a 12 degree hill. They demonstrated that speed was significantly slower during both uphill and downhill conditions. This slower speed was reflected in a decrease in step frequency during uphill walking and a decrease in step length during downhill walking. Stance time was also greater during uphill walking. Despite these indications that hill walking differs from level walking in the anterior-posterior direction, there was no significant difference in step width, indicating no stability changes in the medial-lateral direction. However, because Kawamura et al. (1991) only measured strides on the hill surface, it is unknown if the strides between surfaces elicit similar hill strategies or a distinct transition strategy.

The purpose of the present study was to analyze modifications in temporal-spatial parameters during hill walking transitions. We predicted that in comparison to level walking, the transition strides would indicate the adoption of a distinct gait strategy with a larger base of support to maintain both anterior-posterior and medial-lateral stability. More specifically, we hypothesized that speed and step length would decrease, while stance time and trial to trial variability of speed, step length, and stance time would increase as compared to level walking. We also hypothesized that step width at the heel and toe and the measure of base of support would be greater during hill walking.

2.2 Methods

Participants

Forty healthy college students, 17 men and 17 women (age = 21.78 (*1.70*) yr, height = 1.72 (*0.09*) m, mass = 71.83 (*15.58*) kg, mean (*standard deviation*)) completed the protocol. All of the participants gave written informed consent that followed the guidelines of The Pennsylvania State University Human Research Committee.

Protocol

Each participant completed a standing trial and a series of randomly assigned walking conditions on the level and hill surfaces. All of the walking trials were completed at a self-selected velocity along a 25 m walkway. We utilized a custom-built portable apparatus composed of a 2.4 m ramp inclined at 15 degrees continuous with a 4.8 m plateau (Figure 2.1). The minimum total walking distance was 14.2 m.

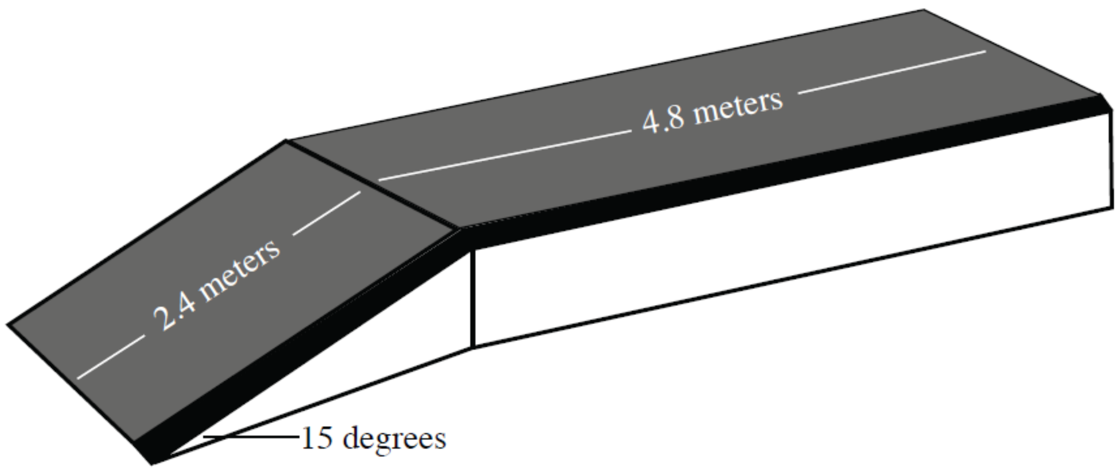

Figure 2.1 Portable ramp apparatus incorporated into a 25 m walkway.

A complete data set was composed of the successful completion of 5 level walking trials and 30 hill walking trials. Due to the limited collection volume of the motion analysis system, we shifted the apparatus to collect the appropriate stride. We collected 5 walking trials during the 2 hill only strides; downhill ramp (DN), and uphill ramp (UP), as well as, each of the following 4 transition strides; level plateau to downhill ramp (L-DN), downhill ramp to level floor (DN-L), level floor to uphill ramp (L-UP), and uphill ramp to level plateau (UP-L). A trial was defined as successful if a full stride of the left leg from toe-off to toe-off was captured in the collection volume. A transition stride was defined to include the left toe off and right foot contact on the first surface and the left foot contact (step 1) and right foot contact (step 2) on the new surface. A stride was defined in this way in order to evaluate the step length and step width during the first double support after the transition. Participants were instructed to begin walking with the left or right leg at the start of each trial to adjust for changes in stride characteristics between conditions.

Kinematics

We collected kinematic data with a six-camera, passive marker 3D photogrammetry system (Motion Analysis Corporation, Santa Rosa, CA). The calibration residual was less than 0.5 mm in a capture volume of approximately 2m x 2m x 2m. Prior to data collection, we placed retroreflective markers on the sacral crest as well as the shoes of each participant superficial to the posterior calcaneus and superior hallux. We collected the marker data at 100 Hz and post-processed the data with EVaRT software (Version 3.21, Motion Analysis Corporation). A custom-written Matlab program (Version

R2006b, The Mathworks, Natick, MA) was utilized for subsequent data processing that included a lowpass filter for the marker trajectories at 7 Hz (fourth-order, dual-pass, Butterworth).

Dependent Variables

After data collection, we evaluated the following temporal-spatial gait parameters during each of the hill and transition strides: (1) speed; the absolute value of the difference in anterior-posterior distance of the sacral crest marker at the first left toe off and second left toe off divided by the stride time, (2) step length; the absolute value of the difference in anterior-posterior marker location from right heel strike to left heel strike, (3) stance time; the time from left heel strike to left toe off, (4) heel and toe step width; the absolute value of the difference in medial-lateral marker coordinates for both the calcaneus (heel) markers and hallux (toe) markers during the first double support. In addition, we calculated the total area between both feet as a measure of base of support (BOS). The heel and toe markers during double support were used to calculate the area of the quadrilateral projected onto the transverse plane from left heel to left toe, left toe to right toe, right toe to right heel, and right heel to left heel. Finally, we evaluated the variability of speed, step length, and stance time by calculating the standard deviation between trials of each condition for each participant. This measure was used to evaluate the trial to trial variability for each participant as opposed to subject to subject variability which is represented for each variable as standard deviation.

Statistical Analysis

All data were analyzed across transition conditions using a repeated measures design (ANOVA). Where appropriate, we performed Newman-Keuls post hoc tests to analyze the differences between conditions and reported all values as mean $\pm$ standard deviation. Significance was defined as $p \leq 0.05$.

2.3 Results

In support of our hypothesis, compared to level walking, the base of support area was 20% greater during the L-DN, L-UP, and UP-L transition strides. Speed was slower during the L-DN and UP-L transitions, step length variability was greater during L-DN, L-UP, and UP-L and step width at the heel was larger during the L-DN and L-UP transitions (Table 1). Also in agreement of our hypothesis, stance time variability was greater and step width at the toe was larger during the L-DN transition. The ANOVA demonstrated that there was a significant main effect for condition for all of the dependent variables. However, these summaries only begin to describe how and when walking humans modulate their gait strategy during surface transitions.

We observed multiple significant changes in the spatial-temporal gait parameters of speed, step length, and stance time during hill transitions compared to level walking (Table 2.1, Figure 2.2). Mean speed during level walking was 1.39 m/s. During both the L-DN transition ($p < 0.001$) and UP-L transition ($p < 0.001$), speed was over 7% slower. In contrast, speed was 6% faster during the DN-L transition ($p < 0.001$). Step length values were significantly different for all transition strides from the average mean of

73.04 cm during level walking. Yet, step length during the L-DN condition was the only transition that supported our hypothesis with a value that was 9% shorter ($p < 0.01$) than level walking. Step length during all other transitions steps was, on average, 8% longer ($p < 0.001$) than level walking. Moreover, mean stance time was 631 milliseconds during level walking but was significantly less during L-DN and DN-L, 3% ($p < 0.001$) and 4% ($p < 0.001$), respectively.

Next, although variability for speed and stance time was only significantly greater during a single condition, L-DN, compared to level walking, variability for step length was greater during multiple hill transition strides (Table 2.1). Step length variability was not significantly different from level walking during DN-L, but was significantly greater during the remaining conditions, in particular, 209% greater during UP-L ($p < 0.001$).

Step width at the heel and toe also exhibited differences during hill transitions compared to level walking (Table 2.1, Figure 2.2). Heel step width was slightly more consistent across conditions with a mean maximum change of 2.49 cm as compared to a change of 3.03 cm at the toe. The two significant differences for step width at the heel were during L-DN and L-UP where step width was 18% ($p < 0.001$) and 10% ($p < 0.05$) larger respectively. Step width at the toe during the L-DN condition was also larger by 48% ($p < 0.01$). However, contrary to our hypothesis, step width at the toe was smaller during both L-UP and UP-L by at least 11% ($p < 0.05$).

As we hypothesized, in comparison to level walking, base of support, which includes step length and step width measures, was greater during the L-DN, L-UP, and

UP-L conditions by a minimum of 25% (Table 2.1). In contrast, due to the shorter step length, during the DN-L condition, BOS was 13% less than level walking ($p < 0.01$).

Compared to level walking, hill walking resulted in many significant differences but depending on the hill condition, these differences were often in different directions and of different magnitudes. During uphill walking, speed decreased while stance time, step length variability, step width at the toe, and base of support all increased. However, during both uphill transitions stance time was not significantly different than level walking and step length was significantly greater. During downhill walking, speed and step length decreased while step width at the toe increased. The DN-L condition resulted in opposite changes as speed and step length increased while stance time decreased. On the other hand, the L-DN condition resulted in significant differences for all variables. Similar to the DN condition, speed and step length decreased and step width at the toe increased. Additionally, stance time decreased and variability for all variables increased as well as step width at the heel and base of support.

In summary, compared to level walking, speed was slower during both DN and UP conditions by 5% ($p < 0.05$) and 10% ($p < 0.001$), respectively while step length was 11% shorter ($p < 0.001$) during the DN condition. Step width at the toe was 52% larger than level walking during the DN condition ($p < 0.01$) and 23% smaller during the UP condition ($p < 0.05$). To add, during the UP condition, BOS was 22% greater ($p < 0.01$) than level walking.

Table 2.1 Temporal-spatial parameters for each of the 5 walking conditions. Top Section: speed, step length, and stance time; Middle Section: variability calculated as the between trial variability of each participant for speed, step length variability, and stance time; Bottom Section: step width at the heel and toe, and base of support area. All data are represented as the mean raw values of each trial averaged for all 40 participants, mean (standard deviation). Italicized values represent level walking. Bold values represent a statistically significant difference as compared to level walking ($p < 0.05$). Shaded regions represent changes consistent with the direction hypothesized.

Dependent Variable	Transition Stride						
	L-DN	DN	DN-L	*LEVEL*	L-UP	UP	UP-L
Speed (m/s)	**1.25** (0.17)	**1.33** (0.19)	**1.47** (0.16)	*1.39 (0.12)*	1.36 (0.16)	**1.25** (0.13)	**1.28** (0.14)
Step Length (cm)	**66.9** (12.9)	**65.1** (9.5)	**81.2** (6.7)	*73.0 (5.5)*	**79.6** (8.5)	70.6 (5.7)	**76.0** (8.0)
Stance Time (ms)	**612** (17)	622 (30)	**605** (25)	*631 (20)*	**633** (22)	**648** (15)	**631** (21)
Speed Variability (m/s)	**0.1** (0.02)	0.04 (0.02)	0.04 (0.02)	*0.04 (0.02)*	0.03 (0.02)	0.04 (0.02)	0.04 (0.03)
Step Length Variability (cm)	**3.2** (1.1)	1.9 (0.0)	2.1 (1.0)	*1.8 (0.8)*	**2.5** (1.4)	**3.1** (1.2)	**3.5** (2.7)
Stance Time Variability (ms)	**14** (6)	13 (7)	11 (5)	*11 (5)*	13 (5)	12 (8)	13 (8)
Step Width Heel (cm)	**11.9** (2.9)	10.7 (2.9)	9.4 (3.4)	*10.2 (2.9)*	**11.2** (3.7)	10.2 (3.8)	11.1 (3.6)
Step Width Toe (cm)	**8.9** (5.3)	**8.8** (5.9)	6.4 (4.7)	*6.9 (4.8)*	6.1 (4.5)	**5.8** (4.7)	6.5 (5.6)
BOS Area (cm^2)	**416.2** (145.8)	384.7 (163.4)	**312.2** (111.5)	*364.5 (132.2)*	**427.7** (160.9)	**421.7** (153.0)	**430.8** (184.7)

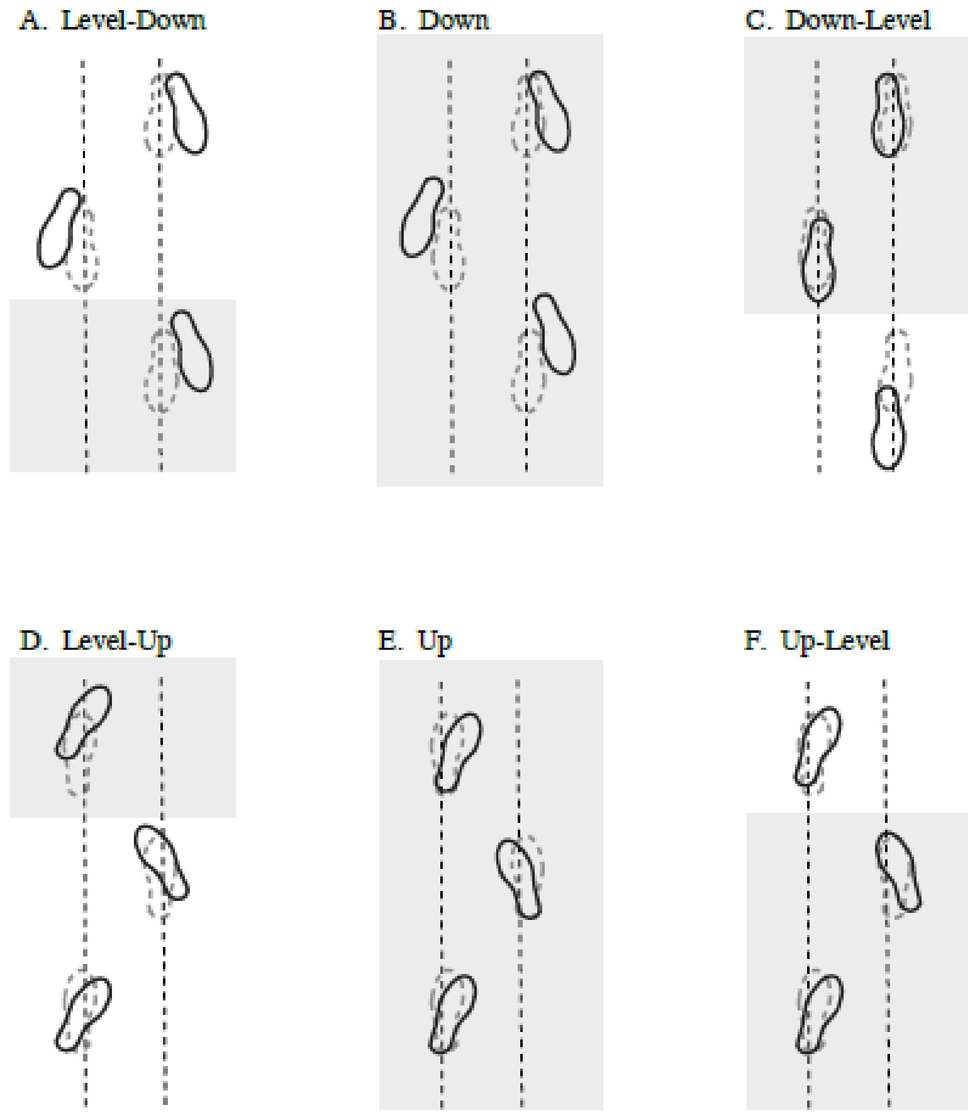

Figure 2.2 Illustration of level walking stride (dashed line) and transition strides (solid lines) for A. Level-Down, B. Down, C. Down-Level, D. Level-Up, E. Up and F. Up-Level. Shaded regions represent a hill surface and non-shaded regions represent a level surface. The foot placement locations represent actual change in step length as well as step width at the heel and toe.

2.4 Discussion

Our results illustrate that healthy, young adults did adopt a distinct gait strategy different from both level and hill walking during all of the transition strides. During the L-DN transition, speed, step length, and stance time were significantly less than level walking while variability, step width, and base of support were all greater (all values $p < 0.05$). With the exception of stance time, these results agree with our hypothesis that during hill walking speed and step length would be less than level walking, while stance time, variability of speed, step length, stance time, step width and base of support will all be greater. In contrast, during the DN-L transition, speed and step length were greater than level walking while stance time and base of support were less (all values $p < 0.05$). During both uphill transitions, L-UP and UP-L, speed was less than level walking, and step length variability and base of support were both greater, as hypothesized.

To begin, speed was significantly slower than level walking during the L-DN and UP-L transitions as hypothesized (Table 1). However, the average speed of 1.47 m/s during DN-L was 6% faster. This increase in speed, due in part to gravity, was the product of an 11% greater step length. In fact, the only transition with a significantly shorter step length was L-DN. It is possible that this stride was unique because transitioning from a level to a downhill surface does not require increased toe clearance. Chen et al. (1991) observed that step length increased in attempt to safely clear an obstacle during level walking. In fact, clearing an obstacle of minimal height, such as a line of tape on the floor, resulted in a 10% greater step length and this length increased more as the obstacle height increased (Chen et al., 1991). These results correspond closely to the 4-11% longer step length during the DN-L, L-UP, and UP-L conditions because of the increase in toe clearance due to the edge of the ramp.

Step length variability was also significantly greater during L-DN, L-UP, and UP-L compared to level walking (Table 1). Chen et al. (1991) reported that step length variability was greater when stepping over an obstacle and Maki et al. (1997) concluded that this increase was an indication of increased fall risk. Therefore, it is likely that participants perceived these strides as an obstacle and they attempted various strategies to maintain anterior-posterior stability during the transition.

Step width at the heel and toe exhibited significant changes in multiple conditions (Table 2.1). As hypothesized, the step width at the heel was larger during L-DN as participants adopted an increased base of support to improve medial-lateral stability. It is possible that there are metabolic determinants to these step width modifications. Donelan et al. (Donelan et al., 2001) utilized a simple model to determine that the cost of step-to-step transitions is minimized at narrow step widths. In a subsequent study, Donelan et al. (2004) manipulated step width while measuring metabolic cost and modeled the walking mechanics. They concluded that walking humans prefer a step width that minimized metabolic cost while maintaining balance. So it appears that during L-DN, the participants in the current study prioritized the maintenance of balance over the minimization of metabolic cost as there was an increase in step width during this condition.

Similarly, toe step width was significantly larger during L-DN transition but was significantly smaller during both L-UP and UP-L (Table 1). We did not originally anticipate the smaller toe step width during the uphill walking conditions. It is possible that this modification is a strategy utilized to improve propulsion. Erdemir and Piazza (Erdemir and Piazza, 2002) support this idea and stated that internal/external rotation of the foot has the potential to modify the force generating capacity of the ankle extensors. Previous research has evaluated the effects of toe position and speed. For example, Ho et al. (Ho et al., 2000) discovered a correlation between narrow toe position and faster walking speed in children. Furthermore, Fuchs and Staheli (1996) observed that sprint runners utilize a similar in-toeing strategy. In sum, a smaller toe step width during stance may be functionally important for generating propulsive force.

Overall, our step width data, particularly at the toe, did not correspond with the ramp walking data of Kawamura et al. (Kawamura et al., 1991). They found no significant difference in step width between any of the uphill or downhill conditions in comparison to level. However our methodology differed in four critical aspects. First, they collected walking data on a ramp, but did not collect data during the transitions between these conditions. It is likely that the transitions between surfaces pose a greater risk of falling than the hill independently. Second, Kawamura et al. collected data on a 12-degree ramp, which was less steep than our 15-degree apparatus. So there may be a decisive hill angle at which participants widen step width in order to maintain balance. Third, they measured kinematics indirectly via a force plate on the level ground below the ramp apparatus. We measured foot placement at the toe and heel using marker data. It is possible that internal and external foot rotation was a crucial detail that was not detected from force data. Fourth, Kawamura et al. instructed their participants to complete the protocol in bare feet whereas our subjects wore recreational shoes. Thus, the cutaneous receptors stimulated during barefoot walking may have aided the participants in selecting a narrow step width due to the increased sensitivity.

Prentice et al. (Prentice et al., 2004) also conducted a thorough investigation on hill walking and more specifically the uphill transitions that provides ideal insight for the present study. They collected walking kinematic data during the approach to a ramp and during the first step on the ramp for 3, 6, 9, and 12-degree conditions. Both limb and trunk motion was modified in a scaled fashion in order to navigate surface slope transitions. During the initial portion of swing, the leg trajectory was exaggerated to ensure clearance, while during the terminal portion of swing, the trajectory was specific for the present slope (Prentice et al., 2004). Our data for the transition steps show similar trends. The participants could not optimally modify their gait patterns until they received additional sensory feedback regarding the specific condition. For example, during the L-UP transition, heel step width was larger than during level walking, while there was no change in toe step width. As progression up the hill continued, the toe step width was significantly smaller.

Finally, our measure of base of support, which included step length, heel step width, and toe step width, was significantly larger for both L-DN, L-UP, and UP-L transitions (Table 2.1). Interestingly, during the UP-L condition, toe step width was smaller for propulsion but step length was longer and heel step width was larger. These differences resulted in a greater base of support, which shows how both the need for increased propulsion and increased stability can be accounted for simultaneously during hill transitions.

The L-DN transition likely poses the highest level of fall risk as demonstrated by the adoption of a cautious gait strategy (Maki, 1997; Balash et al., 2007). During this transition, speed and step length were less than level walking while step length and stance time variability were greater to enhance anterior-posterior stability. To add, step width at both the heel and the toe were significantly larger to improve medial-lateral stability. Finally, despite selecting a shorter step length, base of support area was larger to improve overall stability. It is likely that the participants modified their gait at this transition due to a greater perceived risk of falling.

In the future, we plan to quantify these spatial-temporal gait parameters during walking hill transitions in older adults. Numerous studies have concluded that balance decreases with age as evidenced by decreased step width variability and larger step width magnitude (Maki, 1997; Brach et al., 2005; Schrager et al., 2008). If these two findings correlate to hill walking than it is possible that the transitions between level and hill surfaces could pose the highest risk for instability, thereby leading to falls.

Chapter 3

Altering Footwear Type Influences Gait During Level Walking and Downhill Transitions

3.1 Introduction

Cutaneous and proprioceptive feedback from the lower leg provides the central nervous system with information to maintain stability during standing and walking (Nurse et al., 2005; Hijmans et al., 2007). During walking, humans utilize this somatosensory information, as well as visual and vestibular feedback, to safely traverse changing terrain. Individuals with deteriorated sensory systems or peripheral nervous system disorders are at an increased risk of falls and fall related injuries, particularly older adults (Hijmans et al., 2007). Furthermore, the CDC has reported that falls while walking are the leading cause of accidental injury for all age groups (CDC, 2007). Therefore, it is important for individuals of all ages to attempt to reduce this risk through proper gait adjustments in response to cutaneous, proprioceptive, visual, and vestibular cues.

One proposed method to improve stability during standing and walking tasks is by selecting proper footwear. A variety of footwear shapes and textures have been shown to alter walking patterns (Simeonov et al., 2008; Menant et al., 2009). In fact, several footwear insoles have been developed in attempt to enhance stability during postural and walking tasks by increasing cutaneous and proprioceptive stimuli with electrical noise insoles (Collins et al., 2003) and raised border insoles (Perry et al., 2008) which both improved cutaneous sensation and medial-lateral stability. The raised border insoles tested by Perry et al. (Perry et al., 2008) did not result in any changes in speed, step length, stance time, or step width during level walking or hill platform walking. Palluel et al. (2008) examined spiked insoles commonly found in athletic sandals as a method of providing continuous mechanical stimuli to increase stability during standing and walking tasks. The study determined that spiked insoles increase postural control as measured by smaller deviations of center of pressure during standing and level walking (Palluel et al., 2008). A second study examining the affects of textured insoles tested multiple continuous surface textured insoles on frontal and sagittal plane stability during quiet standing as well as step width during level walking. No significant differences were found between any of the insole conditions for any of the three tests (Wilson et al., 2008). However, it is possible that the postural and walking tasks chosen did not pose a great enough challenge to the participants to require additional afferent feedback to aid in stability.

Changes in stimuli at the foot have also been shown to alter muscle activity during walking and running. For instance, Nurse et al. (2005) investigated the differences between textured and non-textured insoles on muscle activity during level walking. They found that tibialis anterior activity during the first 20% of the stance phase and soleus activity during the late stance phase were each 13% less during the textured insole condition as compared to the non-textured condition (Nurse et al., 2005). Similar to textured insoles, as compared to standard shoes, barefoot walking or running could also increase cutaneous feedback and alter lower limb muscle activity. During barefoot running as compared to running in shoes, activity of the tibialis anterior is reduced prior to foot strike but is greater following foot strike (von Tscharner et al., 2003). To examine reduced plantar sensation, Nurse and Nigg (Nurse and Nigg, 2001) and Eils et al. (2004) compared level walking while barefoot to level walking following a period of plantar surface icing. Although both studies found that altering sensory feedback resulted in changes in muscle activity, these changes were not necessarily consistent. Eils et al. (2004) found that during the iced condition, soleus and gastrocnemius activity was greater during early stance whereas tibialis anterior activity was less, and during mid stance and late stance all three muscles resulted in less activity during the ice condition as compared to the barefoot condition. In contrast, Nurse and Nigg (Nurse and Nigg, 2001) found that during mid stance the muscle activity of the tibialis anterior was greater than it was during barefoot walking.

Increased cutaneous and proprioceptive feedback could be of greater importance for motor control during hill walking as several kinematic changes are made in order to actively maintain both anterior-posterior and medial-lateral stability. An observational study noted that downhill walking resulted in slower speeds and smaller step lengths and that these differences increased linearly as a function of the slope of the ramp and age of the participant (Sun et al., 1996). Additionally, Kawamura et al. (1991) examined temporal spatial parameters such as step length, stride width, and speed during hill walking at a 12 degree incline and decline. Kawamura et al. reported that during downhill walking there was a decrease in step length and speed but no difference in step width (Kawamura et al., 1991). Hill walking also requires changes in lower limb muscle activity and duration (Klint et al., 2008) have shown that the triceps surae muscle length and activity are modulated within 1 step onto a 3 degree decline in response to proprioceptive and cutaneous feedback. More specifically, Lay et al. (2007) examined

EMG activity for downhill walking at 8.5 and 21 degree grades. During downhill walking, there was a significant increase in tibialis anterior activity, and the onset of peak muscle activity for the gastrocnemius and soleus muscles began earlier in stance phase and remained active for a greater percent of stance (Lay et al., 2007). Although hill walking elicits changes in temporal-spatial parameters and muscle activation, these changes vary as uphill and downhill slopes require modifications of propulsion, anterior-posterior stability, and medial-lateral stability.

One possible walking task that may require additional sensory information to maintain adequate stability is transitioning between hill walking and level walking. In a previous study we examined uphill and downhill walking as well as the transition strides between level and hill surfaces. Changes in temporal spatial gait parameters in the frontal and sagittal plane provide evidence that even healthy young adults are presented with anterior-posterior and medial-lateral stability challenges during hill walking. Specifically, the stride presenting the greatest difficulty as represented by decreased speed and step length and increased step width is the stride from an elevated level surface onto a declined surface (Gottschall et al., submitted). Since downhill walking, unlike uphill walking, does not require additional propulsive demands, the additional requirements for this stride are limited to stability demands. Therefore, changes in walking patterns as a result of footwear type may be most discernible during a downhill transition stride.

Hill walking and footwear type can both impact gait patterns and muscle activity, but little research has been conducted on the influence of footwear during hill walking tasks similar to those experienced during daily walking. Therefore, the purpose of this study was to determine if altering footwear to increase or decrease cutaneous stimuli could influence stability during a hill walking task. We tested multiple footwear types, similar to balance enhancing footwear that have previously been tested, during a 15 degree hill walking task previously identified as a walking task which results in cautious gait (Gottschall et al., submitted). Previous results have been inconsistent but have provided evidence that compared to regular footwear, textured insoles increase medial-lateral stability during level walking and reduce muscle activity of the TA and LG. Our research has also shown that the transition from a level to downhill requires additional anterior-posterior and medial-lateral stabilization; speed and step length are reduced while step width and lower limb muscle activity increases. Therefore, we hypothesize that compared to foam insoles, textured insoles will result in greater speeds and step lengths and narrower step width and less stance time, TA activity, and LG activity and that these changes will be more evident during the downhill transition stride as compared to the level stride. Additionally, we hypothesize that the greater differences between ice and barefoot conditions will be in the same direction as the changes between foam and textured insole conditions.

3.2 Methods

Participants

Ten healthy college students, 5 men and 5 women (age = 20.2 (*1.23*) yr, height = 1.72 (*0.06*) m, mass = 69.72 (*10.31*) kg, mean (*standard deviation*)) completed the protocol. All of the participants gave written informed consent that followed the guidelines of The Pennsylvania State University Human Research Committee.

Protocol

Each participant completed one initial standing trial and five trials in each of the five randomly assigned footwear conditions on the level and hill surface. All of the walking trials were completed at a self-selected velocity on a 25 m walkway. For the downhill transition, we utilized a custom-built portable apparatus composed of a 2.4 m ramp inclined at 15 degrees continuous with a 4.8 m plateau. The minimum total walking distance for each trial was 14.2 m.

A complete data set was the successful completion of five level walking trials and a total of five hill walking trials in each of the five footwear conditions: (1) personal athletic shoes, (2) personal athletic shoes with Dr. Scholl's 2X Air Pillow Insoles®, (3) personal athletic shoes with custom-built checker patterned textured foam insoles, (4) barefoot, and (5) barefoot following an icing protocol. Each of the three footwear conditions were randomly ordered and all level and ramp trials were completed before switching footwear. However, the barefoot and iced conditions were always the fourth and fifth conditions in order to prevent the ice condition from altering the results of the other footwear conditions. The icing protocol required the participant to stand on crushed ice enclosed in plastic for 10 minutes before the first walking trial, and one minute between every two walking trials. This method has previously been verified to reduce plantar surface sensation by monofilament pressure tests (Eils et al., 2004). The hill walking trials consisted of the stride from the elevated level surface to the downhill ramp.

A trial was defined as successful if a full stride of the left leg from toe-off to toe-off was captured in the motion analysis capture volume. A stride was defined in this way (-40% to 60%) in order to evaluate the swing phase over the hill transition and stance phase while the right foot was on the elevated level surface and the left foot was on the hill surface during the first double support phase (0-10%) and while both feet were on the hill surface during the second double support phase (50-60%).

Kinematics

We collected kinematic data with a six-camera, passive marker, 3D photogrametry system (Motion Analysis Corporation, Santa Rosa, CA). The calibration residual was less than 0.5 mm in a capture volume of approximately 2m x 2m x 2m. Prior to data collection, we placed retroreflective markers on the sacral crest and lateral malleoli as well as the shoes of each participant superficial to the posterior calcaneus and replaced markers over the skin for barefoot and ice conditions. We collected the marker data at 100 Hz and post-processed the data with EVaRT software (Version 3.21, Motion Analysis Corporation, Santa Rosa, CA). A purpose-written Matlab program (Version R2006b, The Mathworks, Natick, MA) was written for subsequent data processing that included a lowpass filter for the marker trajectories at 7 Hz (fourth-order, dual-pass, Butterworth).

Electromyography

We measured electromyography (EMG) signals using a wired amplifier system (Bortec Octopus AMT-8, Calgary, AB, Canada) with a bandpass filter setting of 5~500 Hz. We collected the data at 2000 Hz using EVaRT software, which also synchronized the marker and muscle data. Prior to electrode placement, we prepared the skin with fine sandpaper and rubbing alcohol. We placed 1 cm x 1.5 cm bipolar, silver-silver chloride, surface electrodes (Vermed, A10041, Bellows Falls, VT) over the tibialis anterior (TA), soleus (SOL), and lateral gastrocnemius (LG) muscles of the left leg according to the recommendations by Cram and Kasman (1998). For the TA, we placed the electrodes in parallel at half the distance from the patella to lateral malleolus. For the SOL, we placed the electrodes at an oblique angle below the lateral gastrocnemius. For LG, we placed the electrodes in parallel at one-third the distance from the head of the fibula to the calcaneus. The inter-electrode distance was 2 cm. We verified that the position of the electrodes was functionally correct and that cross talk between the muscles was negligible with a series of functional tests suggested by Winter et al. (1994), and Cram and Kasman (1998).

For the temporal analysis, we generated a linear envelope (Winter, 1991) obtained by low pass filtering (dual-pass, fourth order, Butterworth) the rectified EMG data at 30 Hz. For the amplitude analysis, we full-wave rectified the band-pass filtered signals, calculated the mean EMG amplitude (mEMG) and averaged five strides for each condition. Each stride was further divided into eight time segments. For each trial, heel strike was calculated by marker location and stance and swing phase was determined for the left leg. Swing phase was then categorized into: (1) initial swing from 0-34% of swing phase, (2) mid swing from 34-60% of swing phase, (3) terminal swing from 60-100% of swing phase. Stance phase was categorized into: (1) initial stance 0-20% of stance phase, (2) mid stance from 20-57% of stance phase, and (3) late stance from 57-81% of stance phase, (5) terminal stance from 81-100% of stance phase. For each subject, we normalized the mEMG for each experimental condition based on the mEMG for the level walking condition in personal athletic shoes.

Dependent Variables

We evaluated the following temporal-spatial gait parameters during each of the transition strides: (1) speed; the absolute value of the difference in anterior-posterior distance of the sacral crest marker at first left toe off and second left toe off divided by the stride time, (2) step length; the absolute value of the difference in anterior-posterior heel marker location from right heel strike to left heel strike, (3) stance time; the time from left heel strike to left toe off, (4) step width at the ankle; the absolute value of the difference in medial-lateral marker coordinates for malleoli markers (ankle) during the first and second double support of each stride.

Muscle activity was evaluated during separate intervals within each stride. The TA was analyzed during initial swing, terminal swing, initial contact, and terminal stance. These four areas are during times of peak activation as the TA controls toe clearance and initiates swing phase. The SOL and LG were both analyzed during late stance as they are activated for propulsion. Additionally, the LG was also analyzed during terminal swing as it is active to aid in anterior-posterior stability during foot placement (Winter, 1983).

Statistical Analysis

All data were analyzed across transition conditions within footwear type and across footwear type within transition conditions using a repeated measures design (ANOVA). Where appropriate, we performed Newman-Keuls post hoc tests and paired Student's t-tests to analyze the differences between conditions and reported all values as mean $\pm$ standard deviation. Significance was defined as $p \leq 0.05$.

3.3 Results

In support of our hypothesis, all variables exhibited significant differences between regular footwear and barefoot conditions and the mean step length was significantly longer during the textured insole condition as compared to the regular footwear condition. Also in support of our hypothesis, the hill transition resulted in greater anterior-posterior and medial-lateral instability as measured by significant differences in all dependent variables except stance time. As compared to level walking, the downhill transition resulted in slower speeds, shorter step lengths, wider steps (Figure 3.1, Figure 3.2), and greater activation of the TA at the end of stance phase and of the LG at the end of swing phase (Figure 3.3).

More specifically, walking speed in regular footwear was on average 1.31m/s during the level stride and 1.22m/s during the downhill transition stride, a 7% decrease ($p < 0.005$, Figure 1). During level walking, the average speed was significantly faster during both foam insole and textured insole conditions as compared to the ice condition ($p < 0.05$). The foam and textured insole conditions were 2% and 4% faster than the regular footwear condition respectively while the ice condition was 4% slower than the regular footwear condition. The textured insole condition was also significantly faster than the barefoot condition ($p < 0.05$). Similarly, during the downhill transition the regular, foam, and textured footwear conditions were all significantly faster than both the barefoot and ice conditions. The foam and textured insole conditions were each 1% faster than the regular footwear condition, whereas the barefoot and ice conditions were 7% and 11% slower than the regular footwear condition ($p < 0.05$). However, in either level or downhill transition stride, there were no significant differences between any of the shoe conditions or between the barefoot and ice conditions (Figure 3.1).

In contrast, step length measures did result in significant differences between shoe conditions and between barefoot and ice conditions. During the regular footwear condition, step length was an average of 69.6 cm over level ground and 65.9 cm over the downhill transition, 13% shorter ($p < 0.001$, Figure 3.1). During level walking the regular footwear average step length was 3% shorter than the textured insole condition and 5% longer than the ice condition ($p < 0.01$). Furthermore, both the foam and textured insole conditions were longer than the barefoot condition ($p < 0.01$); as compared to the regular footwear condition, foam and textured insole conditions were each 3% longer while the barefoot condition was 3% shorter. During the downhill transition stride, step length during the regular footwear condition was 9% longer than the barefoot condition and 17% longer than the ice condition ($p < 0.05$). Although not significantly different, step length during the foam insole condition was longer than the regular footwear condition and textured insole conditions were less than 1% shorter ($p = 0.19$ and 0.74 respectively). Both regular and textured insole conditions resulted in significantly longer steps as compared to both barefoot and ice conditions. Finally, step length during the barefoot condition was 8% longer than the ice condition ($p < 0.05$).

Stance time in regular footwear during level walking was 0.692s and during the downhill transition was 0.687s and was not significantly different (Figure 3.1). During level walking, stance time during the regular footwear condition was 5% longer than the barefoot condition ($p < 0.01$). Although not significant, stance time during the regular footwear condition was 1% shorter than the foam insole condition and 1% longer than the textured insole condition. Both foam and textured insole conditions were significantly longer than the barefoot condition. Yet, during the downhill transition the only significant differences were between the foam insole condition and the barefoot condition ($p < 0.05$). Stance time during the regular footwear condition was shorter than the foam insole condition and longer than the barefoot condition ($p = 0.50$ and 0.14 respectively). Contrary to both speed and step length, stance time resulted in no significant differences between level and downhill transition strides.

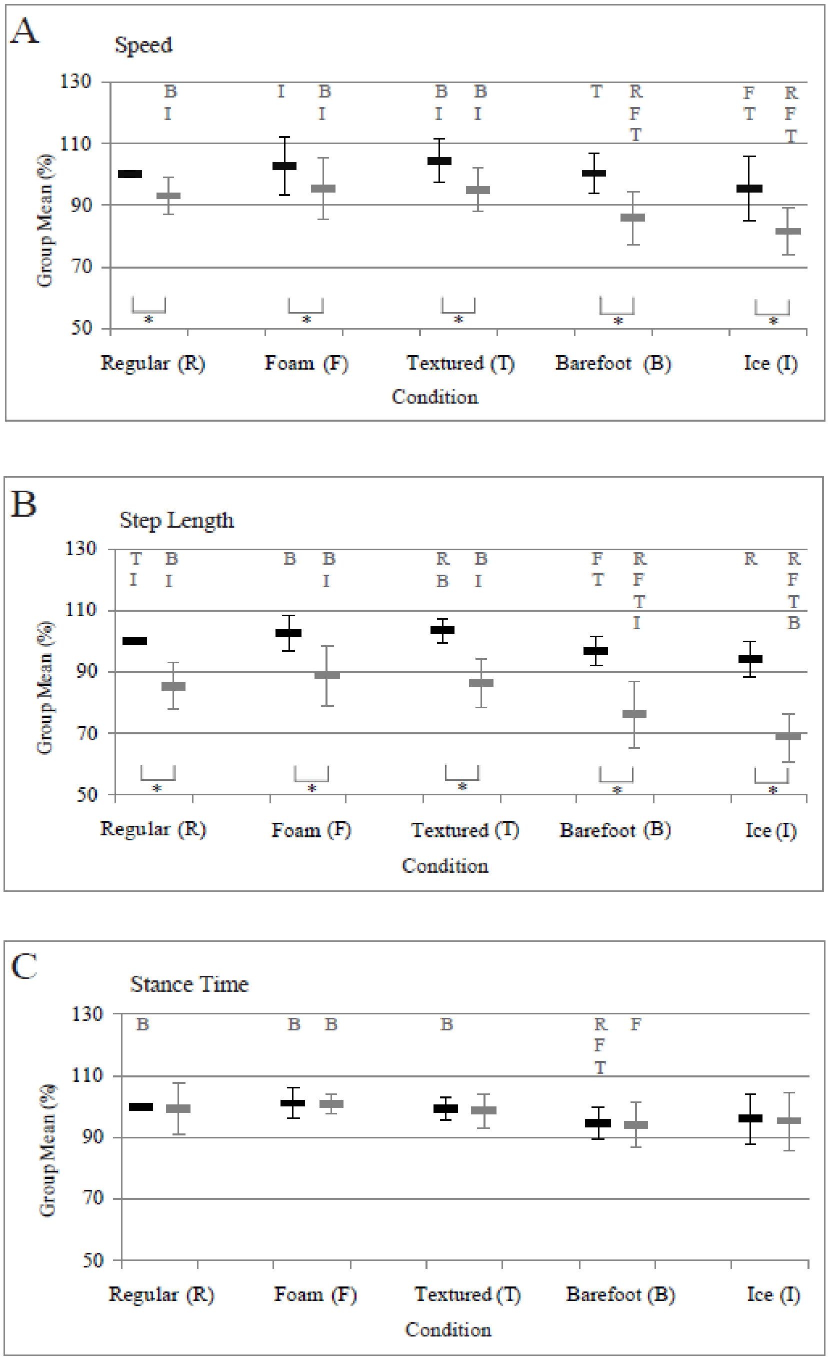

Figure 3.1 Speed (A), step length (B), and stance time (C) represented for each footwear condition paired between level (black) and downhill transition (gray). Mean and standard deviation of all 10 participants' average walking trials. Values represented as a percent of each participant's mean for all trials normalized to level walking in regular footwear. Significant differences between level and downhill pairs represented by a star (*) across the bottom ($p \leq 0.05$). Significant differences between footwear conditions represented by the letter of the paired condition (R, F, T, B, I) across the top ($p \leq 0.05$).

There were also multiple differences in step width values for both phases of double support. During the first double support phase the step width during the regular footwear condition was 0.235m during level walking and 0.232m during the downhill transition (Figure 3.2). This difference was not significant, and only the textured insole condition resulted in a large difference between level and downhill walking ($p = 0.054$). The regular footwear condition was 3% wider than the textured insole condition during level walking and 3% narrower during the downhill transition stride. Compared to the regular footwear condition, the iced condition was 2% wider and significantly greater than the textured insole condition during level walking ($p < 0.05$). During the downhill transition stride, the regular footwear condition was 5% narrower than the foam condition ($p = 0.05$) and 6% wider than the barefoot condition ($p < 0.01$). Additionally, during the downhill transition the barefoot condition resulted in a significantly narrower step width as compared to the foam and ice conditions ($p < 0.05$). The barefoot condition was 6% narrower than the regular footwear condition while the ice condition was only 1% narrower.

During the second double support phase the mean step width for the regular footwear condition was 0.223m during level walking and 0.247m during the downhill transition, 10% wider ($p < 0.005$, Figure 3.2). Additionally, the downhill transition stride resulted in a wider step as compared to level walking during the foam insole, textured insole, and ice conditions ($p < 0.05$). During level walking, the regular footwear condition was at most 1% different than any other footwear condition and there were no significant differences between footwear conditions. Similarly, during the downhill transition stride the regular footwear condition was less than 1% different than either foam insole or textured insole conditions. However, during the downhill transition stride the regular footwear condition was 8% wider than the barefoot condition ($p < 0.05$).

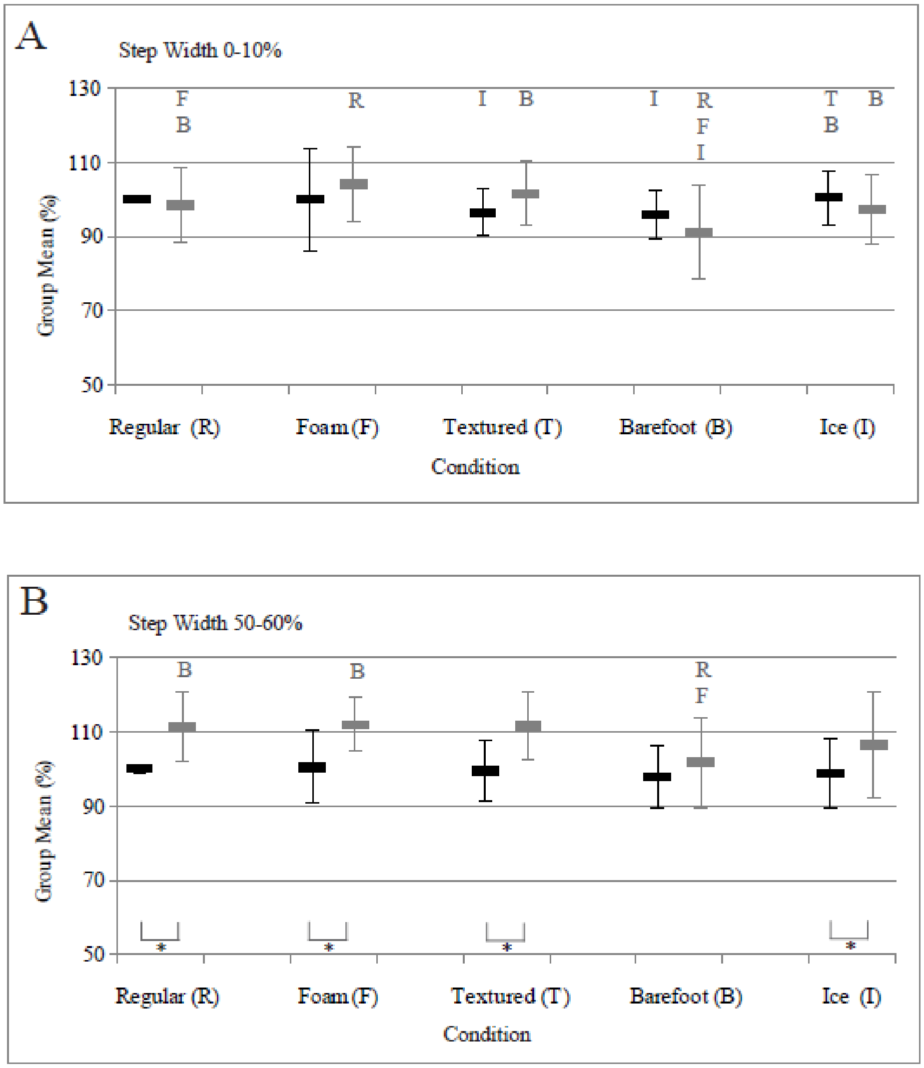

Figure 3.2 Step width at the ankle during first double support phase (A), and second double support phase (B) represented for each footwear condition paired between level (black) and downhill transition (gray). Mean and standard deviation of all 10 participants' average walking trials. Values represented as a percent of each participant's mean for all trials normalized to level walking in regular footwear. Significant differences between level and downhill pairs represented by a star (*) across the bottom ($p \leq 0.05$). Significant differences between footwear conditions represented by the letter of the paired condition (R, F, T, B, I) across the top ($p \leq 0.05$).

Significant differences for TA, SOL, and LG muscle activity were also prevalent during several stride phases across multiple conditions. There was a significant difference between a footwear condition and barefoot or ice condition in at least one stride phase for both level and barefoot walking in all muscles. To note, during the initial swing phase of level walking, TA activity was 8% greater during the regular footwear condition as compared to the textured insole condition ($p < 0.05$). Additionally, during the pre-swing phase of level walking, TA activity of the foam insole condition was 30% greater than the textured insole condition and 43% greater than the barefoot condition (Table 3.1). The downhill transition stride resulted in greater muscle activity during the pre-swing phase of the TA and the terminal swing phase of the LG during all footwear conditions ($p < 0.05$, Figure 3.3). The TA activity during pre-swing was 69% greater during the downhill transition stride in regular footwear and the difference became increasing larger in other footwear conditions (Table 3.1). The downhill transition stride was more than twice the value of level walking during the ice condition. Similar trends were evident in the LG activity during terminal swing as the downhill transition stride was 40% greater during the regular footwear condition and up to 118% greater during the barefoot condition. In contrast, the downhill transition stride resulted in less muscle activity during the terminal stance phase of both the SOL and LG for all footwear conditions as well as the terminal swing phase of the TA during barefoot and iced conditions ($p < 0.05$, Figure 3.3). Lateral gastrocnemius muscle activity during terminal stance was approximately 50% less during the downhill transition stride as compared to level walking for all footwear types. The SOL decrease in activity was nearly as consistent but averaged less than a 40% decrease during the downhill transition stride for all footwear conditions.

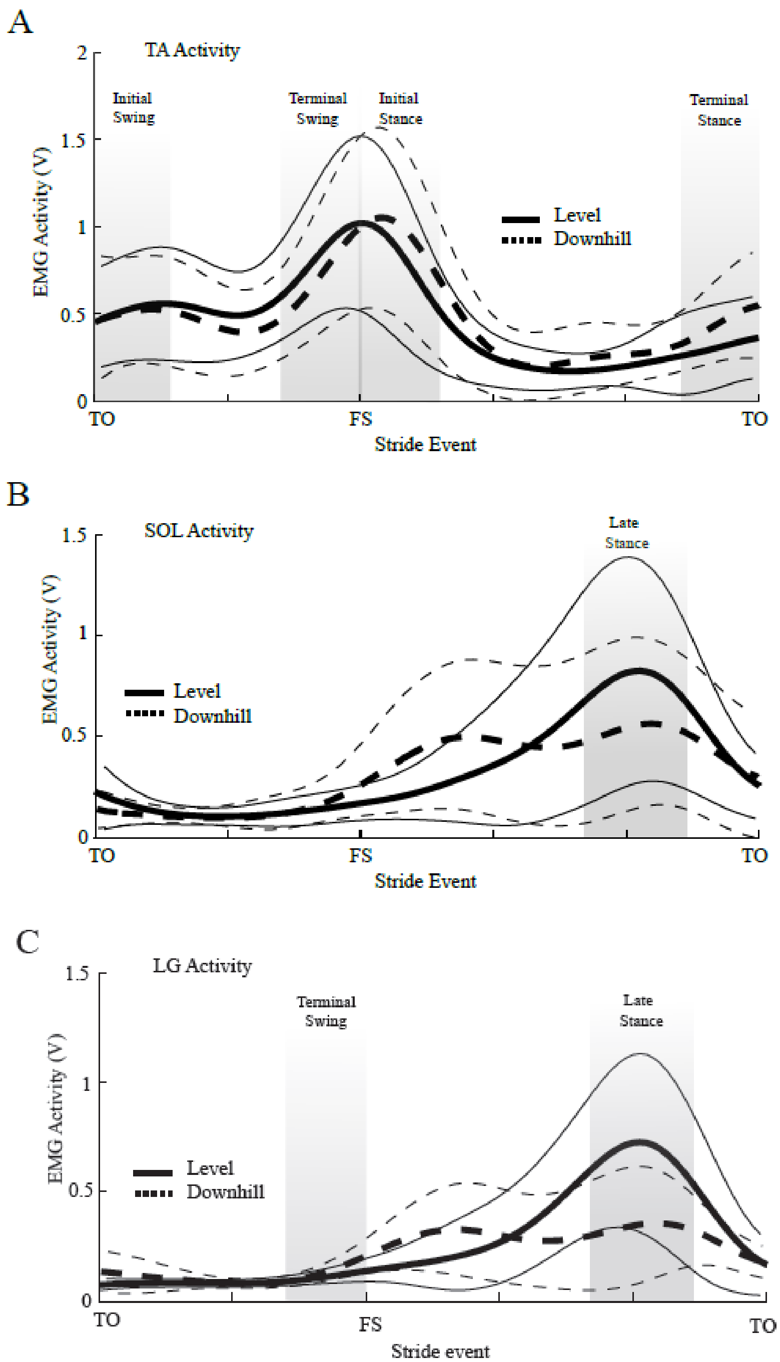

Figure 3.3 Muscle activity for the tibialis anterior (TA) (A), soleus (SOL) (B), and lateral gastrocnemius (LG) (C) represented for regular footwear condition only. Mean (bold lines) and standard deviation (thin lines) for all 10 participants' average walking trials during level strides (solid lines) and downhill transition strides (dashed line). Stride is presented from left toe off to left toe off (TO). The highlighted bands of each stride for each muscle are further analyzed during all footwear conditions.

Table 3.1 Muscle activity for the TA, SOL, and LG during level strides (A) and downhill transition strides (B). Mean values for all 10 participants represented as a percent of each participant's mean for all trials normalized to level walking in regular footwear. Significant difference between level and downhill pairs represented by shaded regions and significant differences between footwear conditions represented by the letter of the paired condition (R, F, T, B I) within each stride condition ($p \leq 0.05$).

Muscle	Condition / Phase	Level Stride: Regular	Foam	Textured	Barefoot	Ice
TA	Initial Swing	100.0 T	103.0	82.2 R I	100.1	116.7 T
TA	Terminal Swing	100.0	94.0	99.8	90.7	91.6
TA	Initial Stance	100.0 B I	89.7 I	93.4 B I	69.1 R T	63.9 R F T
TA	Terminal Stance	100.0	128.1 T B	97.9 F	84.6 F	100.4
SL	Late Stance	100.0 I	109.3 B I	106.2 B I	94.0 F T I	84.1 R F T
LG	Terminal Swing	100.0	105.0	99.8	102.4	128.0
LG	Late Stance	100.0 B	107.4 B I	101.6 B	81.4 R F T	91.1 F

Muscle	Condition / Phase	Downhill Transition Stride: Regular	Foam	Textured	Barefoot	Ice
TA	Initial Swing	103.6	104.3	99.1 B I	115.9 T	129.6 T
TA	Terminal Swing	102.2 B I	96.2 B	100.7 B I	70.7 R F T	68.0 R T
TA	Initial Stance	100.1 B I	104.0 B I	107.6 B I	67.0 R F T	70.3 R F T
TA	Terminal Stance	169.0	175.0	189.0	211.7	275.7
SL	Late Stance	64.6	65.4	68.6 B I	58.3 T	55.0 T
LG	Terminal Swing	140.8	151.1	131.1 I	219.9	180.2 T
LG	Late Stance	53.1 B I	50.9 B I	53.7 B I	44.2 R F T	43.0 R F T

3.4 Discussion

Our hypothesis that compared to foam insoles, textured insoles would result in increased speed and step length and decreased stance time, step width, and lower leg muscle activity was not confirmed. However, it is evident that walking in shoes compared to walking barefoot resulted in changes in gait patterns which could affect the safety of a specific walking task. To a lesser extent, it is also evident that modifying the type of footwear worn can lead to similar gait modifications. Due in part to the variability between participants, it is less clear if footwear modifications will result in consistent changes for all populations, or if increased or decreased cutaneous feedback was the cause of the observed gait changes.

While there were no significant differences between textured and foam insoles or barefoot and iced conditions for speed, step length, and stance time, several measures followed our hypothesis. For both speed and step length, mean values for textured insoles were greater than foam insoles and barefoot insole mean values were greater than the iced conditions. However, speed, step length, and stance time were greater during the regular footwear condition as compared to the barefoot condition. Contrary to results presented by Perry et al. (Perry et al., 2008), this suggests that footwear insoles did affect participants' speed and step length during both level and hill walking. Unlike the footwear conditions tested in our research, the insoles tested by Perry et al. (2008) did not alter sensation over the whole plantar surface of the foot and therefore likely did not provide enough feedback to result in altered speeds and step lengths.

Step width during level walking was significantly narrower during both textured and barefoot conditions as compared to the ice condition during level walking. These results are consistent with findings by Perry et al. (2008) who determined that ridged insoles improved medial-lateral stability as compared to non-textured insoles. We found no significant differences between textured insoles and regular footwear, but results from Perry et al. may be more accurate as they evaluated stability during single support by measuring the minimum distance from the center of mass to the lateral border of the foot. Palluel et al. (2008) also found significant improvements in medial-lateral stability while wearing spiked textured insoles in both young adults (age 21-32) and older adults (age 62-80). These results were measured under reduced vision conditions which may have increased participants' reliance on cutaneous feedback. However, Wilson et al. (Wilson et al., 2008) determined that there was no difference between eyes open and eyes closed conditions when measuring postural stability or step width during level walking.

In addition to level walking, our study also examined the first two steps onto a downhill surface. During this transition stride, wider step widths improve medial lateral stability and may in fact be a desired change to improve medial-lateral stability. If so, it would be advantageous to have a wider step width during both the first double support phase, when one foot is on each the level and downhill surface, as well as during the second double support phase, when both feet are on the downhill surface. Yet, the textured insole condition was the only condition during the first step width measure where the downhill condition was substantially wider than the level condition ($p = 0.054$). Following heel strike, foot rotation could occur to increase base of support measured by step width at the ankle. Therefore, it follows that textured insoles provided additional feedback that improved the response to a terrain change.

It is important to determine if these improvements in medial-lateral stability are temporary changes in response due to comfort affected by footwear modifications or a more permanent response due to changes in cutaneous feedback. Wilson et al. (2008) concluded that over a four week period there was no difference between regular insoles and any of the three different textured insoles. In contrast, our results showed some significant differences in step width between footwear conditions. The study by Wilson et al. (2008) did not compare footwear conditions within subjects, but rather across clinical groups. It is possible that a large variance in step width between participants resulted in a lack of significant difference between footwear conditions. Each of the participants in our study walked in five different footwear conditions within a three hour time period. This could have resulted in differences due to drastic changes in footwear comfort that is specific to each participant's sensitivity and preference as opposed to changes in cutaneous feedback.

Activity of the TA during stance phase has previously been shown to decrease during an iced condition as compared to a barefoot condition during level walking (Eils et al., 2004). Our results did not indicate a difference in muscle activity during early stance in the iced condition as compared to the barefoot condition during downhill transition strides or level strides. There was a significant decrease between ice condition and all three footwear conditions during the loading response but there was a similar decrease during the barefoot condition as well. Similar to findings by Eils et al. (2004) and Nurse and Nigg (2001), our results show that increased TA muscle activity occurs in the iced condition during initial swing resulting in additional toe clearance. Although not significant, TA activity during pre-swing and initial swing was greater during the iced condition as compared to the barefoot condition during both level and downhill walking. Furthermore, during initial swing TA activity was lowest during the textured insole condition and highest during the iced condition during both level and downhill walking. Decreased feedback during the iced condition caused the participants to overcompensate toe clearance whereas during the textured insole condition a more efficient toe clearance distance was selected which minimized TA activity during the initial swing phase.

Additionally, during the downhill transition, SOL activity was significantly greater during the textured insole condition as compared to the barefoot and ice conditions. This is likely due to increased speeds during the downhill textured condition as compared to both barefoot and iced conditions. Moreover, the textured insole condition resulted in the lowest LG activity during terminal swing for both level and downhill strides and was significantly less than the iced condition during the downhill transition stride. Therefore, the textured insole improved anterior-posterior stability as less muscle activity was required for foot placement during the downhill transition stride.

The difference between a footwear condition and the regular footwear condition was as much as twice as large during the downhill transition condition as compared to the level condition for speed, step length, and step width, and in some cases EMG activity differed by more than a factor of four. However, step length was the only instance where there was a significant difference between barefoot and ice conditions during the downhill transition condition and not the level condition. Muscle activity of the TA during pre-swing was also significantly different between textured and foam insoles during the level condition and not the downhill condition. The hill transition did increase stability requirements, but there were not significant changes that became evident during the hill transition stride that were not observed during the level stride. It is possible that visual and vestibular feedback is prioritized during a hill transition walking task and that cutaneous feedback provides cues for minor balance adjustments.

Despite the varied response to footwear types, there were consistent results between ramp conditions which confirm those found by Gottschall et al. (submitted). Speed during the level condition was greater than the downhill transition for all footwear conditions. Although an increase in propulsive forces during the downhill condition, speed remained slower as the participants approached and crossed the transition between the level and downhill surface.

Additionally, step length was smaller during the downhill transition as compared to level walking during all footwear conditions. Again, despite navigating a terrain change that may require obstacle clearance and therefore a larger step length, participants selected a slower, smaller, and more stable step during the hill transition. Finally, in contrast to Kawamura et al. (1991), step width during the second double support was greater during the downhill transition as compared to level walking for all footwear conditions except barefoot walking. Kawamura et al. (1991) utilized force plates located in the ground under the ramp apparatus to measure step width. This was likely less accurate than the kinematic ankle marker location used in our studies. EMG data during ramp walking also confirmed previous research. Similar to work by Lay et al. (2007), we found that muscle activity of the SOL and LG was reduced during terminal stance of the downhill transition as compared to level walking. This is likely due to the effect of gravity during downhill walking which results in less propulsive forces from the LG and SOL required during push off to maintain a given speed. Additionally, LG activity during terminal swing has been shown to aid in stability during foot placement (Winter, 1983) and activity during that phase was significantly greater during the downhill transition stride for all footwear conditions.

It is difficult to directly measure human sensory systems to determine if changes in gait in response to various insoles are in fact due to a change in afferent feedback (Nurse et al., 2005). However, a common difference when comparing a barefoot to ice condition and a textured to foam insole condition is the amount of cutaneous sensation at the plantar surface of the foot. Therefore, if the results between barefoot and ice conditions mirrored those of textured and foam insole conditions, the theory that textured insoles increase cutaneous feedback could be further supported. Unfortunately, our results showed relatively few significant differences between barefoot and ice conditions or textured and foam insole conditions and the pairs never resulted in significant differences within the same variable. As a result, it is difficult to make comparisons between the two cases. Yet, there are some positive results that would support further research to investigate the similarities between the two cases.

Footwear fit, shape, and hardness have been shown to alter walking stability (Perry et al., 2007; Menant et al., 2008), rate of injury (Mundermann et al., 2001), and perceived comfort (Mundermann et al., 2002). The purpose of our study was to determine if a textured insole that provided comfortable cushioning could also improve walking stability by increasing cutaneous feedback. Although our data shows that footwear conditions alter gait patterns and lower leg muscle activity during level walking and a more difficult hill walking task, there is not enough evidence to support the hypothesis that textured insoles will increase stability as compared to other footwear conditions.

Chapter 4

Child Temporal-Spatial Gait Characteristics and Variability During Uphill and Downhill Walking

4.1 Introduction

Gait development is a gradual process in children since muscle and bone growth, as well as motor skill improvements are long-term developments. Past research has investigated various child age groups ranging from 1 to 15 years old in an attempt to determine temporal-spatial gait parameters that characterize gait development. Stride speed, stride length, step width, and percent stance time are basic variables typically used to characterize maturity of child gate. These gait parameters are dependent upon multiple factors including body size and walking experience (Kingsnorth and Schmuckler, 2000) which are both correlated to age (Norlin et al., 1981). As children mature, speed and stride length increase while step width and percent stance time decrease (Rose-Jacobs, 1983; Sutherland, 1997; Dusing and Thorpe, 2007).

Speed is an informative temporal-spatial gait characteristic as it is a product of both stride length and stride frequency. Walking faster or slower than preferred speed can alter several other gait characteristics (Kang and Dingwell, 2008). Sutherland (1997) found that walking velocity rapidly increased until age 3.5 - 4 at which point a more gradual increase continued. In agreement with results found by Sutherland (1997), a more recent study examining children between the ages of 1 and 10 years old walking at a self-selected speed found that when speed and stride length are normalized to leg length, a gradual plateau occurs beginning around the age of 4 years (Dusing and Thorpe, 2007). This maturation pattern can be attributed to the importance of gains in motor control at younger ages followed by leg length and body height growth at older ages (Sutherland, 1997).

However, Beck et al. (1981) concluded that, when normalized to height, stride length was not significantly different between any age group (2 through 15 years old) and was on average 76% of the participant's body height while walking at 1.04 m/s. The controlled walking speed used in this study may explain this difference in results. Independent of leg length and age, step length is modified to maintain anterior-posterior stability during walking (Winter and Eng, 1995) and therefore may be a sign of mature gait in children (Kingsnorth and Schmuckler, 2000). Although it is evident that speed and step length are important markers for gait development across age groups, it is difficult to use these measures alone to quantify gait maturity since long term skeletal growth will continue to alter these parameters (Sutherland, 1997).

Step width may also be an appropriate measure of walking maturity as it is a primary determinant for medial lateral stability (Maki, 1997). Dusing and Thorpe (2006) reported that step width decreased in children between 1 and 3 years old and then remained relatively constant after 4 years old. These results were consistent with those found earlier by Rose-Jacobs (1983) who reported that step width was not different between 3 and 5 year olds but step width variability did change between age 3 and 5.

Increased time in single stance, or decreased percent of gait cycle in stance, indicates increased stability since greater balance control is required during single limb support (Sutherland, 1997). Sutherland (1997) reported that by age 3.5 – 4, the percent of single limb support is approximately equal to that of normal healthy adults. Similarly, Beck et al. (1981) found that above the age of 2, there is no difference in swing or stance percent at the same walking speed. Additionally, the variability of percent time in double support, a comparable measure to single stance time, decreased in children over the age of 4 (Dusing and Thorpe, 2007).

Variability of gait characteristics is also an important measure of both gait maturity and stability (Hausdorff et al., 1999). Adjustments of temporal-spatial gait parameters to increase medial-lateral and anterior-posterior stability can be attributed to both fear of falling and fall risk (Maki, 1997). Speed, stance time, stride length, stride width mean, and stride-to-stride variability were reported to correlate well to self-reported fear of falling as well as future fall frequency. Changes in mean values correlated to reports of fear of falling whereas both stride length and stride width variability were accurate predictors of future falls (Maki, 1997). In a similar study of older adults, Hausdorff (2001) found that stride time and swing time variability were also strong predictors of future falls. Hausdorff (1991) also studied gait variability in children between 3 and 14 years old. Stride time variability for children ages 11 - 14 was similar to that of healthy adults but increased significantly for 6 to 7 year olds and again for 3 to 4 year olds. Since stride-to-stride variability predicts both fall risk and child age, it follows that walking patterns of children between 3 and 7 are not fully matured despite the appearance of normal mean temporal-spatial gait parameters.

In addition to growth, walking experience can also affect children's level of gait maturity. For instance, Gill et al. (2009) describes how walking infants approach walking challenges differently based on their level of walking experience. Infants, 15-16 months old with 3-4 months of walking experience, selected smaller stride lengths when walking down slopes and decreased stride lengths as the slope became steeper. Unfortunately, there is little research available studying children and slope walking to compare these results to those for children above 3 years old.

Similar to results presented by Gill et al. (2009) it has been observed that adults decrease stride speed and stride length during downhill walking (Kawamura et al., 1991) and that this decrease becomes more pronounced among older adults as the slope increases (Sun et al., 1996). During uphill walking, speed decreased and stance time increased (Kawamura et al., 1991). It has also been shown that step length variability and step width increase during uphill and downhill walking (Gottschall et al., submitted). These differences during hill walking represent both proactive and reactive gait changes which aid in the control of medial-lateral and anterior-posterior stability. Currently, it is not clear if children possess the same ability to control stability during hill walking.

Therefore, the purpose of this study is to investigate the similarities and differences between walking strategies utilized by children during hill walking and that of adults. Further identifying these differences between children and young adults could provide more details to identify motor learning pathways for children and adults with gait abnormalities. Children ages 3.5 - 5.5 were studied as that age group represents children who have reached stride length, step width, and percent stance time measures which are comparable to adults during level, self-selected velocity walking. We hypothesized that the mean percent change between level and uphill or downhill walking for speed, stride length, step width, and percent stance time would be greater in adults as compared to children. Additionally, we hypothesized that variability of these temporal-spatial gait parameters will be greater in children during uphill and downhill walking as compared to level walking. This variability would be greater in children than in adults during all walking conditions.

4.2 Methods

Participants

Thirty healthy children, 17 boys and 13 girls, completed the protocol (Table 4.1). All of the participants' parents or guardians gave written informed consent that followed the guidelines of The Pennsylvania State University Human Research Committee. All participants were in good general health with no known gait or neurological disorders and were currently enrolled in university managed day care facilities.

Table 4.1 Descriptive statistics of participants for 6 month age groups and the total child group from 3.5 to 5.5 years old. Values represented as mean and (standard deviation).

Age (yrs)	n	Age (days)	Height (cm)	Weight (kg)	Leg Length (cm)
3.5 - 4	6	1369 (55.41)	102.47 (2.71)	16.88 (1.60)	51.36 (2.09)
4 - 4.5	9	1536.78 (36.20)	105.86 (3.90)	17.77 (1.89)	54.51 (2.68)
4.5 - 5	7	1713.14 (57.82)	106.49 (3.18)	17.59 (1.01)	55.03 (3.48)
5 - 5.5	8	1922.50 (60.02)	110.24 (2.18)	18.24 (1.31)	58.31 (2.64)
Total (3.5 - 5.5)	30	1647.23 (210.25)	106.50 (4.01)	17.67 (1.51)	55.01 (3.57)

Protocol

Participants completed five level, uphill, and downhill walking conditions. All of the walking conditions were completed at a self-selected velocity over a 425 cm GAITRite Electronic Walkway (Version 3.9, CIR Systems Inc, Havertown, PA). The hill conditions utilized a portable ramp composed of a 125 cm 15 degree incline with a 125 cm plateau. Participants were permitted to complete walking trials in self selected footwear or barefoot to ensure clean foot strikes as well as participant comfort. Participants were instructed to walk at a comfortable walking pace similar to the speed and manner they used to walk through the halls of the daycare. When necessary, verbal encouragement was given to participants to walk the full length of the electronic walkway along a straight path. Each walking trial was monitored by three researchers who visually determined the validity of the trial. Trials which involved running, jumping, or shuffling and those in which the participants did not remain focused on the walking task were omitted from the five required trials for each condition.

Temporal-Spatial Parameters

The GAITRite electronic walkway was used to record participant footfalls via 1.27 cm square piezoelectric cells built into the 61 cm by 427 cm active collection space at a 60 Hz sampling rate. Multiple studies have been conducted which have validated the use of a GAITRite electronic walkway to record temporal-spatial parameters in children (Thorpe et al., 2005; Dusing and Thorpe, 2007) and adults (Bilney et al., 2003; Menz et al., 2004). A purpose written Matlab program (Version R2006b, The Mathworks, Natick, MA) was written which utilized GAITRite determined footfall time and location to calculate stride speed, stride length, step width, and stance time percent. A stride was defined as toe off to toe off (-40% to 60%). For each trial, we selected the full stride which took place at the midpoint of the level or hill surface to ensure that gait initiation or termination was not captured. Stride speed was defined as the difference in anterior-posterior location of first toe off to second toe off (stride length) divided by the time between the two events. Step width was defined as the absolute value of the difference in medial-lateral heel location between the left and right foot during the first double support phase. Stance time percent was defined as the time from heel strike to toe off (stance time) divided by the total stride time and presented as a percent. Mean values and trial to trial deviation (coefficient of variation) for each walking condition were calculated for each variable. We then normalized each variable to the participant's mean level walking value and reported hill walking as a percent change from level walking.

Variability of temporal-spatial gait parameters and percent change of mean values between level and hill walking for children were compared to data calculated for thirty healthy young adults (age = 21.67 (1.40) yr, height = 175.01 (9.79) cm, weight = 75.11 (16.08) kg, leg length = 97.92 (8.14) cm, mean (standard deviation)) (Gottschall et al., submitted). Each of the adult participants completed five level walking trials and five uphill and downhill walking trials. All walking trials were completed at a self-selected walking velocity over a 25m walkway. Uphill and downhill walking was completed utilizing a portable 15 degree continuous incline 2.4m ramp with a 4.8m plateau. We recorded temporal-spatial gait parameters utilizing a six-camera, passive marker, 3D photogrametry system (Motion Analysis Corporation, Santa Rosa, CA) with retroflective markers placed over the sacral crest, lateral malleoli, and over the shoes of each participant superficial to the posterior calcaneus and distal head of the first metatarsal. We calculated stride speed, stride length, step width, and stance time for each stride. A stride was defined as left toe off to left toe off occurring completely on the level or hill surface (Gottschall et al., submitted).

Statistical Analysis

All data were analyzed across age groups and across level and hill conditions using a repeated measures design (ANOVA). Differences between child and young adult groups were further analyzed using Newman-Keuls post hoc tests for mean values, variability, and percent change from level to hill strides. Significance was defined as $p \leq 0.05$.

4.3 Results

Similar to our hypothesis, the mean percent change between level and hill surfaces was not consistently greater in the child group as compared to the adult group. In further support of our hypothesis, variability was higher in the child group than the adult group and there was an increase in variability within the child group during hill walking as compared to level walking. There were no significant differences in temporal-spatial gait parameters between six month age groups during level, uphill, or downhill walking (Table 4.2). Additionally, during level walking, all temporal-spatial gait parameters for the child group (3.5 - 5.5) were significantly different than the adult group ($p < 0.001$). Compared to the adult group, the mean child group speed was 0.48

m/s slower, stride length was 72 cm shorter, step width was 4 cm narrower, and stance time percent was 3.2 percent shorter. Values of speed and stride length for level walking were also normalized to leg length of each participant and again found to be significantly different between the child and adult groups. However, here the child group walked faster as a percent of leg length (1.65%) compared to the adult group (1.38%) but with shorter strides (133.56% compared to 146.87%).

Table 4.2 Mean and (standard deviation) for stride speed (m/s), stride length (cm), step width (cm), and stance percent (%). Values are represented for children, in 6 month age subdivisions, during level, uphill, and downhill walking. Significant differences within hill conditions and across 6 month age groups are represented as with an asterisk (*). Significant differences between the child age group (3.5 - 5.5) and the adult age group (Gottschall et al., submitted) during level walking are represented as ***bold, italicized*** values ($p < 0.05$).

Condition		3.5 - 4	4 - 4.5	4.5 - 5	5 - 5.5	3.5 - 5.5
Stride Speed (m/s)	**Level**	1.03 (0.19)	0.93 (0.13)	0.83 (0.19)	0.84 (0.18)	***0.90 (0.18)***
	Uphill	0.96 (0.27)	0.89 (0.24)	0.76 (0.24)	0.70 (0.22)	0.82 (0.25)
	Downhill	1.07 (0.11)	0.88 (0.41)	1.02 (0.32)	0.85 (0.42)	0.95 (0.34)
Stride Length (cm)	**Level**	79 (8)	75 (13)	70 (11)	68 (11)	***73 (11)***
	Uphill	76 (9)	69 (9)	68 (12)	66 (16)	69 (12)
	Downhill	76 (7)	64 (12)	71 (14)	61 (20)	67 (15)
Step Width (cm)	**Level**	6 (3)	7 (2)	7 (2)	6 (1)	***6 (2)***
	Uphill	6 (2)	8 (3)	8 (3)	8 (2)	7 (2)
	Downhill	5 (1)	6 (2)	7 (3)	6 (2)	6 (2)
Stance Percent (%)	**Level**	58.1 (2.6)	60.4 (1.7)	60.4 (2.9)	60.3 (1.4)	***59.9 (2.3)***
	Uphill	59.3 (5.3)	60.7 (5.9)	60.7 (5.9)	64.6 (2.8)	62.1 (4.7)
	Downhill	55.5 (4.7)	55.4 (6.0)	55.4 (6.0)	56.2 (6.4)	55.3 (5.2)

Although mean values differed during level walking between the child and adult age groups, the percent change from level to hill walking was rarely different between child and adult groups (Table 4.3). The only significant difference between the two groups was the change in percent stance time during downhill walking. Compared to level walking there was a greater reduction in stance time percent for the child group versus the adult group ($p < 0.001$). There were, however, several significant differences between level and hill walking within age groups. For the child age group, step width and percent stance time increased during uphill walking ($p < 0.05$) and percent stance time decreased during downhill walking ($p < 0.001$). For the adult age group, speed decreased during uphill walking, stride length decreased during downhill walking, and stance time percent increased during uphill walking ($p < 0.05$).

Table 4.3 Mean values of temporal-spatial gait parameters for children and adults (Gottschall et al., submitted) represented as a percent change from level walking. Significant differences within age groups and between level and uphill or level and downhill walking are represented with an asterisk (*). Significant differences within hill condition and between age groups are represented as ***bold, italicized*** values ($p < 0.05$).

Condition	Child Group		Adult Group	
	Uphill	Downhill	Uphill	Downhill
Stride Speed	91.6	105.8	90.0 *	95.5
Stride Length	95.4	89.8	99.8	94.5 *
Step Width	120.6 *	95.2	99.2	105.6
Stance Percent	103.7 *	***92.6*** *	102.8 *	***98.6***

Furthermore, during hill walking, measures of temporal-spatial gait parameter variability for the child age group increased in several instances. As compared to level walking, downhill walking resulted in significantly greater variability in speed, stride length, and percent stance time ($p < 0.05$, Table 4.4). Compared to level walking, there were no significant differences in step width variability during downhill walking, and no significant differences in variability for speed, stride length, step width, or percent stance time for uphill walking.

Compared to the adult group, all measures of variability were greater in the child group ($p < 0.01$, Table 4.4). Speed, stride length, step width, and percent stance time variability were greater within the child group during level, uphill and downhill walking. Hill walking increased the difference between age groups with the largest difference for all variables occurring during downhill walking.

Table 4.4 Coefficient of variation for temporal-spatial gait parameters for children and adults (Gottschall et al., submitted) during level, uphill, and downhill walking. Significant differences within age groups and between level and uphill or level and downhill walking are represented with an asterisk (*). Significant differences within hill condition and between age groups are represented as ***bold, italicized*** values ($p < 0.05$).

Condition	Child Group			Adult Group		
	Level	Uphill	Downhill	Level	Uphill	Downhill
Stride Speed	***0.20***	***0.24***	***0.29****	***0.03***	***0.03***	***0.03***
Stride Length	***0.10***	***0.11***	***0.17****	***0.04***	***0.02***	***0.02***
Step Width	***0.49***	***0.47***	***0.53***	***0.29***	***0.19***	***0.19***
Stance Percent	***0.05***	***0.07***	***0.10****	***0.02***	***0.02***	***0.03***

4.4 Discussion

In summary, children walked slower with a smaller stride length, step width, and percent stance time as compared to adults during level, uphill, and downhill walking. In support of our hypothesis, child temporal-spatial gait variability increased during hill walking and was greater than adults during all conditions. The hypothesis that the adult group would have greater percent change in temporal-spatial parameters between level and hill walking was not confirmed since children often adjusted these parameters in different magnitude and direction compared to adults.

Temporal-spatial gait parameters of children during level walking are consistent with those presented in previous literature. We found no significant differences between 6 month age groups during level or hill walking and the child population was therefore considered as one group. This is consistent with data showing that increases in walking speed, stride length, and step width slowed after the age of 4 (Sutherland, 1997; Dusing and Thorpe, 2007) and that no differences in single stance time were present beyond the age of 2 (Beck et al., 1981). Moreover, the difference in mean temporal-spatial gait parameters during level walking between our child and adult populations are consistent with previous data which documented that children walk slower and take shorter and narrower steps (Rose-Jacobs, 1983; Sutherland, 1997; Dusing and Thorpe, 2007). However, percent stance time results differ from those of Sutherland (1997) and Beck et al (1981) who both state that by age 3.5, percent of single limb support time is equivalent to that of healthy adults. This may be explained by the difference in methods as our data represents total percent stance time as opposed to percent of single limb support.

Unlike level walking, there is little research available describing normative child gait during hill walking. Gill et al. (2009) described a walking pattern for infants represented by smaller strides during the approach and traverse of a downhill slope (Gill et al., 2009). Although the current study examines an older child age group, stride length was found to be shorter during both downhill and uphill walking but the percent change was not statistically significant. To add, there were few significant differences of percent change from level to hill walking between child and adult groups. This could be a result of differences in strategies used by each group. The adult group walked slower during both uphill and downhill conditions whereas the child group walked slower during the uphill condition but faster during the downhill condition. Future studies utilizing electromyography and muscle strength evaluations could determine if muscle power and the production of propulsive and braking forces are limiting factors during uphill and downhill walking among children.

In addition to differences during level and hill walking, strategies used for obstacle clearance have been shown to differ between young children, old children, and adults (Kingsnorth and Schmuckler, 2000; McFadyen et al., 2001; Vallis and McFadyen, 2005). McFadyen et al. (2001) reported that although children age 7 to 9 years old possess anticipatory obstacle clearance methods similar to those of adults, the gait patterns utilized do not match that of a healthy adult. Vallis and McFadyen (2005) showed that during obstacle avoidance, children age 8 – 12, altered temporal-spatial gait parameters differently than adults when stepping around an obstacle. During steps before the obstacle, children decreased step length whereas adults did not. Similarly, children in the current study decreased step length by a greater percent than adults during both uphill and downhill walking (p = 0.09 and 0.27 respectively). Furthermore, although step width changed equally for both children and adults during obstacle avoidance, step width variability in the child group was greater (Vallis and McFadyen, 2005). In the current study, the child group coefficient of variation for step width was greatest during downhill walking (p = 0.43) and was significantly greater than that of the adult group during level, uphill, and downhill walking. Michel et al. (2010) also showed that children age 6-8 adapted differently to obstacle avoidance than children age 9-12 and that the older children were able to clear more obstacles. It is evident that the young children in our study do not possess a fully mature gait profile. Lastly, Kingsnorth and Schmuckler (2000) reported that for infants (15 - 30 months) walking experience, as defined by number of months spent walking, correlated closely to maximum height of obstacle cleared. However, although stride length and step width differed for age, they were not appropriate predictors of obstacle height clearance. These results further the theory that gait maturity is related to a combination of growth, motor skill, and motor learning which may be important factors for both obstacle clearance and hill walking.

` Variability may be an appropriate measure of motor skill and learning. Hausdorff (1991) found increases in gait variability among young children, reporting that stance time variability was higher in children age 3 - 4 compared to children age 6 - 7 and 11 - 14. Percent stance time, stride length, and stride speed variability was not only greater in children than in adults, but variability was also higher during downhill walking as compared to level walking within the child group. Since variability of these temporal-spatial gait parameters correlate to fall risk (Maki, 1997; Hausdorff et al., 2001) it follows that downhill walking increases the risk of falls among a child population that is already experiencing greater fall risk as compared to healthy adults during level walking.

To note, in a study of healthy adults, Danion et al. (2003) found that during treadmill walking, the coefficient of variance for step length and step frequency during a given speed rarely exceeded 3%. In agreement with these results, the adult group in our study did not exceed a 4% coefficient of variance during overground or hill walking for stride speed, stride length, and percent stance time. Comparatively, children were consistently above the 3% level, typically between 5 and 20%. Likewise, step width variability for both adults and children was much greater than 3%. This may provide further evidence that step width is a major determinant for medial-lateral stability and the measure of variability may be strong predictor of fall risk (Maki, 1997).

Limitations in the measure of coefficient of variation under the current study should be noted. Previous studies analyzing temporal-spatial gait variability utilized treadmill, extended, or circular walkways to calculate stride-to-stride variability (Hausdorff et al., 2001; Danion et al., 2003; Hausdorff, 2007; Kang and Dingwell, 2008). Our study utilized trial-to-trial variability of independent strides as opposed to consecutive strides within one long walking trial. However, reduced walking duration during each trial was required in order to maintain attention levels and reduce fatigue for the child population. Consequently, the use of trial-to-trial variability also resulted in fewer total number of steps analyzed. Each participant completed a minimum of five trials for each condition, but studies have shown that upwards of 400 steps are required to appropriate analyze kinematic variability (Owings and Grabiner, 2003). Therefore, collecting variability measures over consecutive strides during longer walking periods could result in more accurate and significant comparisons of temporal-spatial gait variability between children and adults during level and hill walking tasks.

Still, it is evident from the current results that children age 3.5 - 5.5 do not yet exhibit a mature gait. Growth rates may limit temporal-spatial gait parameters during many walking tasks but leg length and body height should not limit gait variability or effect gait changes during hill walking. Development, including muscle strength, motor control, and walking experience, are more likely factors limiting walking stability among children.

Chapter 5

Conclusion

Multiple changes in gait patterns during hill walking have been documented which improve upon the results currently available in the literature. First, healthy young adults modify their gait patterns to maintain anterior-posterior and medial-lateral stability during uphill and downhill walking as well as the transitions between hill walking and level walking. Second, changes in temporal-spatial gait parameters and muscle activity patterns during hill walking can be further altered by footwear type which may result in greater stability. Lastly, the differences between level and hill walking observed in healthy young adults are magnified in children. These results aid to further characterize walking profiles during complex tasks such as hill walking.

Anterior-posterior and medial-lateral stability is maintained during hill walking and hill transitions through multiple changes in temporal-spatial gait parameters. Healthy adults manage these changes in order to meet several demands based on the type of hill transitions encountered. These demands include obstacle clearance, propulsion demands, and stability maintenance, and can be prioritized individually or compensated for simultaneously depending upon the specific demands created by the walking task. Specifically, the level to downhill transition poses the greatest challenge to both anterior-posterior and medial-lateral stability. Compared to level walking, speed and step length were significantly less during the L-DN stride and speed, step length, and stance time variability significantly increased, indicating an adaptation in anterior-posterior stability. Additionally, step width at the heel and toe as well as the area of base of support was significantly greater during the L-DN as compared to level walking conditions, indicating an adaptation in medial-lateral stability.

During the level to downhill transition, anterior-posterior and medial-lateral stability adaptations can be improved through the use of proper footwear. There are large differences in muscle activity and walking patterns between barefoot and shoe walking on both a level and hill surface. Furthermore, there is evidence that supports the hypothesis that a textured insole improves anterior-posterior and medial-lateral stability. During the L-DN stride, the textured insoles resulted in reduced muscle activity controlling foot placement and increased medial-lateral stability control during the first double support step width measure. It is possible that a textured foam insole is an appropriate alternative to thick insole padding for individuals who require additional comfort and support and also need to limit fall risk.

Hill walking can be especially challenging for children. Our study first confirmed previous results that during level walking, children walk slower and take shorter and narrower steps as compared to healthy adults. Second, this study found that compared to adults, children walk with greater speed, step length, stance time, and step width variability during level and hill walking. More importantly, during downhill walking, this gait variability is significantly increased as compared to level walking. This increase, which is unique to the child group, indicates that anterior-posterior stability is challenged during downhill walking in an age group that is already experiencing anterior-posterior and medial-lateral stability challenges during level walking.

There are multiple complexities that may have influenced both the results of these studies and the safety of individual's real world walking transitions. First, when navigating hills and hill transitions, multiple challenges must be met. Balance, obstacle clearance, and propulsion demands must all be satisfied to safely navigate changing terrains. However, these criteria may be met at varying levels and also prioritized in different orders. Because certain aspects which may influence an individual's selected walking pattern, such as task goals, perceived risk, and speed-accuracy preference, are difficult to measure, it is not possible at this time to determine which factors contribute the most to walking patterns during hill transitions. In fact, multiple factors may be influencing variables such as speed, step length, and step width simultaneously but with different weights.

Similar affects may have occurred when studying footwear insoles. Each participant likely had varying levels of sensory feedback and different preferences for footwear comfort. This is especially evident by the large between subject variability for each footwear condition. It is likely that fit, comfort, experience, and preference were all factors which influenced the results of the various footwear insoles in addition to the hypothesized changes in sensory feedback. However, long term studies, surveys, and larger sample sizes would be needed to determine which factors may contribute to improved dynamic balance.

Finally, walking experience and risk assessment can also influence gait patterns during hill walking and hill transitions. As Maki (1997) has shown, changes in mean values of speed, step length, and step width correlate strongly to fear of falling. Levels of perceived fear during each walking task were likely to impact the results of all subjects.

However, this factor may be increasingly important when studying older adults who may be injured more severely following a fall. Experience and risk assessment were also likely influential in the child population. Despite having less experience navigating hills, children may have approached the downhill walking task with less caution as they did not prioritize the potential risk of falling.

Future studies should focus to further the understanding of gait during complex walking tasks. I recommend that similar studies involving hill walking be completed with participants from special populations including those with reduced proprioception, neurological disorders, or gait asymmetries. The effectiveness of textured footwear should especially be tested within these special populations. Additionally, further investigating gait variability as a predictor of fall risk and indicator of compromised anterior-posterior and medial-lateral stability will be advantageous. To meet this goal, treadmill walking on various inclines and declines could be utilized to record a larger number of strides for measures of stride to stride variability. This increased knowledge base could lead to an improved understanding of how the central nervous system controls walking, which could eventually result in interventions to improve the mobility of individuals with gait disorders.

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Appendix

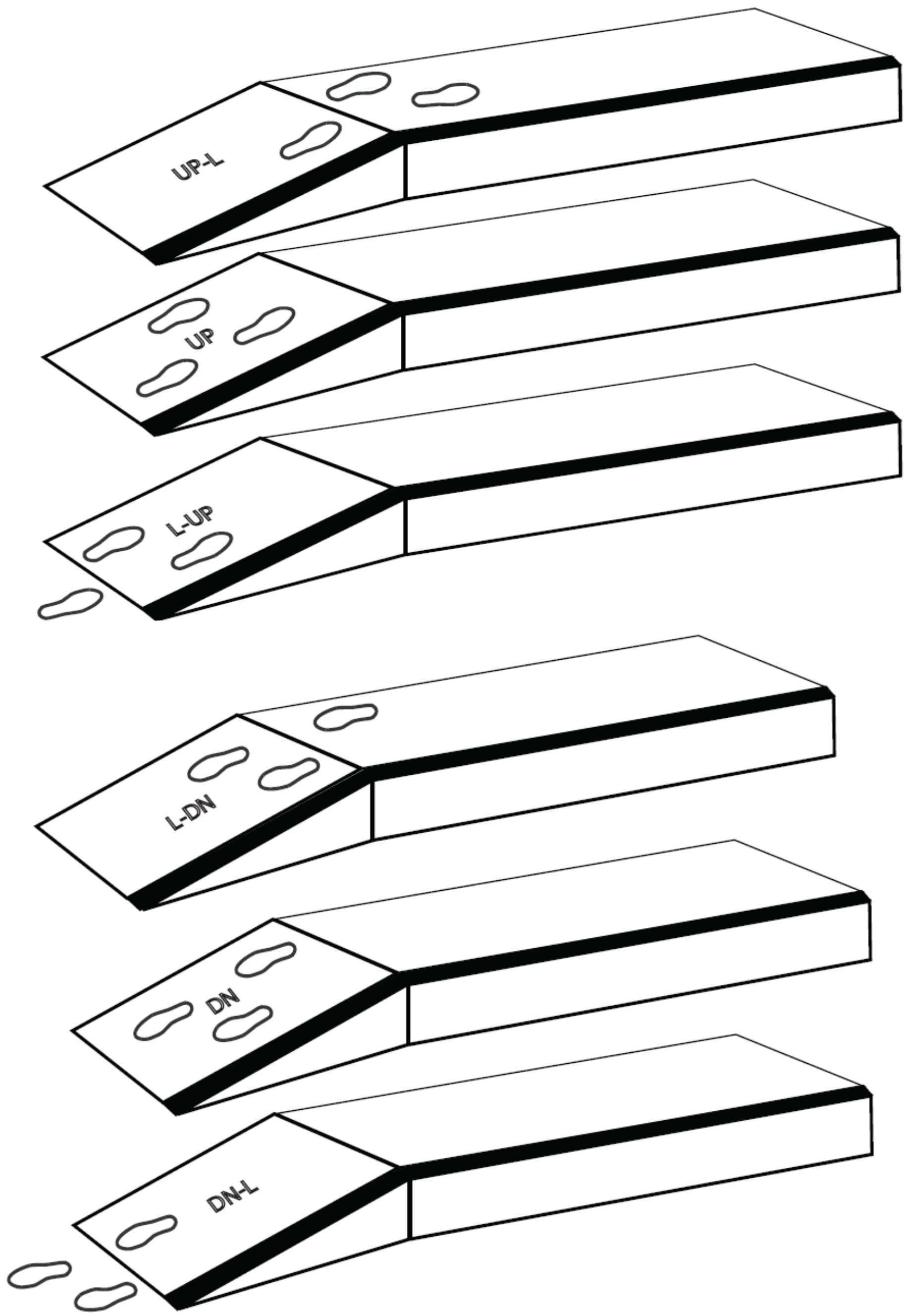

Figure A.1: Transition steps on ramp selected from a larger walking trail extending the length of the walkway and ramp. A single stride was then selected which included the left leg swing phase over the edge of the transition. In this manner, the strides depicted begin at the left toe off of the original surface (not shown) to the left toe off of the left footfall on the new surface (shown).

Variable	Hypothesis	L-UP	UP	UP-L	L-DN	DN	DN-L
Speed	Less		-	-	-	-	+
Step Length	Less	+		+	-	-	+
Stance Time	Greater	+	+	-	-		-
Speed Var.	Greater				+		
Step Length Var.	Greater	+	+	+	+		
Stance Time Var.	Greater				+		
Step Width Heel	Greater	+			+		
Step Width Toe	Greater	-	-		+	+	
Base of Support	Greater	+	+	+	+		-

Figure A.2: Summary of transition step significant results highlighted from previous results in Table 2.1. Each variable is followed by the hypothesized change in direction from level walking values to transition walking. Significant differences for each variable and each condition are marked with a plus (+) when greater than level and minus (-) when less than level walking ($p \leq 0.05$). If the significant difference is in the direction hypothesized it will be colored. Significant differences in the opposite direction hypothesized will remain black. Consistent changes in agreement with the hypothesis are evident as speed is less than level walking and step length variability is greater than level walking during 4 conditions each. The L-DN condition also resulted in the most consistent signficant differences.

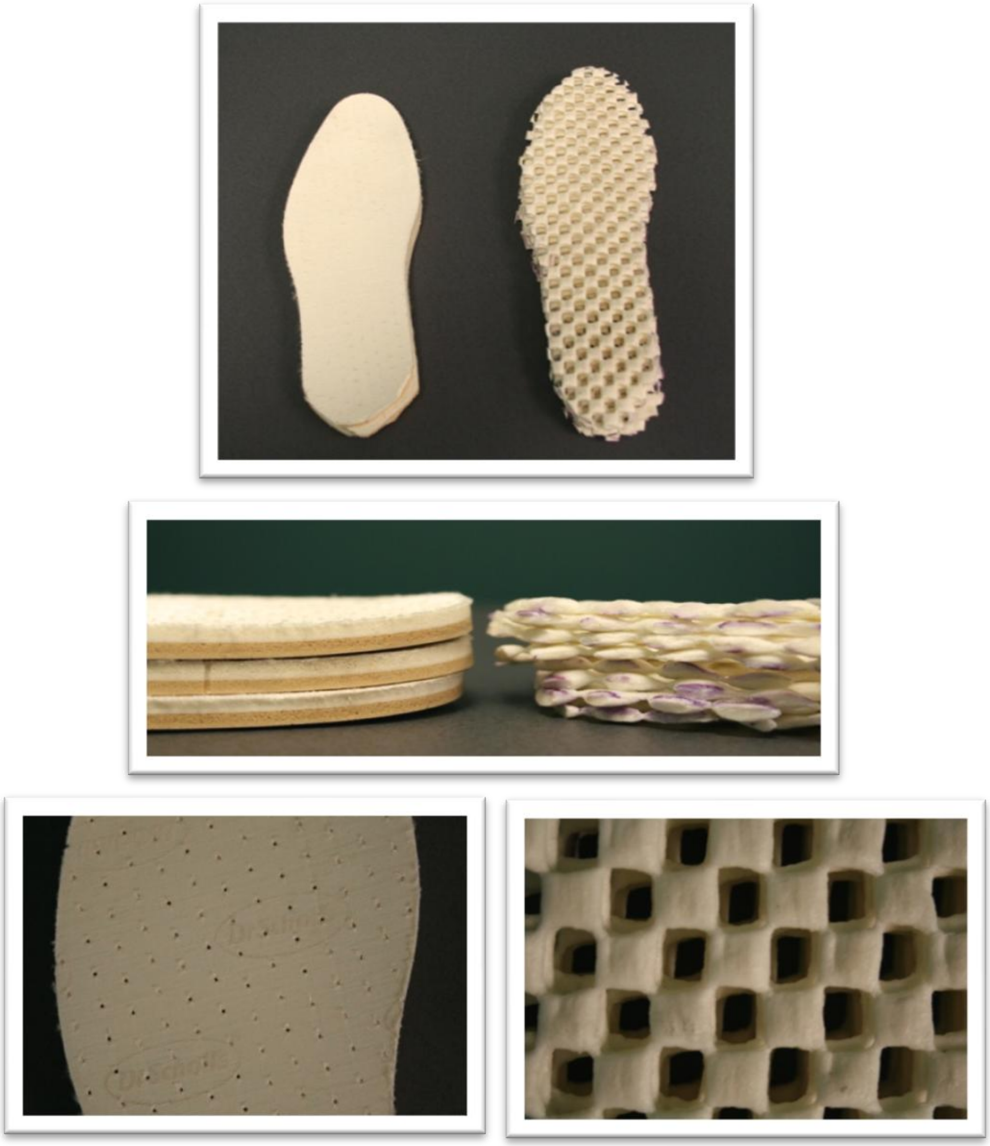

Figure A.3: Footwear insoles. Foam insoles appear on the left hand side and textured insoles on the right. The insoles were constructed out of similar materials to offer comprable thickness, cushion, shape and size. In this manner, the main difference between the two insoles is the surface texture that contacts the plantar surface of the foot.