Biomechanical Perspectives on Human Walking and Slip, Trip and Fall Events: A Literature Review

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Introduction

Although observation of human motion dates back to the time of Aristotle, advances in technology have allowed for great progress in the understanding of gait biomechanics over the last 150 years. Applications of this research include development of injury prevention strategies, design of walking aides, improvement in sporting techniques, and design of footwear. Understanding the basic biomechanics of human walking can assist the safety professional in evaluating an environment for potential safety concerns. This paper is meant to serve as a brief introduction to the aspects of locomotion biomechanics that may be most critical to such evaluations.

Biomechanics of Normal Walking

Walking has been defined as "to go along or move about on foot at a moderate pace; specif., a) to move by placing one foot firmly before lifting the other, as two legged creatures do (Merriam-Webster, 1501)." The gait cycle describes the motion of each leg during one stride (period of time between heel strike of one foot and the subsequent heel strike of the same foot) and provides fundamental terminology to describe the differences in the various gait patterns that will be discussed throughout this paper. The gait cycle is divided into two phases (which refer to a single leg), known as the stance phase and the swing phase (Figure 1). A typical gait cycle begins with the heel contact, which is also the beginning of the stance phase. The individual is in a period of double support, or being supported by both legs, until toe-off of the opposite foot occurs at approximately 10% of the total time of the cycle. While the individual is in a single-limb stance, the individual's center of mass (COM) passes over the stance foot as the opposite leg moves into position for its heel contact, which occurs halfway through the cycle. A second period of double support occurs and lasts for approximately another 10% of the cycle (between approximately 50% and 60%, as shown in Figure 1). The leg then experiences toe-off, beginning swing phase, which comprises approximately the last 40% of the cycle. Swing phase is further divided into three periods: initial swing, mid-swing, and terminal swing. When the swing leg is again at the point of heel contact, a single gait cycle has been completed.

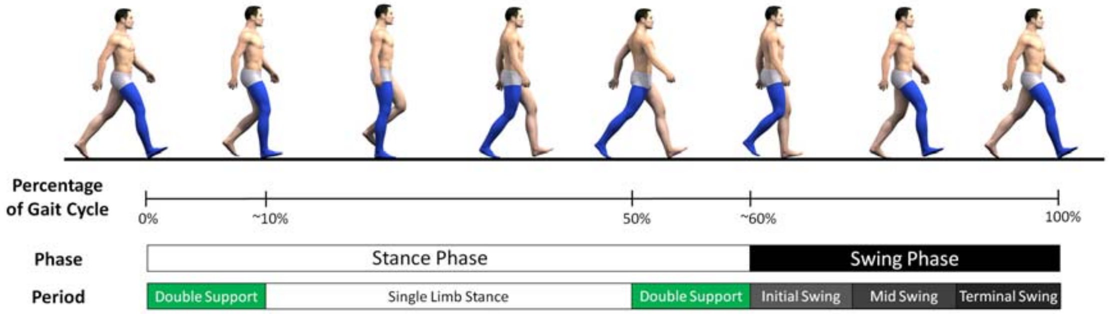

Figure 1. The gait cycle is divided into phases and periods.

As seen in Figure 2, a single step occurs between a heel strike of one foot and the next heel strike of the opposite foot, and a step length is the distance between the contralateral heel strikes. A single stride occurs between a heel strike and the subsequent heel strike of the same foot, and a stride length is measured as the distance between two adjacent heel strikes of the same foot. Similarly, step width (also known as the base of support) is the distance measured side-to-side between the midline of each heel, as shown in Figure 2. Cadence is defined as the number of steps taken in a given time interval and is often reported as steps per minute.

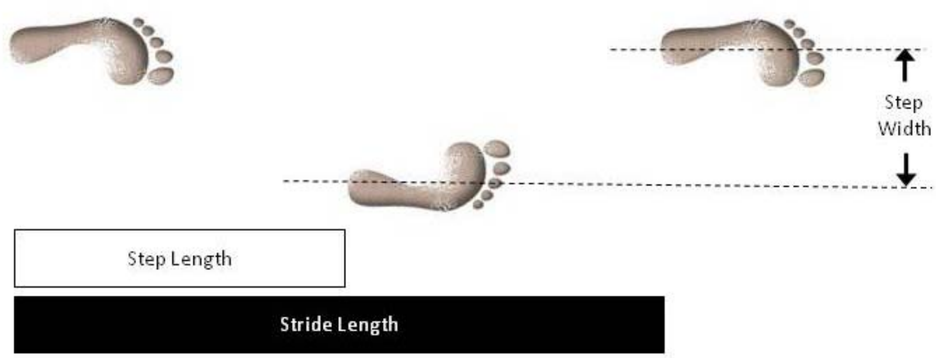

Figure 2. The length of a step is measured from heel strike of one foot to heel strike of the opposite foot. The length of a stride is measured from heel strike of one foot to the next heel strike of the same foot.

The walking speed of an individual (in m/s) can be calculated by multiplying their stride length (meters) and their cadence (steps/minute) and dividing the result by 120. Studies have shown that walking velocity is dependent on many variables, including leg length and age, and average walking speed is reported to be approximately 80 meters per minute (Hageman and Blanke, 1986, 1385; Blanke and Hageman, 1989, 146; Winter, 1990, 347). Researchers have investigated the energy expenditure and efficiency associated with walking and running, and analyses of gait data have shown that preferred walking speed is also the most energy efficient (Inman, 1968, 34; Cavagna and Kaneko, 1977, 475; Waters and Mulroy, 1999, 207-213). Additionally, the velocity that corresponds to the preferred walk-to-run transition occurs approximately at the velocity where walking actually takes more energy than does running (Cavagna and Kaneko, 1977, 475; Diedrich and Warren, 1995, 183).

During each swing phase, the swing foot experiences a minimum vertical toe clearance of approximately 1 to 2 centimeters (Winter et al., 1990, 347; Winter, 1992, 47; Mills and Barrett, 2001, 433; Moosabhoy and Gard, 2006, 497; Mills et al., 2008, 103; Best and Begg, 2008, 1148-9). Factors such as wearing multifocal spectacles increase within-subject variability in minimum toe clearance (Johnson et al., 2007, 1469). Walking on a sloped surface also changes the minimum toe clearance variability, and contrasting findings exist as to whether advancing age affects toe clearance (Khandoker et al., 2007, 4890; Winter et al., 1990, 347; Mills et al., 2008, 103). Minimum toe clearance and toe clearance variability are important to understand when evaluating if an individual may be likely to experience a trip in a given environment.

Each step results in the foot imparting a force against the ground, thereby also resulting in a ground reaction force directed from the floor to the foot. The ground reaction force initiates at heel strike, is initially directed backwards with respect to the individual, and serves as a braking force during that time. As the individual's center of mass moves forward, the point of action of the ground reaction force moves from the heel towards the toes, and the force is directed upward. Due to the motion of the center of mass as it passes over the stance foot, the magnitude of the ground reaction force slightly decreases at approximately halfway through the stance phase before again increasing. Late in the stance phase, the ground reaction force is directed in the direction of travel, acting as a push-off force just prior to toe-off. Figure 3 shows the ground reaction force through the stance phase of gait and illustrates why the force through time during a single stance phase is sometimes referred to as a "Butterfly Diagram" (Cohen et al., 1980, 96).

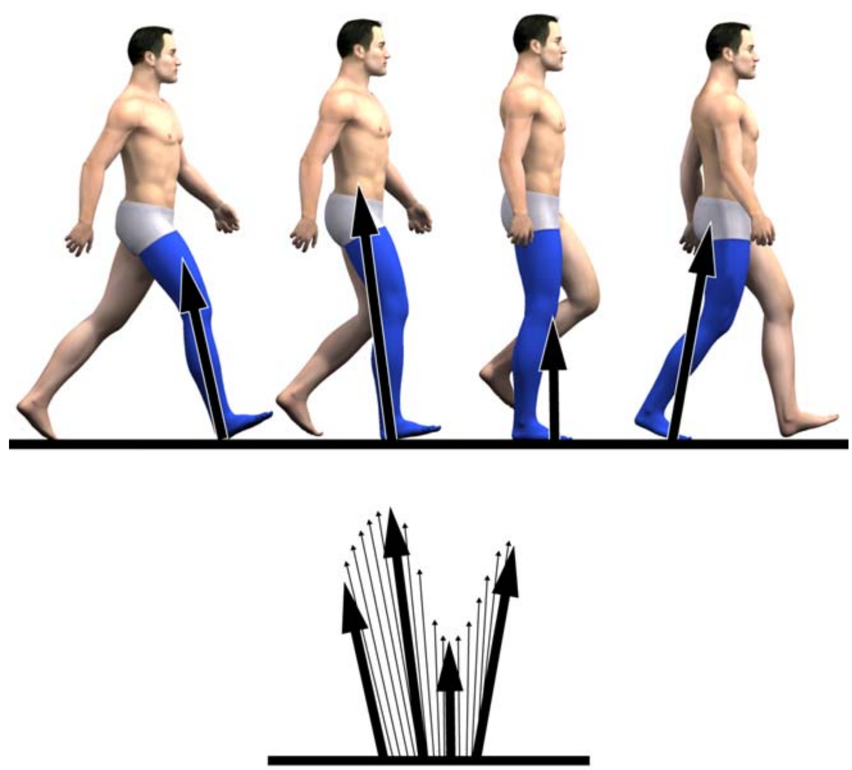

Figure 3. The composite of the ground reaction force through the stance phase of gait is sometimes referred to as a "Butterfly Diagram."

Developmental Gait

Children begin to walk at approximately one year of age, after which they undergo rapid development and maturation of their gait pattern for the next several years. Toddlers encounter several challenges in learning how to walk and developing a mature gait pattern, not the least of which is due to growth, resulting in constantly changing body dimensions and proportions.

Sutherland et al. conducted testing of 186 normal children between one and seven years of age and provided a detailed comparison of differences in gait between children and adults. They reported that at one year, children exhibit differences in their gait pattern when compared to adults, including a higher step frequency, lack of swinging opposing extremities, flexed knee during stance, and plantar-flexed foot at foot-strike (Sutherland et al., 1980, 345). During swing phase, there is diminished dorsiflexion, and increased hip flexion, pelvic tilt, hip abduction, pelvic rotation, hip-joint rotation, and knee-joint rotation. Additionally, there is external rotation of the hip throughout the cycle, and the duration of single-limb stance is less than what is seen in adult gait.

By age two, Sutherland et al. (1980, 346) noted a markedly more mature gait. Pelvic tilt, hip abduction, and hip external rotation are decreased, and the knee experiences greater flexion after foot strike and extends prior to toe-off. At the ankle, there is less plantar flexion such that the initial contact is made with the heel, and the ankle dorsiflexes during swing. About 75% of two year olds demonstrate reciprocal swinging of the opposite limb. By the age of seven, the gait of children is similar to adults' mature gait. A few differences still exist, including a greater cadence, slower walking velocity, increased pelvic rotation, increased hip joint rotation, and increased swing phase hip abduction.

Sutherland et al. (1980, 347-8) listed the five determinants of mature gait: duration of single-limb stance, walking velocity, cadence, step length, and ratio of pelvic span to ankle spread. Table 1 represents Sutherland's findings regarding how these parameters develop between the ages of one and seven years old.

Table 1. Rapid changes occur in a child's gait pattern between the ages of one and seven years old (*Source*: Data summarized from Sutherland et al., 1980, 347-352).

Determinant of Mature Gait	Change Seen from One- to Seven-Year-Old Children
Duration of single-limb stance	Increases from approx. 32% in 1 y.o. to approximately 38% in 7 y.o.; Variability is also reduced
Walking velocity	Increases from approx. 0.6 m/s in 1 y.o. to approx. 1.2 m/s in 7 y.o.
Cadence	Decreases from over 175 steps/min in 1 y.o. to less than 150 steps per minute in 7 y.o.; Variability is also reduced
Step length	Increases from approx. 0.2 m in 1 y.o. to approx. 0.5 m in 7 y.o.
Ratio of pelvic span to ankle spread	Increases from approx. 1.3 in 1 y.o. to approx. 2.3 in 3 ½ y.o., after which it remains relatively constant

Sutherland (1997, 169) presented data supporting a stabilization in the gait pattern that occurs at approximately 4 years of age, which he attributed to both a maturation of the central nervous system and a child's physical growth. He attributed further gait maturation after age 4 to physical growth. Adolph et al. (2003, 494) provided an extensive review of contributing factors to how an infant's gait develops and concluded that experience was one of the strongest predictors for adopting a more mature gait pattern. Because some important gait parameters (e.g., step length and walking velocity) are so closely tied to stature, Todd et al. (1989, 200) has developed graphs of expected variations in these related parameters for growing boys and girls.

As briefly described above, mechanical efficiency is an important aspect of the adult gait pattern and preferred velocity. An investigation of the role of mechanical energy in toddlers' gait showed that the amount of positive work performed during a gait cycle to lift and accelerate the body's center of mass was the most significant contributor of the total mechanical work in this age group, and the specific work performed was greater than that done by adults for the same distance traveled (Hallemans et al., 2004, 2430). Because mechanical efficiency is a hallmark of normal gait in adults, it seems that achieving this efficiency is also an underlying goal of a child's developing gait.

Aging Gait

Elderly individuals face a unique set of challenges to achieving a normal and consistent gait pattern, including a change in their body composition involving an increase in their percentage of body fat (Hughes et al., 2002, 476), declining muscle strength (Hughes et al., 2001, B209), and musculoskeletal and neurological changes (Harmon, 1981, 7124-8). As a result, elderly adults adopt a more conservative gait pattern, demonstrated by shorter step and stride length, more time spent in stance, reduced ankle range of motion, reduced pelvic obliquity, reduced walking velocity, and increased step timing variability (Menz et al., 2003, 139-40; DeVita and Hortobagyi, 2000, 1806; Hageman and Blanke, 1986, 1385; Malatesta et al., 2003, 2251; Murray et al., 1969, 169-70), although contrasting findings have been reported with respect to changes in cadence as a function of age (Oberg et al., 1993, 210; Winter et al., 1990, 347; DeVita and Hortobagyi, 2000, 1806). Across walking speeds, the energy cost of walking was higher for the elderly than for younger controls, indicating that the gait pattern adopted by the elderly is less efficient than in younger individuals (Malatesta et al., 2003, 2251).

Bohannon et al. timed both comfortable and maximum gait speed in men and women aged 20 to 79 years and showed a slight decrease in comfortable walking speed in both genders as a function of age and a more marked decrease in maximum walking speed as a function of age as shown in Figure 4 (Bohannon et al, 1997, 17).

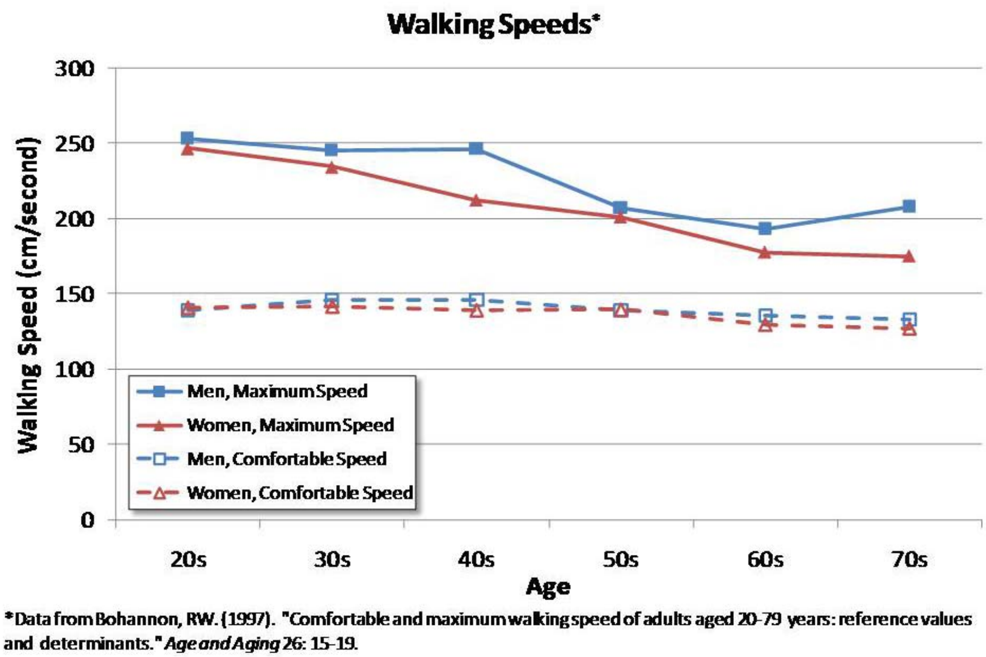

Figure 4. Walking speeds in men and women decrease as a function of age (*Source:* Data from Bohannon, 1997).

When comparing young men and elderly men, Winter et al. (1990, 347) found no significant difference in cadence between the two age groups, but found that stride length was significantly decreased in the elderly group (1.55 m for young men versus 1.39 m for elderly men). Additionally, Winter et al. found that elderly men spent a significantly higher percentage of the gait cycle in stance (65.5 % in elderly men compared to 62.3% in young men), but did not find a significant difference in toe clearance or toe clearance variance between the two age groups (1.27 cm in young versus 1.11 in elderly). Other studies have contrasted this finding, reporting an age effect on toe clearance (Menant et al., 2009, 394). Elble et al. (1991, 1-2) also compared young adults and neurologically healthy elderly adults. The research group reported that the average velocity of elderly walking was 20% less than that of young walking and that the fast velocity of elderly walking was 17% less than the fast velocity of young walking (Elble et al., 1991, 2). They attributed this change in velocity to comparable differences in stride length, as they reported that cadence did not differ between young and elderly for either fast or for natural walking. Increased hip rotation was discussed as the main contributor for an increased stride length and corresponding faster walking velocity.

Shumway-Cook et al. (2002, 672-4, 677, 679) observed two groups of community-dwelling elderly adults (≥ 70 years) during three trips that they made: to the grocery store, to the physician, and a recreational trip. Nineteen of the adults were considered "without disabilities," as evidenced by their ability to walk ½ mile or climb stairs without assistance), and seventeen were considered "with disabilities." The researchers stated that there was not a uniform decrease in encounters with environmental challenges due to mobility disability of the older adult navigating their environment. The distance walked, environmental factors (traffic lights, precipitation, temperature, outdoor light), and attentional demands (familiarity, distractions) were not found to show significant differences between the two. Although walking speed could not be specifically determined within this study, there was a significant difference between the percentage of subjects without disabilities who were able to maintain the speed of those around them (100%) compared to the subjects with disabilities (10%). Subjects without disabilities carried more packages (1.56 (SD 1.5) versus 0.98 (SD 0.8)), and those packages weighed more (6.7 pounds (SD 9.3) versus 1.5 pounds (SD 1.8)) than those with disabilities. Although there was not a significant difference between the percentage of subjects in each group who encountered one flight of stairs, more of the subjects without disabilities encountered two flights of stairs (47% versus 4% trips encountered), and less of the subjects without disabilities utilized elevators (10% versus 25% trips encountered). More of the subjects without disabilities encountered grass (37% versus 23% trips encountered) and obstacles (13% versus 3% trips encountered). Ninety-five percent of subjects without disabilities traveled unaccompanied, while 24% of subjects with disabilities traveled without a companion. Although there was no difference in the number of "stops" and "back ups" between the two groups, the subjects without disabilities were more likely to change their head orientation, reach forward, reach up, reach down, and change directions. This study demonstrated that elderly adults with mobility impairments make more conservative choices while navigating their environment when participating in activities of daily living.

Gait Termination

Although the majority of gait research has focused on steady-state walking on a level surface, more effort has been made to further understand gait termination in the last decade. Ground reaction force data have shown that push-off forces decrease and braking forces increase during gait termination (Jaeger and Vanitchatchavan, 1992, 1234). When subjects were asked to stop in one step, with one foot beside the other, muscle activity associated with braking began approximately 150 milliseconds after a visually cued stop signal in the stance limb and began approximately 330 milliseconds after the signal in the swing limb (Crenna et al., 2001, 1063 and 1067). The stance leg's muscle braking activity was characterized as a distal-to-proximal (foot-to-hip) activation sequence, while the swing leg's activity was characterized as proximal-to-distal (hip-to-foot). In comparing normal walking, planned stopping, and unplanned stopping, it was found that the peak decelerating ground reaction force was highest for the lead limb in unplanned stopping (Bishop et al., 2004, 136). Additionally, the deceleration force increased as an individual's cadence increased, but cadence did not affect the duration of muscle activity relative to the time of peak loading.

Younger subjects walked faster, and also were able to stop more quickly, as compared to their elderly counterparts (Tirosh and Sparrow, 2004, 247-9). Although elderly subjects were more likely to stop in two steps compared to their younger counterparts (who were more likely to stop within one step of the visual cue), the average stopping distance for all conditions examined was not significantly different between the two age groups. A subsequent study by the same research group indicated that because elderly subjects were more likely to need an extra step to stop during gait termination for normal walking speeds, this resulted in longer stopping time and stopping distance (Tirosh and Sparrow, 2005, 282-4). Similar to Crenna's work, Tirosh and Sparrow found the muscles responsible for braking to respond more quickly in the stance leg than in the swing leg. The stance leg responded to the stopping cue through activation of the tibialis anterior and inhibition of the soleus. The swing leg responded to the stopping due through the activation of the soleus and inhibition of the tibialis anterior to control ankle plantar flexion; in addition, the gluteus medius and vastus lateralis were activated before the subsequent heel contact.

Menant et al. (2009, 68-9) investigated the effects of age, flooring surface condition, and shoe type on parameters of gait termination, including walking velocity, time to last foot contact, total stopping time, stopping distance, step length post-stopping-cue, step width post-stopping-cue, and base of support length. It was found that older individuals exhibited slower walking velocity, longer time to last foot contact, greater stopping distance, and greater step width post-stopping cue than their younger counterparts. Individuals walked more slowly on irregular and wet surfaces than on the control surface, the total stopping time and distance were greater on the wet surface than the control surface, and the base of support was decreased on the wet surface than on the control surface. Although several types of shoes were investigated (standard, elevated heel, soft sole, hard sole, flared sole, bevelled heel, high-collar, and tread sole), the only significant difference found with respect to the standard shoe across age groups and surface conditions was an increased total stopping time in the soft sole shoe. There was a significant surface x shoe interaction found between the standard and high-collar shoe, with the total stopping time being less on wet surfaces for individuals wearing high-collar shoes. Perry et al. (2007, 96) has demonstrated that increases in the shoe midsole hardness increased the range of the medial-lateral (side-to-side) center of mass movement relative to the lateral border of the individual's base of support during the first single-support phase following the cue to stop.

When individuals' vision was occluded through the use of special glasses, the decrease in velocity (slowing) of the forward progression was delayed (Perry et al., 2001, 31). When cutaneous sensation was decreased through hypothermic anesthesia (by submersing the soles of their feet in ice water), the step length of the second step was increased. Reduced sensation also increased the loading rate of the first foot contact after the stop cue as well as the loading rate of the final foot contact.

Navigating Turns

Hase and Stein (1999, 2914) described two strategies employed for turning: the step turn and the spin turn. The step turn was utilized when the individual turned to the opposite side as the foot which was in front at the time of the stimulus. This turn was more easily executed because the base of support was wider throughout the maneuver. The spin turn was utilized when the individual turned to the same side as the foot which was in front at the time of the stimulus, and it involved spinning the body around the leading foot. The muscle activations necessary to execute the turn occurred within a step after the cue to turn, and most individuals were able to complete the turn without breaking their established gait rhythm.

Glaister et al. (2008, 3091) described the initiation step, apex step, and termination step involved in turning (as shown in Figure 5). When compared to straight walking, the initiation step was characterized by a medial impulse, a larger braking impulse, and a smaller propulsive impulse. The apex step involved a large lateral impulse, a braking impulse similar to straight walking, and a large propulsive impulse. During the termination step, a medial impulse was present throughout stance, the braking impulse was smaller than straight steps or other turning steps, and the propulsive impulse was larger than straight steps or the initiation step but was smaller than the apex step.

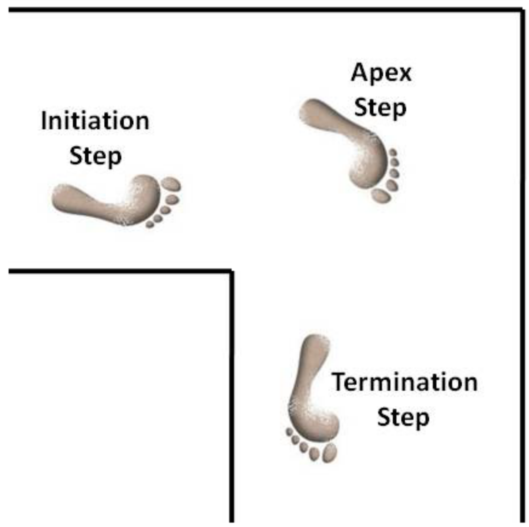

Figure 5. A turn is composed of an initiation step, an apex step, and a termination step.

Stair Negotiation

Ascending and descending stairs require a different gait pattern than level walking. Riener et al. (2002, 32-3 & 43) asked ten subjects to ascend and descend three inclinations of a five-step stair case, ranging from 24 degrees to 42 degrees. Stair inclination did not affect most of the temporal gait cycle parameters and ground reaction forces evaluated, but a significant difference in joint angles, joint moments during stance, and joint powers was observed. The authors commented that they did not observe a shift in the motor patterns as a function of inclination angle, suggesting that there may be an inclination angle where a switch between level gait and stair gait would be seen.

As with level gait, stair gait can be divided into stance phase and swing phase. During stair ascent (going up stairs), the stance phase comprised 65% ± 4% of the gait cycle, which can be further subdivided into foot contact, weight acceptance, vertical thrust, single-limb support, forward continuance, and double support (Zachazewski et al., 1993, 414-5). Swing phase consisted of the remaining 35% of the stair ascent gait cycle, and is further subdivided into foot clearance and foot placement. Stair ascent was characterized by McFadyen and Winter as "pulling and pushing the body through concentric contractions of the rectus femoris, vastus lateralis, soleus and medial gastrocnemius (1988, 738)." Typical kinematics of ascending stairs are shown in Figure 6.

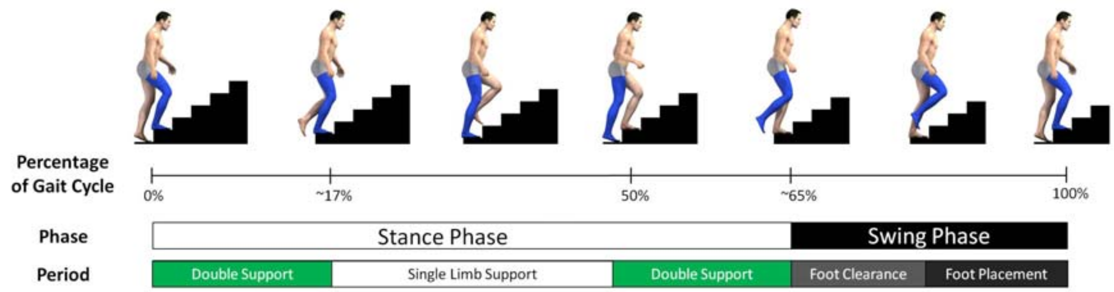

Figure 6. Stair ascent is divided into the stance and swing phase.

During stair descent (going down stairs), the stance phase comprised 68% ± 2% of the gait cycle, which can be further subdivided into weight acceptance, forward continuance, and controlled lowering (Zachazewski et al., 1993, 417). The swing phase of stair descent involved leg pull through and foot placement. Stair descent was characterized by McFayden and Winter as controlling the force due to gravity through eccentric contractions of the rectus femoris, vastus lateralis, soleus, and medial gastrocnemius (McFayden and Winter, 1988, 738). Typical kinematics of descending stairs are shown in Figure 7.

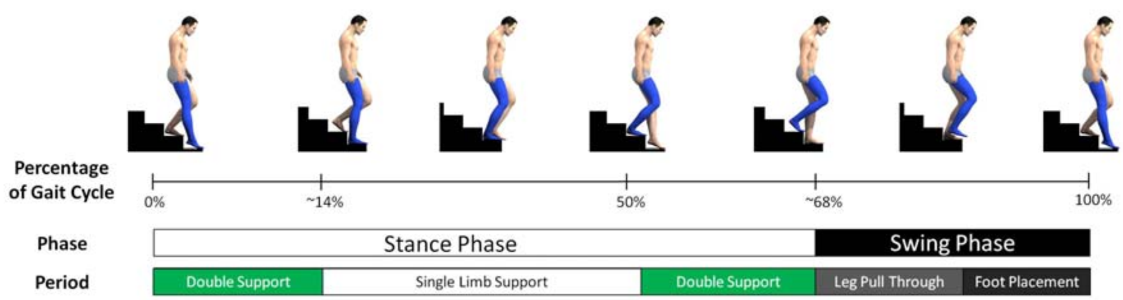

Figure 7. Stair descent is divided into the stance and swing phase.

Researchers instrumented hip implants in four patients and took measurements during activities of daily living, including stair ascent and descent (Bergmann et al., 2001, 861). It was shown that the average hip joint contact force was 251% body weight for stair ascent and was 260% body weight for stair descent; in comparison, the average joint contact force was 238% body weight when walking at approximately 4 km/hr (Bergmann et al., 2001, 867).

Footwear

A literature review conducted by Menant et al. (2008, 1167) indicated that walking barefoot, in socks, or in high-heels increases the risk of falling in the elderly, and they recommended that elderly individuals should consider low-heeled shoes with firm, slip-resistant soles while they are inside or outside of their home. Similarly, Koepsell et al. (2004, 1497-8) reviewed the type of footwear worn at the time of 327 falls by persons 65 years and older and found an increased fall risk for people who were barefoot or in stocking feet, and fall risk was the lowest for those wearing athletic shoes. In a study involving ninety-five older people, Sherrington and Menz

(2003, 213) asked participants to identify their footwear at the time of a fall-related hip fracture. The most common footwear were recorded to be: slippers (21.6%), walking shoes (16.5%), sandals (8.2%), and barefoot (7.2%).

As part of a study, an occupational therapist visited the homes of discharged patients to conduct routine home assessments. Of the 178 homes visited, a recommendation to change footwear was made in 43 homes (or 24% of those visited). At the 12-month follow-up, only 54% of those patients to whom footwear changes had been recommended had actually modified their behavior (Cumming et al., 1999, 1398). A separate study involving 170 home visits of patients who had previously fallen revealed that "foot and footwear" was identified as a problem by the nurse in 37% of the cases (Lightbody et al., 2002, 206). A telephone survey of elderly people was utilized to investigate their likelihood of wearing a "sturdy" shoe (defined as "shoes with laces and a nonskid sole, that is, walking or athletic shoes or men's dress shoes") at some point during the week. Sixty-four percent of respondents stated that they had worn sturdy shoes daily, 14% stated they had worn them four to six times during the week, and 22% responded wearing them three times per week or less (Dunne et al., 1993, 246). Interestingly, when asked what shoe they were wearing at the time of the telephone call, 9% of those 85 and older reported wearing dress shoes, while only 2% of the respondents 65 to 84 years old were wearing dress shoes. Additionally, the older respondents were less likely to ever wear sturdy shoes.

The Functional Reach Test, Timed Up & Go Test, and 10-Meter Walk Test are standard ways to test an elderly person's ability to perform tasks they are likely to encounter in everyday life. When female subjects ranging in age from 65 to 93 years old were asked to perform these tasks in three shoe conditions (barefoot, walking shoes, dress shoes), the type of shoe that they were wearing did have a significant effect on their performance (Arnadottir and Mercer, 2000, 20 & 22). A similar study involved completing the Berg Balance Scale while women who were 60 years and older were either barefoot or wore their own shoes. Horgan et al. (2009, 64) reported that of the 14 functional tests on 100 women aged 61 to 95 years old, wearing shoes compared to being barefoot improved results in ten of the tasks (sit-stand, standing unsupported, standing unsupported with eyes closed, standing unsupported with feet together, forward reach, looking over shoulder, turning 360 degrees, placing alternate foot on step, tandem stand, and single-leg stance).

Investigation of hip joint loading was accomplished through the use of an instrumented hip implant in a single patient, which showed that walking with sport shoes, leather shoes, hiking boots, and clogs produced small changes when compared to walking with bare feet (Bergmann et al., 1995, 818-20 & 825). They also found that shoes with hard soles tended to create higher hip loads. Light et al. (1980, 478) reported that measurements of accelerations in the tibia and skull resulting from heel strike during gait are dependent on footwear, with a hard leather heel producing a different response than a shock-absorbing heel. Finally, Kerrigan et al. (1998, 1400) reported that females wearing high-heeled shoes experienced different forces across the patellofemoral joint and altered lower extremity joint kinematics when compared to the barefoot condition. These studies indicate that the construction of the shoe affects the loading experienced by the individual.

Although many studies regarding footwear involve elderly populations, a study involving women aged 21 to 40 years old showed that younger women adopt different strategies to negotiate stairs depending on their footwear (Heller et al., 2008, 673). The women were more likely to utilize the available hand rail, keep their foot within the confines of the stair tread, and proceeded more slowly when wearing high heels compared to when wearing sneakers or flipflops while ascending and descending stairs. Another study involved subjects ranging in age from 19 to 50 years old participating in a balance-beam task with different shoe conditions (Robbins et al., 1994, 118-21). The authors reported that shoe midsole thickness and hardness both affected the likelihood of a balance failure, and significantly more balance failures occurred when barefoot than when wearing shoes. Shoes with the thickest and softest midsoles were related to the poorest stability and shoes with the thinnest and hardest midsoles were related to the best stability.

Six shoe designs (standard shoe, standard shoe with soft sole, standard shoe with hard sole, high collar shoe, elevated heel shoe, and tread sole shoe) were worn by both young and elderly (>65 years) men and women, and gait data were collected (Menant et al., 2008, 1971). A summary of the significant kinematic and kinetic differences is shown in Table 2.

Table 2. Several gait parameters can be affected by differences in shoe construction (*Source:* Data summarized from Menant et al., 2008, 1974).

Gait Parameter	Significant difference when compared to standard shoe condition
Step width	Hard sole decreased
Double support (% stance time)	Elevated heel and hard sole increase
Walking velocity	Soft sole and tread sole increase
Lateral center of mass base of support margin	Soft sole and high collar increase
Posterior center of mass base of support margin	Elevated heel and hard sole decrease
Vertical loading rate	Elevated heel decrease

A similar study was published by Menant et al. in 2009 (395), which showed that additional parameters, including walking velocity, toe clearance, and heel velocity, were affected by shoe type. Other parameters, including cadence, step length, and shoe-floor angle, were not significantly affected by shoe type.

Distracted Gait

Many activities may distract an individual while they are walking including walking with a friend and carrying on a conversation, reading a map, being “lost in thought,” eating, listening to music, or using an electronic device. Researchers have recently begun to investigate how individuals who have chosen to engage in a distracting activity while walking may interact differently with their environment. Bungum et al. (2005, 275-6) performed an observational study at an intersection near a university. Approximately 20% of those observed were engaged in distractive behaviors, including eating, drinking, smoking, using headphones, or using a cell phone, and the authors reported that those who were distracted exhibited less cautionary behaviors while crossing the street. Another observational study concluded that females who were talking on a cell phone were less likely look at traffic before or while crossing the street and less likely to wait for traffic to stop before crossing (Hatfield and Murphy, 2007, 197). The study also revealed that males crossed unsignalised intersections more slowly and crossed signalised intersections more quickly when talking. Females crossed the unsignalised intersections at approximately the same speed and crossed signalised intersections more slowly when talking. Nasar et al. found that when individuals were engaged in a cell phone conversation, they were less able to notice out-of-place objects in their surroundings than when they were not engaged in conversation (Nasar et al., 2008, 70). The findings of Kuzel et al. (2008, 667) supported those of Nasar, in that with an increasingly difficult cell phone conversation, subjects were less able to detect objects or recognize details and were likely to take more steps over a longer period of time while covering the same distance. These studies suggest that biomechanical parameters, such as number of steps taken to walk a given distance and walking speed, are affected by engaging in distractive behavior. Additional research needs to be done to more fully characterize the biomechanical implications of distracted gait.

Slips, Trips, and Falls

Because the focus of this paper is to describe biomechanical factors that may play a role in the risk of falling, the biomechanics of falls will only be touched upon, but many resources exist that explore this topic much more fully. Slipping has been defined by Gronqvist (2008, 687) as follows: “Slipping is a sudden loss of grip, resulting in sliding of the foot on a surface due to a lower coefficient of friction than that required for the momentary activity, often in the presence of liquid or solid contaminants.” Figure 8 demonstrates typical kinematics of a slip event, characterized by a weight-bearing limb sliding on the surface and the individual falling as a result of not being able to maintain their center of mass over their base of support.

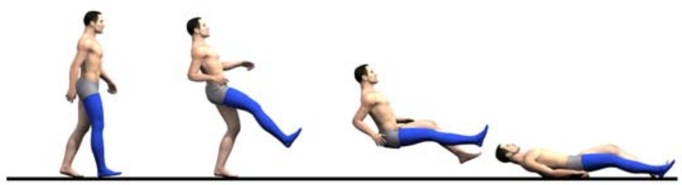

Figure 8. A slip involves the sliding of the foot (shown in blue) on a surface, often in the presence of a contaminant.

A trip is caused when an object near the ground arrests the forward motion of the swing limb, and a fall occurs when the center of mass is not maintained over the individual’s base of support. Typical fall kinematics resulting from a trip are shown in Figure 9.

Figure 9. A trip occurs when an object near the ground arrests the forward motion of the swing limb (shown in blue).

Conclusion

Although walking is one of the most common activities among humans, there are many challenges that one is likely to encounter over the course of a day, which may be compounded by the challenges of being either a child or an elderly adult. Understanding the various aspects of human gait is essential for the safety professional as they are determining how individuals may interact with a specific environment.

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