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The Effect of an Incline Walking Surface and the Contribution of Balance on Spatiotemporal Gait Parameters of Older Adults

Richard A. Ferraro
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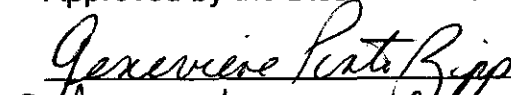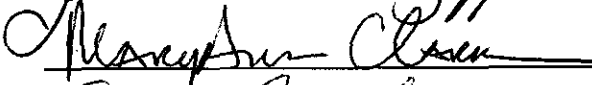
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Submitted in partial fulfillment of the
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ABSTRACT

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Chair, Dr. Genevieve Pinto-Zipp

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Seventy-eight active, older adults participated in this study (mean age, 77.8; SD, 4.8). The Berg Balance Scale (BBS) and Dynamic Gait Index (DGI) were used in this study to assess overall stability. A third measure, the Gait Stability Ratio (GSR), was calculated directly from output measures provided by the GaitRite® computerized walkway system. Acting as their own controls, all subjects walked five times each on a level surface and an inclined walkway. Dependent t-tests were used to determine statistical significance between level and incline surfaces for cadence, step length and velocity. A repeated measures ANOVA was performed to determine differences in means for the higher risk subgroups comparing their level and incline walking patterns. The level of significance was set at $p = 0.05$.

Results of this study indicate that cadence, step length and velocity significantly decreased on inclines while GSR increased relative to subjects' level ground walking patterns. While cadence and velocity support previous incline studies with younger subjects, the decrease in step length suggests a different pattern adapted by older adults on inclines (Kawamura et al., 1991; McIntosh et al., 2005). In the higher risk subgroups, only the results from the repeated measures ANOVA using the DGI showed a significant increase is GSR on the inclined surface indicating decreased stability relative to level ground.

These findings are important and have significant clinical value. Increased GSR measured on inclines indicate more time spent in double support and suggests the primary goal, even in healthy adults, is stability. These results suggest that clinicians incorporate more challenging balance activities for healthy older adults such as dual tasks and varying terrain obstacle courses.

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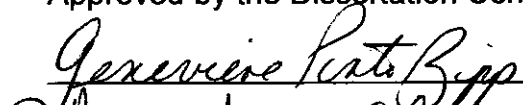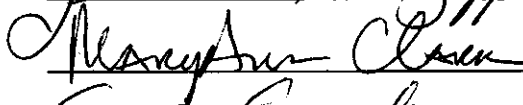

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are committed. To my wife, Laurie, thank you for all that you have sacrificed. Thank you for listening, encouraging and calming. You are amazing and I truly would not be in this position in my career without your continuous support, love and understanding. Many times when I wavered about my fortitude and abilities you were the source from where I drew my strength. I love you.

To all of my coworkers, friends, family and subjects who volunteered their time and energy in order to help me, I am greatly indebted to you all. I have realized that a journey of this length cannot be completed without the kind and generous hearts of others.

DEDICATION

This work is dedicated to the memory of my mother, Beverlee Ferraro, who instilled in me the meaning of dedication and hard work from an early age as she raised a family on her own and made many sacrifices along the way. Every day she reminded me of the importance of a good education. Rest in peace mom, I did it!

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Chapter I

INTRODUCTION

Background of the problem

Recent studies of population trends indicate an increasing percentage of older adults in the United States with the largest percentage of growth of any age range in the 65 years and over age group (Summer, Friedland, Mack & Matthieu, 2004). Included in that growth are those over the age of 85 who are expected to increase by 7.3 million by the year 2020 (Fowles & Greenberg, 2006). Often termed the "baby boomer effect", this large increase in the population of older adults (those born between 1946-1964) will have major implications on the economic and healthcare systems in the next two decades. Despite the evidence that the population of the United States as a whole is getting older, adults 65 years and over, on average, are maintaining higher levels of activity than previous generations (Kasper & O' Malley, 2007).

The U.S. Department of Health and Human Services (2006) reports that most older adults (65 and over) are less dependent on others and less likely to use assistive devices than those in past decades. Older adults are maintaining active lifestyles well into their eighties due to the advent of new medications and an overall greater emphasis placed on health and wellness (Kasper & O' Malley, 2007). Although self-reports of increased activity levels in older adults are more common, research shows a rapid decline in the physiological efficiency

throughout the body from the sixth to eighth decades of life which ultimately increases the demand placed on older adults while they walk (Fuller, 2000).

Generally, overall strength decreases with age due to various changes that have been observed at all levels of the body. At the cellular level of muscles there is evidence that the cross sectional area decreases with age (Williams, Higgins & Lewek, 2002). In addition, relative to non-contractile tissue there is less functional, contractile tissue present in the muscle fibers of older adults (Kent-Braun, Ng and Young, 2000). Fast-twitch fibers (type II fibers), which are important for faster walking speeds, are also found in lesser numbers in older adults (Frontera & Hughes, 2000). The decreasing number of functional motor units has also been linked to the decreased efficiency of muscle activity seen in older adults (Cunningham, et al., 1987; Gajdosik & Vander Linden, 1996). Ultimately, all of these changes at the microscopic level contribute to a progressive decline in isolated, maximum voluntary contractions as well as muscular endurance in the lower extremity. As a result there is a progressive decrease in the efficiency of gait between the sixth and ninth decades of life (Vandervoot, et al. (1992).

In addition to the strength decreases associated with increasing age, numerous studies have documented a progressive loss in overall joint range of motion (ROM) due to an increase in non-contractile tissue, decreased muscle length, and more dense capsular tissue around joints (Gajdosik & Vander Linden, 1996; Gajdosik, 1997; Gajdosik, Vander Linden & Williams, 1999). Other causes of decreased joint motion that occurs with increasing age include less than

optimal joint alignment and pain due to osteoarthritic changes (Valderrabano et al., 2007). Decreased ROM at the ankle has been correlated with decreased balance scores on the Timed Up and Go and gait portion of the Tinetti Performance Oriented Mobility Score (POMA) (Mecagni et al., (2000). Similar findings regarding dorsiflexion ROM limitations have also been found to contribute to gait deviations such as decreased step length and walking speed in adults (Mueller et al., 1995).

Despite the evidence that suggests that age-related physiological changes have an effect on gait and function in healthy older adults, relative to younger adults, there are few studies examining older adult gait patterns. Even fewer studies show the effects of various walking surfaces on the gait patterns of older adults. In the majority of studies examining older adults results suggest this population walks at a slower speed, and with increased cadence, increased double support time, and a decreased step length when compared to younger populations (Winter et al., 1990; Kang & Dingwell, 2007). Winter (1991) concluded that these shorter, more frequent steps are a response to older adults' perceptions of instability. Rogers, Cromwell and Grady (2008) confirmed this conclusion demonstrating that older adults slow down and take more steps than younger controls as they ambulate in more challenging conditions (Rogers, Cromwell & Grady, 2008). Paradoxically, as a result of trying to make themselves more stable, older adults may be increasing their risk for falls as cadence; step width, stride to stride parameters and stride length all become

more variable as walking speeds decrease. (Maki 1997; Hausdorff, Rios & Edelberg, 2001; Kang & Dingwell, 2007).

There are many possible explanations for the differences in gait between younger and older adults and most are still debated in the literature. Researchers generally agree, however, that older adults prioritize stability whereas younger populations prioritize forward progression, mobility and efficiency while walking (Winter et al., 1990; Cromwell & Newton, 2004; Rogers, Cromwell & Grady, 2008). It is not clear whether this shift in gait paradigms is the cause or the result of many of the spatiotemporal and kinematic differences that are apparent in older adult gait patterns on level ground. Although the results from studies on level ground walking are both informative and descriptive, generalizations of these results may not apply to walking on different surfaces such as inclines.

In comparison to level ground walking, the ability to walk on inclines requires a different motor pattern in the lower extremity. This motor pattern requires increased force output by lower extremity musculature and increased range of motion particularly at the ankle (Saunders, Inman & Eberhart, 1953; Hirakozu & Yamamuro, 1987; Andersson & Forssberg, 1989; Vogt & Banzer, 1999; Leroux, Fung & Barbeau, 2002). Relative to level ground, motor patterns of younger healthy subjects show differences during incline walking such as increased torque occurring at the hip and ankle musculature as these muscles act to simultaneously stabilize each respective joint and propel the mass of the

body upward (Leroux, Fung & Barbeau, 2002; 2006; Lay, Haas, Nichols & Gregor, 2007).

Generally, a resultant decrease in cadence and velocity occurs while both step length and stride length increase as the slope of the incline increases (Kawamura, Tokuhiro & Takechi, 1991; Lange & Hintermeister, 1996; Sun, Walters, Svensson & Lloyd, 1996; McIntosh, Beatty, Dwan, Vickers, 2005). Data also suggest that there is a larger excursion of movement of the center of mass while walking up and down inclines relative to level ground. This larger movement excursion is indicative of greater balance requirements (Kawamura, Tokuhiro and Takechi, 1991). Despite strong evidence to suggest the presence of different gait patterns among young healthy adults on incline surfaces when compared to level ground, no evidence exists on whether these patterns also occur in older adults.

Although gait and balance have been thoroughly investigated independent of one another, the link between balance and walking is not clearly understood. Gait patterns of younger adults on level ground are characterized by phases of instability that allow for forward progression and lateral shifting of the body's center of mass with each step (Nashner, 1980; Perry, 1992). Older adults on level ground decrease their velocity, take shorter steps and minimize their lower extremity motion to increase stability while they walk (Winter, Patla, Frank & Walt, 1990; Cromwell, Newton & Forrest, 2001; 2002). While these patterns on level ground are well described, far less is known about the effect of incline walking on various aspects of balance during gait. Based on the existing

literature with young adults on inclines it is logical to expect even greater differences in older adults (> 70 years old) as they walk on inclines due to the effects of decreased ROM, strength, and balance.

Statement of the problem

While data exist for healthy younger populations walking on level ground, much less is known about the gait patterns of older adults (> 70 years old). Even more apparent is the lack of data describing how older adults negotiate inclines. Based on the rapid increase in the numbers of older adults in this country, the physiological decline that is associated with aging, and the evidence that suggests age-related changes in gait patterns on level ground, a closer analysis of older adults' walking patterns on common community surfaces such as inclines is merited.

Purpose of the Study

The purpose of this study is twofold: 1.) To examine the effects of an incline surface on spatiotemporal gait parameters among older adults and 2.) to identify any relationships that exists between balance and incline spatiotemporal gait parameters.

Hypotheses:

H1: Cadence in older adults will decrease during incline walking as compared to level walking.

H2: Step length in older adults will increase during incline walking as compared to level walking.

H3: Gait velocity in older adults will decrease during incline walking as compared to level walking.

H4: Gait stability ratios (GSR) will increase during incline gait as compared to level ground walking.

H5: Older adults with lower Berg Balance Scale scores (< 45) will demonstrate an increased GSR during incline walking compared to level ground.

H6: Older adults with lower Dynamic Gait Index scores (≤ 19) will demonstrate an increased GSR during incline walking compared to level ground.

Chapter II

Review of the literature

Population trends

The United States is currently going through a dramatic demographic transformation due to a progressively aging population. The "baby boom" generation (those born between 1946-1964) is nearing retirement age and adults, in general, are living longer. Based on these two trends, forecasts for the next twenty years include the largest percentage growth for those over the age of 65 (Summer et al., 2004). The population over 65 was 35 million in 2000, 40 million in 2006 (a 15% increase in that population) and is expected to increase to 55 million in 2020 (36% increase). Similar projections are expected for those over the age of 85. In 2000, the population over 85 numbered 4.2 million, increased 40% to 6.1 million in 2006 and is expected to grow to 7.3 million by 2020 (Fowles & Greenberg, 2006).

While population studies indicate a trend towards a greater percentage of older adults (those adults 65 and older), demographic studies indicate that older adults in the U.S. are more active today when compared to previous generations. These active lifestyles include regular participation in sports, employment and community activities. The advent of new and improved medications and health awareness affords this population a greater ability to maintain active lifestyles longer in their lifetime as well as a decreased dependence on others (Kasper & O'Malley, 2007). The U.S. Department of Health and Human Services reports

that among non-institutionalized older adults ages 65-74 only 11% reported

needing assistance to ambulate (2006). These figures increase to 21% among

ages 75-84 and then again to 41% in persons older than 85 years of age (Kasper

& O'Malley, 2007). Although these numbers inherently rise as age increases,

they indicate that a large percentage of older adults continue to ambulate without

assistance. Less reliance on assistive devices and less assistance required from

others supports the notion of the growing independence of older adults in today's

society. Therefore, it is crucial for clinicians to have a comprehensive

understanding of gait and, more specifically, how gait parameters change with

age across different environmental conditions.

History of gait analysis

Gait has been studied using various instrumentation and data collection

methods such as electromyography (EMG), force plates (kinetics), motion

analysis (kinematic) and electronic walkways (spatiotemporal) which have

contributed to our growing understanding of the complexities of locomotion.

As far back as 340 BC, Aristotle began to document various movements

of humans and animals. These movements were analyzed using more accurate

methods of analysis as they were developed to understand and document the

various complexities of the human gait pattern. Contributions from others

including Cartan, Descartes, Newton, Euler and Weber continued to improve the

quality of gait analysis in three dimensional planes. Theories developed based on

3-dimensional planes of motion (Cartesian planes) and Newtonian mechanics

and physiology. Muybridge took observational analysis one step further and

began recording gait and other human activities with sequential cameras capturing temporal characteristics associated with movements. This lead to the development of the first biomechanics laboratories established by Inman and Eberhart in the early 1900's (Baker, 2007).

One biomechanical method used to measure the timing, duration and intensity of muscles is electromyography (EMG). EMG requires the use of in-dwelling needles or surface electrodes on the motor points of various muscles to quantify output. EMG output affords researchers the ability to assess the quality and timing of deep and superficial muscles during various activities and contractions. Much of the contemporary integration of clinical electromyography (EMG), observational gait analysis and digitization are credited to Perry and Sutherland.

Perry (1992) clearly demarcated the sub-phases of gait. The following description is based on her model using stance and swing phases as references for the timing of various muscle activity. The complete gait cycle (GC) was described along a continuum from 0-100%, with 0% representing initial contact and 60% representing the period of terminal stance. The stance phase of gait equals 60% of the total time to complete one gait cycle and encompasses the entire time that the foot is in contact with the ground. Stance is sub-divided into four phases: initial contact, loading response, mid- stance and terminal stance. Based on this temporal continuum, initial contact comprises 0-2% of the GC and requires 10-15 degrees of ankle dorsiflexion range of motion (ROM) defined as the final stages of swing and into initial contact (Murray, 1967; Perry, 1992;

Ostrosky, 1994). This motion is maintained by the concentric action of the pretibial muscles dorsiflexing the ankle.

As the body moves forward over the foot, the loading response phase of stance begins (0-10% GC). This is characterized by an eccentric contraction of the pretibial muscles as the foot is lowered to the ground in a plantarflexion arc (Skinner et al., 1985; Perry, 1992). Simultaneously, the tibia is drawn forward and the combination of these two actions contributes to limb progression as the body's weight rolls forward onto the heel. This controlled ankle plantarflexion also acts as a shock absorption mechanism minimizing forces translated proximally upward through the limb (Skinner et al., 1985).

The next phase, mid-stance (10-30% GC), is the time period when the foot is entirely in contact with the ground as the tibia and center of mass of the body continue over a fixed foot. Muscle activity during midstance phase includes primarily the soleus and secondarily the gastrocnemius as passive dorsiflexion is restrained by these posterior tibial muscles (Skinner et al., 1985). This muscle activity serves to slow down and control tibial advancement as the body prepares for propulsion (Sutherland 1980; Skinner et al., 1985). During this stage approximately 5-10° of passive ankle dorsiflexion is required for smooth translation of the body's forward momentum (Perry, 1992).

During terminal stance, (30-50% GC) the heel of the foot begins to rise as the body prepares for the swing phase. Maximum ankle dorsiflexion ROM is required (10-12°) as the tibia approximates the fixed foot during forward progression and the center of mass moves ahead of the forefoot (Close & Inman,

1952; Sutherland et al., 1980). Pre-swing phase then ensues as the ankle moves through the largest plantarflexion arc of the entire gait cycle peaking at ranges of 15-25°.

According to Perry (1992), pre-swing (50-60% GC) requires limited muscular contributions from the lower limb. Pre-swing is also referred to as the weight release or weight transfer period as an abrupt transfer of body weight unloads the limb (Close & Inman, 1952). Concurrent with initial contact of the contralateral limb is an ipsilateral increase in ankle plantarflexion and knee flexion occurring to prepare for unloading and clearance of the foot.

In the initial swing phase of gait (60-73% GC) the limb rises off the ground as the primary goal of foot clearance and advancement of the limb from its trailing position is attained (Perry, 1992). In an attempt to minimize energy expenditure foot clearance is maintained at approximately one centimeter above the ground (Perry, 1992). Muscle activity in this sub-phase includes primarily the hip flexors and secondarily from the ankle dorsiflexors as both groups return to neutral in preparation for initial contact. Mid-swing (73-87% GC) is driven by active hip flexion as the swinging limb is advanced to a position anterior to the body. The knee is extended primarily as a response to gravitational forces and the ankle regains a neutral position relative to the tibia. In the terminal swing phase (87-100% GC), completion of limb advancement occurs with full knee extension and heel contact with the ground completing the gait cycle (Perry, 1992).

During each phase of gait, proper timing and amplitude of various muscle activity ensures efficiency and smooth transitions between each phase. Therefore, it is necessary to understand the individual muscular contributions in terms of maximum voluntary contractions (MVC) in the lower extremity. MVC is typically measured by one isometric contraction of a muscle is performed while holding against an external force. The percentage of force relative to the maximum that a muscle can produce provides researchers with an overall understanding for each muscle's contribution during level ground walking. Initiation of the gait cycle begins with the onset of pretibial muscle activity in the pre-swing phase (Close & Inman, 1952). The extensor hallucis longus (EHL) contracts initially at 8% of MVC while tibialis anterior and extensor digitorum longus (EDL) follow rapidly in mid-swing, reaching levels of 35% MVC. Maximum activity in these muscles occurs at initial contact as the tibialis anterior reaches 45%, EHL 35%, and EDL 25% MVC respectively (Simon et al., 1978; Perry, 1992).

There is slightly more variability among ambulators in the triceps surae activity but it is generally accepted that 93% of the plantarflexor torque is provided by the soleus and gastrocnemius (Perry, 1992). Soleus activity begins during the loading response and rises to 25% of MVC throughout midstance. At the onset of terminal stance there is a rapid increase in amplitude to 75% MVC (45% GC) indicative of the power requirements needed for push off. Soleus activity drops quickly to zero by the onset of double stance. The medial head of the gastrocnemius parallels the activity of the soleus while lateral head

gastrocnemius activity is delayed until midstance. As a whole, the rise in the activity of the gastrocnemius is slower and less intense than that of the soleus, peaking during terminal stance at 60% MVC (Simon et al., 1978; Perry, 1992).

While all of the aspects of the gait cycle in younger healthy adults are well understood, some caution should be used when applying these conclusions to older adult populations. There are many studies that demonstrate the rapid decline of various systems in the body with increasing age. Because of the decline in physiological efficiency, older adults who have range of motion, balance, visual or strength deficits present with gait patterns that are different than younger adults. The knowledge gained from studies of older adult gait provide information on the numerous physiological changes seen with aging such as decreased strength (Vandervoot, et al., 1992), decreased range of motion (Boone, Walker & Perry, 1981; Valderrabano et al., 2007), and decreased balance, contributing to a greater incidence of falls in this population (Daubney & Culham, 1999). Evidence suggests that these physiological changes contribute to variations in gait parameters such as decreased step and stride length (Winter et al., 1990; Perry, 1992; Kang & Dingwell, 2007), increased percentage of double support (Whittle, 2002) and a shift in the purpose of gait in older adults from efficient advancement to an increase in stability (Cromwell & Newton, 2004; Rogers, Cromwell & Grady, 2008). Therefore, it is important to understand normal physiological changes that occur as humans advance in age is paramount.

Age-related physiological changes

Physiological changes in aging muscle tissue are well documented in the literature and begin at the microscopic level. Williams, Higgins and Lewek (2002) demonstrated that skeletal muscle cross sectional area (CSA) decreases with age. Not only does the CSA decrease, but research suggests that less contractile tissue and more non-contractile tissue exists in the muscles of older adults (65-83) as compared to younger adults (26-44) (Kent-Braun, Ng & Young, 2000). In older adults, type II fast-twitch muscle fibers demonstrate greater selective atrophy than do type I fibers (Frontera & Hughes, 2000). Larssen, Sjodin and Karlsson (1978) showed that adults ages 20-29 had 39% type I, slow-twitch fibers while the adults age 60-65 had 66% type I fibers in their lower extremities. In addition to selective fiber atrophy, an overall decrease in the number functional motor units in older adults has been reported (Cunningham, et al., 1987; Gajdosik & Vander Linden, 1996).

Although somewhat variable in number, type II fast-twitch fibers are found in large numbers in the gastrocnemius and tibialis anterior, two muscles that are very active in critical sub-phases of the gait cycle (Porter, Vandervoot & Lexell (1995). The tibialis anterior controls the foot from the initial contact phase to the loading phase while the gastrocnemius aids in the pre-swing and propulsion stage of gait. If, as suggested in the literature, selective atrophy occurs in type II muscle fibers of these lower limb muscles, different gait patterns would ensue. A resultant gait pattern with poor foot clearance during swing, decreased knee

control during stance phase and decreased eccentric control of the foot at heel strike would most likely result (Perry, 1992).

Type II muscle fibers also contribute to overall strength during a muscle's maximum voluntary contraction (MVC). A MVC is a measure of strength often defined in terms of force (i.e. lbs, Newtons, kg.) or as a moment around a joint (i.e. Newton-meters, etc.) (Bernard, 2006). MVC was shown to significantly decline in both men and women between the sixth to ninth decades of life (Vandervoot, et al., 1992). Declines in dorsiflexion strength were greater in men where mean values dropped from 43Nm to 29Nm as compared to females where mean values dropped from 27Nm to 19Nm (Vandervoot et al. 1992). Evidence of decreases in MVC translate into difficulties maintaining contractions at the stages of gait that have the highest requirement of peak muscle torques such as midstance (for plantarflexors) and initial contact to the loading response duration (for dorsiflexors). It is these two phases of gait that facilitate smooth transitions from swing to stance, ensuring maximum efficiency and safety. Without the ability to maintain a strong isometric contraction in the lower leg muscles, the ability to propel the body mass forward in the pre-swing phase and stabilize the lower limb during initial contact is compromised (Perry, 1992). The result is an inefficient and unstable gait pattern.

Functionally, limited ankle dorsiflexion strength has been linked to decreased postural stability as well as a decreased ability to adapt to changing surfaces during gait (Daubney and Culham, 1999). High correlations were found between dorsiflexion and eversion strength in predicting Berg Balance Scale

(BBS) scores, and dorsiflexion strength alone was shown to be a valid predictor of falls in adults with a mean age of 74.82. Equally important is the contribution dorsiflexion strength plays in reactive balance control. Dorsiflexion is the initial response to a destabilizing force by providing a forward directed counter movement (Wolfson, Whipple, Amerman & Kleinberg, 1986). Daubney and Culham (1999) contend that distal lower extremity muscle strength may predict functional balance scores, such as those on the Berg Balance scale and Timed Up and Go (TUG) test. They found that in the group reporting no falls, dorsiflexion and eversion force accounted for 58% and 48.4% of the scores on the BBS and TUG respectively. They suggest the main factor in predicting falls with these two tests is decreased strength in these two muscle groups (Daubney & Culham, 1999). Further findings from this study suggest that fallers have lower dorsiflexor and hip extensor force output than non-fallers (Daubney & Culham, 1999). The results of this study suggest a strong correlation between balance and strength are fundamental to understanding gait and how it contributes to an individual's functional level.

In addition to age related physiological decline in muscle and strength, passive extensibility also decreases with age. An increase in non-contractile tissue in the calf tendons and muscles is the primary reason for this length change. Shortening of calf muscles at the ankle acts to limit dorsiflexion range of motion (Gajdosik & Vander Linden 1996; Gajdosik, 1997; Gajdosik, Vander Linden & Williams, 1999).

Animal studies suggest multiple factors contribute to overall joint stiffness. In a classic study done with cats, Johns and Wright (1962), found joint stiffness in mid-ranges is primarily due to the joint capsule (47%) and muscle (41%) and less to tendons (10%) and skin (2%). However, at the end of available range of motion the effects of stiffness, specifically from the tendons, became more significant. Peacock (1966) demonstrated no increases in range of motion after removing the layers of fascia, muscles and superficial bands of dense connective tissue from a four week old flexion contracture in a canine. It was not until the volar capsule was incised that a sudden and visible increase in motion resulted, implicating the joint capsule as the major contributor to joint stiffness.

Dense connective tissue restrictions also contribute to age related decreases in joint ranges of motion in humans. Vandervoot et al. (1992), documented decreases in active ankle dorsiflexion in men and women in two age groups, 55-60 years old and 81-85 years old. Mean values for men decreased from 20.7 to 10.1 degrees between age groups while values for women decreased from 20 to 13.5 degrees between groups. Boone, Walker and Perry (1981) also showed a gradual decrease in active ankle dorsiflexion due to aging. Dorsiflexion decreased from 12.4 degrees in subjects 40-54 years old to 8.2 degrees in subjects ages 61-69 years old. In that same study, plantarflexion ROM also decreased.

The limitations in ankle dorsiflexion have a clear impact on the gait pattern such as reduced toe clearance during the swing phase. Further, the lack of dorsiflexion range of motion leads to changes in gait parameters such as

decreased step and stride length (Winter et al., 1990; Perry, 1992; Kang &

Dingwell, 2007), increased percentage of double support time (Whittle, 2002) and

a shift in the priorities of gait from efficient advancement of the body's center of

mass as younger adults to maximizing stability with increasing age (Cromwell &

Newton, 2004; Rogers, Cromwell & Grady, 2008).

Osteoarthritis (OA), a common joint disease affecting articular cartilage,

also occurs more frequently with increasing age. Over time OA and its

associated pain causes a decrease in strength and joint ROM as joint capsules

tighten (Valderrabano et al., 2007). Osteophytic formation, less than optimal joint

alignment and resultant ankle pain have been cited as the primary reasons why

OA is particularly debilitating in the elderly (Valderrabano et al., 2007). In a study

comparing older adults with ankle OA, total ankle replacements and age-

matched controls, those with ankle OA showed a significant decline in six of

seven measurable gait parameters including step length, stride length, cadence,

and velocity (Valderrabano et al., 2007).

Subjects with and without osteoarthritis (OA) also demonstrate differences

when comparing ground reaction forces in various planes of motion. Those with

OA demonstrated the largest relative reduction in force occurring in the

transverse plane while attempting to stabilize the ankle. This was attributed to

localized atrophy, weakness of the lower leg muscles and stiffness of the

surrounding soft tissues (Valderrabano et al., 2007). Researchers agree that age

related physiological changes decrease overall motion, strength and efficiency of

all joints (Boone, Walker & Perry, 1981; Perry, 1992; Vandervoot et al., 1997;

Daubney & Culham, 1999; Valderrabano et al., 2007). Therefore, researchers have begun to compare the differences in gait patterns on level ground between younger and older adults. As a result of these efforts there are some common results in the changes observed in older adults' walking patterns.

Step width variability, defined generally as the differences between data sets as measured between each heel was found to be significantly greater in older adult adults (Maki & McIleroy, 1997; Owings & Grabiner, 2004; Brach et al., 2007). Maki and McIlroy (1997) suggest that variability in step width is the primary contributor to falls in the elderly. Similarly, Hausdorff, Rios and Edelberg (2001) suggest that one predictor of falls in the elderly (> 70 years old) is increased variability in other parameters of gait and an inability of the body to adapt quickly enough to changes in speed or terrain. However, this study highlights stride time variability rather than step width variability as the primary predictor of falls. Here 40% of older adults who reported falling during the previous twelve months demonstrated stride time variability of 106ms while those who did not report any falls had stride time variability of 49ms (Hausdorff, Rios & Edelberg, 2001). They also found that stride time variability was significantly correlated with factors such as strength, balance, gait speed, functional status and even mental health. However, none of these other factors was found to be predictors of falls.

In an attempt to separate the effects of gait speed and age on gait variability, Kang and Dingwell (2007) studied eighteen older (mean age = 72) and younger (mean age = 23), height and weight-matched subjects as they walked

on a treadmill at speeds ranging between 80% and 120% of their preferred

speed. By controlling for treadmill speed for both older and younger subjects,

they were able to determine that older adults exhibited greater variability in trunk

roll (defined as movement of the pelvis in the frontal plane) ($p = 0.003$), trunk

pitch (defined as a sagittal plane motion similar to trunk flexion)($p = 0.022$), step

length ($p = 0.005$) and stride length ($p = 0.018$) independent of speed. However,

it was found that speed did increase variability of stride time, frontal plane hip

and knee motions, knee internal/external rotations and all trunk motions across

all subjects. Therefore, the increased variability in the older adults was attributed

more to decreased leg strength and passive ranges of motion than from the

slower speeds as they walked (Kang & Dingwell, 2007).

In an earlier gait study, no significant differences in velocity were found

between groups of healthy older (mean age = 72) and younger adults (mean age

= 25) (Grabiner, Biswas & Grabiner, 2001). However, significant main effects

were found for age and stride width ($p = 0.007$), and age and step length ($p =$

0.002). These findings support previous studies that suggest that step width

variability may be the most important outcome variable in older adults when

identifying those at risk for falls (Gabell & Nayak, 1984; Hausdorff et al., 1997;

Maki, 1997).

Variability in cadence also occurs in older adults as a more stable gait

pattern is established. Measured as the number of steps per minute, cadence

varies more in older than younger adults. Intuitively, the primary concern of older

adults is maintaining a stable gait pattern, especially when walking on different

surfaces (Rogers, Cromwell & Grady, 2008). Shorter stride lengths and an increased double support time during gait ensues (Winter et al., 1990). Cromwell and Newton (2004) propose that shorter stride lengths decrease forward progression of the body and limit time spent in single limb stance and thus, influences balance.

Regardless of the compensatory strategies used to establish a more stable gait pattern, overall variability increases in older adults during different phases of gait. Some older adults maintain stability without increasing cadence or decreasing their rate of speed, while others decrease cadence and slow their pace resulting in increased overall variability during gait. Slower gait patterns may result from the perception of how much stability is needed, resulting in shorter and more frequent steps (Winter, 1991). Although older adults (mean age = 68; range 65-85 years) showed greater evidence of adopting a more stable gait pattern on all surfaces, younger (mean age = 27.2; range 21-35 years) and older subjects slowed and took more steps under challenging sensory conditions (Rogers, Cromwell & Grady, 2008). Therefore, as older adults experience greater challenges to their balance, greater variability in gait parameters often results.

Variability in muscle activity exists when comparing the gait patterns of older and younger adults. As a measure of the timing and amplitude of muscle activity, EMG burst patterns in the lower extremity muscles also show greater variability in older adults when compared to younger adults. Tirosh and Sparrow (2005) compared older men (mean age = 74) to younger men (mean age = 23)

and showed fewer muscles activated in the swing limb of older adults with less frequent responses in the soleus and gluteus medius as compared to younger controls. In general, older adults also demonstrated slower responses in EMG burst activity in the stance leg during gait termination EMG burst following a visual stopping stimulus (215ms for older subjects and 176ms for younger subjects). Tirosh and Sparrow (2005) suggest that a failure to activate soleus and gluteal muscles compromises the extensor torque needed to maintain the center of gravity anterior to the base of support. If this same burst activity in the soleus and gluteus medius can be generalized to larger populations of older adults on level ground, one might hypothesize that an even greater hip extensor torque would be needed to control the forces of gravity when walking on inclines. As seen during level ground walking, resultant adaptations made by physiologically less efficient older adults on inclines might be quite different when compared to incline gait profiles of younger, healthy adults.

Reduced amplitudes of EMG recordings were also found when healthy able-bodied subjects, ranging in age from 23-58, walked over level ground and on treadmills but at manually selected slower rates (.60ms-.80ms) relative to each subject's preferred walking speed (Nymark et al., 2005). All lower limb segments decreased in overall motion at slower walking speeds. Foot flat posturing at heel strike (less dorsiflexion), loss of knee flexion during weight acceptance, decreased plantarflexion at toe off, loss of trunk forward lean and decreases in flexion/extension of the hip were all resultant effects of slower walking speeds (Nymark et al., 2005). When these subjects walked on treadmills

or over ground at manually selected slower walking speeds, changes in EMG recordings were noted. EMG amplitudes were significantly lower than when walking at normal rates of speed. In addition, proximal musculature around the hip joint demonstrated increased periods of co-contraction and did not show the distinct peaks and transition periods that occurred at natural speeds. Similar to the pattern of co-contraction seen with the hip musculature comparable phasic patterns, with decreased amplitudes, were found at the medial gastrocnemius and tibialis anterior relative to the natural speeds (Nymark et al., 2005). Ultimately, researchers concluded that control of foot position at heel strike (tibialis anterior) and forward propulsion (medial gastrocnemius) result in phasic activities of these lower limb muscles (Nymark et al., 2005).

Winter et al. (1990) investigated the gait patterns of 15 healthy and fit older adults (mean age = 68) using kinematic and kinetic data collection methods with the primary goal of identifying differences in gait patterns of older adults and compare them to younger adult gait patterns previously established (Winter et al., 1990). In addition, these researchers identified consistent motor patterns in adults beyond measuring gait parameters such as velocity, step length and step width variability.

One major difference between younger and older adults was a drastic reduction in the vigor or power generation of push-off at 40-45% of the gait cycle (terminal stance). There was a significant reduction in mechanical energy generation while absorption of energy increased showing an overall decrease in the push off mechanism. Winter et al. (1990) proposed that the terminal stance

phase is normally destabilizing and that this decreased "vigor" may be an attempt by older adults to reduce the potential for instability. Another explanation for this phenomenon is a decrease in ankle plantarflexion strength. With the increasing effects from gravity at this phase of the gait cycle such minimal deficits in strength may translate into a much larger reduction in power generation, making walking at faster speeds or on inclines more difficult (Winter, 1990). Winter (1990) also suggested that decreased push off induces significantly shorter step lengths and increases double support times in older subjects. Double support time for older adults (mean age = 68; range 62-78 years old) was 31% of the total gait cycle as compared to 24.6% in the younger population (mean age = 24.6; range 21-28 years old). In contrast, cadence and stride length differences in healthy older adults were not significant when compared to younger gait profiles. Researchers attributed these non-significant differences to the high fitness levels and relative good health of the older adults who participated in this study (Winter, 1990). The results contributed to the data collected on level walking comparing younger and older adult populations has provided important information regarding the different characteristics during the gait cycle between the two groups.

Gait profiles of older adults on level ground show significant differences in kinetic, kinematic and spatiotemporal outcomes such as increased time spent in double support, shorter step lengths, wider base of support, slower balance reactions and altered phasic activity of lower extremity musculature (Winter, 1990; Kawamura, Tokuhiro & Takechi, 1991; Sun et., 1996; Daubney and

Culham, 1999). Speculation continues regarding the origin of these differences but most researchers agree that the gait characteristics and objectives of older adults vary when compared to younger adults. The different patterns may be due to the inherent physiological limitations associated with aging such as decreased range of motion, strength and balance (Vandervoot et al., 1992; Tirosh & Sparrow, 2005; Kang & Dingwell, 2007; Valderabanno et al., 2007). In addition to the physiological changes with age there is also a change in the priorities during gait. This shift from propulsion to stability also contributes to the measurable and observable differences seen among older adults when compared to younger adults (Winter, 1991; Cromwell & Newton, 2004).

Although level ground walking is well understood among different age groups, few studies have been completed investigating incline walking, particularly among older adults. Results from incline gait analysis will provide information on surfaces that require more balance requirements, strength and range of motion to negotiate safely (Lay et al., 2007; Cromwell & Newton, 2008). Resultant gait patterns on incline surfaces are certain to change as older adults negotiate these common community barriers due to the physiological limitations as well as the increased demand required to walk on inclines,.

In the community, man-made inclines are often chosen as a replacement for stairs because inclines are often perceived by older adults as an easier obstacle to negotiate. Due primarily to decreased physiological efficiency seen in many older adults, negotiating an incline surface becomes more challenging than level walking. For younger adults this increased challenge on incline surfaces is

evident in studies and is measured in terms of increased torque (Lay et al., 2006; Tokuhiro, Nagashima & Takechi, 1985), and decreased stability in lower extremity musculature (Cromwell & Newton, 2004).

Investigation into the adaptations that take place while walking on inclines and torque analysis is effectively captured using electromyography (EMG). EMG allows researchers to demonstrate a variety of muscle characteristics on inclines relative to level ground walking. Tokuhiro et al., (1985) effectively captured EMG activity of various muscles in the lower extremity including the tibialis anterior (TA), gastrocnemius (Gc), rectus femoris, semitendonosis, and gluteus maximus on 3°,6°,9°, and 12° surfaces. Among other findings they reported significant differences in the tibialis anterior and gastrocnemius output between level and incline surfaces.

In healthy subjects on level surfaces, TA activity begins at toe off, continues through the swing phase and ceases with an eccentric lowering to foot flat and Gc activity is seen from late stance to toe off during level walking (Close & Inman, 1952; Sutherland et al., 1980; Tokuhiro et al., 1985; Perry, 1992). However, on inclines and declines (at 6°, 9° and 12°), both muscles showed longer phasic activity as the slope became greater. While walking on 6° inclines, EMG activity of the TA began in late midstance and continued until foot flat. At 9° and 12° TA activity began at 70% of midstance and continued until foot flat (versus at toe off during level walking). Among healthy subjects, with progressively increasing incline slope angles (with a maximum of 39% grade)

longer burst durations in the anterior tibialis, gluteus medius, rectus femoris, semimembranosis, and vastus medialis occurred (Lay et al., 2007).

Interestingly, while walking down slopes, the Gc showed a similar pattern of longer duration contractions initially activated at heel strike and ceasing at 70%, 80% and 50% of push off at 6°, 9°, and 12° degrees, respectively. In essence, the Gc and TA contract simultaneously in order to stabilize the talocrural joint and assist in toe clearance (Tokuhiro et al., 1985). This pattern of increased burst duration in the lower extremity was confirmed in a more recent study on inclines at a greater slope.

Lay et al., (2007) found that overall power generation during incline walking occurred for 68% of stance phase as compared to 43% during level ground walking. During decline walking, power absorption, defined by eccentric or decelerating muscle activity, occurred for an average of 83% of stance, compared to only 53% on level ground. These changes resulting from incline and decline walking indicate an increase in the total work done by the lower extremity musculature. The increased power requirements support previous studies that suggest different patterns of muscle activity when healthy younger adults walk on inclines compared to level ground walking. In addition, while walking on inclines, EMG magnitude and duration of seven of the eight muscles recorded showed significant increased activity while decline walking produced increased activity in three muscles when compared to level walking (Lay et al., 2006; 2007). More specifically, the greatest increase was recorded in gluteus maximus activity in upslope walking at 259% of the mean gluteus maximus

activity found in level walking. This increased activity and subsequent power generation was suggested as one of the primary sources in moving the body's mass up the slope (Lay et al., 2007). As found with kinematic results in the Leroux et al., (2002; 2006) study, these EMG patterns seen in healthy adults provide further impetus to look more closely at older adults on inclines.

Unfortunately, despite increased attention to incline walking the majority of results can only be generalized to younger adults (< 50 years old). Little data exists on older adults as they negotiate inclines. Despite the omission of older adults in contemporary incline studies the results found among younger adults provide a foundation for future comparison between these two age groups. Similar to level ground gait studies incline gait studies, have used a variety of data collection methods with different hypotheses.

As with level walking, incline gait studies include data collection on the head, neck and trunk (Cromwell, 2003; Leroux, Fung & Barbeau, 2002, 2006) as well as the lower extremity stance and swing (McIntosh, et al., 2005; Tokuhiro, Nagashima & Takechi, 1985; Kawamura, Tokuhiro & Takechi, 1991; Sun et al., 1996). Each study's findings contribute to understanding how the body adapts to varying levels of inclines.

In an effort to include a more holistic view on incline gait analysis a general understanding of what occurs at the head, neck and trunk is necessary.

Cromwell (2003) described head, neck and trunk segments to be less stable when subjects ascended inclines producing the greatest challenges to head stability. When subjects walked on inclined surfaces, head, neck and trunk

positions varied more than in trials on level surfaces. This research suggests in the head, neck and trunk certain movement strategies develop in an attempt to stabilize the head and eyes. On inclines, stabilization of the head and neck is thought to be more difficult leading to greater variability and different movement strategies. More specifically, Cromwell (2003) concluded that these patterns developed in an attempt to minimize eye movement and further suggested that the otoliths of the inner ear may also be a contributing factor to a specific pattern development. By stabilizing the head and minimizing movement of the otoliths of the inner ear, subjects attempt to offset the increased balance demands that are induced by incline surfaces.

The relationship of the trunk and pelvis during gait on level ground was initially described by Saunders, Inman and Eberhart (1953) who hypothesized that pelvis and trunk together minimize displacement of the center of gravity. Using kinematic techniques Leroux, Fung and Barbeau (2002) expanded on movement patterns of the trunk and pelvis and included the hip, knee and ankle, while investigating postural strategies of younger adults (mean age = 34; range 25-54 years old) during walking on 0°, 5° and 10° inclined surfaces. All subjects displayed minimal trunk forward bending during level walking. By bending the trunk forward, subjects moved their center of mass slightly ahead of their center of foot placement and, subsequently, the force of gravity assisted in accelerating the body forward. In addition, the increase in forward bending at the trunk during uphill walking is thought to assist the lower limbs in generating more momentum to counteract the increased resistance due to gravity (Leroux, Fung & Barbeau,

2000). As the slope of the incline increases a larger anterior hip rotation and relative hip flexion occurs. These observations are in agreement with previous studies that showed the same relative trunk positioning when subjects walked on inclines (Hirokazu & Yamamuro, 1987; Vogt & Banzer, 1999).

Leroux and associates (2006) compared eight subjects who suffered an incomplete spinal cord injury (SCI) (ASIA Level D) to an age and gender-matched control group on incline surfaces that varied from -10° (decline) to 10° (incline). Postural adaptations that developed on inclines and declines in patients with decreased control at distal segments of their limbs were analyzed. Previous findings on SCI patients found that this population used mainly hip strategies when adapting to an incline which differed from healthy subject patterns that showed clear and consistent patterns of adaptations at the hip, knee and ankle (Leroux et al., 1999). In a more recent study, Leroux et al. (2006) suggested that this unique adaptation by the SCI population may be due to decreased control at the more distal joints of their limb due to the resultant impairment of their injury. One of the results was a significant decrease in plantarflexion moments when comparing results of the SCI subjects to the control group. Although the patterns of trunk and pelvic strategies were similar between the experimental and control groups the total excursion was much larger in the SCI group (Leroux et al., 2006). It was theorized that these larger excursions were due to propioceptive, sensory and motor loss in the experimental group's lower extremities.

While walking on declines, the forward lean of the trunk and pelvis in normal subjects decreases to accommodate for the decreasing grade of incline in an effort to oppose the forces of gravity. However, incomplete SCI subjects maintained forward trunk flexion on declines which was atypical when compared to control subjects on the same decline angles (Leroux et al., 2000; 2006). These researchers suggest that maintaining a flexed posture on decline surfaces may be a compensatory mechanism for instability in the lower extremities in the SCI group. Biomechanically, the flexed posture may prove to be an unsafe postural adaptation because the subject's center of mass shifts anterior instead of posterior to offset the effects of the ramp and gravity. With the weight of the head and trunk anterior to the foot placement, subsequent momentum and gravitational forces on the body may be too much to overcome for patients with spinal cord damage (Leroux et al., 2002; 2006). These findings lead to speculation that older adult subjects, who also demonstrate decreased distal control due to age related changes, may also behave differently than younger healthy subjects when negotiating inclines and declines.

In addition to the clear differences represented by EMG studies, spatiotemporal data collection during gait also suggests differences in walking patterns between level and incline surfaces. McIntosh and colleagues (2005) measured eleven healthy male subjects' gait patterns using spatiotemporal, kinetic and kinematic data analysis. Their design included a walkway randomly varied to grades of 0°, 5°, 8°, 10°. Spatiotemporal measures revealed that cadence increased slightly with a greater decline angle that plateaued at angles

greater than -5 degrees. However, the opposite was true while walking uphill as cadence decreased slightly with progressively larger incline angles. These findings support earlier studies which also noted decreases in cadence with increases in incline grades (Sun et al., 1996; Kawamura et al., 1991). Kawamura et al. (1991) evaluated 17 healthy young men (mean age = 25.3), and found significance only between the greatest and least incline angles of 12 and 3 degrees (p <0.01).

McIntosh et al., (2005) noted initial increases in stride length on inclines of 0°, 5°, 8° and then plateauing at the 10° inclination. Step length distances increased significantly on inclines by an average of 68cm on level ground to 70cm and 71cm on 6° and 9° slopes respectively (Kawamura et al., 1991). Sun et al., (1996) showed significant decreases in uphill step length versus downhill on the same slope angles across all three age groups studied (10-35, 35-55 and 55-75) (p < 0.001). Researchers in this study speculated that the differences in their findings are a result of the varying angles of the dock's ramp as the ebb and flow of the tide changed or as the result of the observational techniques used by the researchers.

Results of the Sun et al. study (1996) were based on 2,400 observations of pedestrians in three different estimated age groups (10-35, 35-55 and 55-75) while they walked over ramps whose angles changed frequently with the ocean's tide. Data were collected over an eight-week period at various times in the day. During these times researchers continually measured various ramp angles as the tide changed. Step length, cadence and walking speed were recorded and

compared between groups and between ramp angles. The largest threat to validity derived from the observational data collection methods used in this study. The most significant results indicated that walking speed was significantly affected by age ($p < 0.001$) while step length and velocity of the older group (55-75) were more affected by downhill slopes ($p < 0.005$) (Sun et al., 1996).

In addition, Sun et al., (1996) found that subjects' preferred walking velocity decreased with increasing grades of both inclines and declines and concluded that the angle of slope has significant effects on velocity. These results support the findings of Kawamura et al. (1991) who used similar incline grades. However, decreased velocity differed from the results shown in a later study using varying inclines from -10° (downhill) to 10° (uphill) (McIntosh et al., 2005). Regardless of whether subjects walked on declines or inclines walking speed in eleven healthy subjects (mean age = 22.4) increased as the grade of slope increased. McIntosh et al., (2005) provided explanations for the differences in results citing a shorter walkway that did not allow subjects to attain their preferred gait velocity before recording began. The authors also suggest that increase in gait speed might be attributed to a younger and healthier population who did not perceive the walkway as hazardous.

Despite significant evidence to suggest changes in spatiotemporal gait parameters of younger adults on inclines, little data exists on older adults walking patterns on inclines. Efforts have centered on establishing normative data on incline gait patterns of younger adults walking on inclines using level ground walking as a baseline to compare results (Tokuhiro et al., 1995; Sun et al., 1996;

Kawamura et al., 1996; Lange & Hintermeister, 1996; Leroux et al., 1999; McIntosh et al., 2005; Lay et al., 2006, 2007). Sun et al., (1996) were the only researchers who provided data on a population with a mean age over fifty. However, results of this study should be viewed with some caution due to the observational, field study design.

Evidence suggests different walking patterns when young healthy adults walk on inclines when compared to level ground walking. Various data formats including kinetic, kinematic and spatiotemporal data suggest that differences are present when comparing level ground walking between younger and older adults. Many reasons for these differences have been proposed, including age related physiological declines in strength, range of motion and balance as well as a shift in older adults' motor planning to a more stable gait pattern. Similar to strength and range of motion during gait, balance has been documented extensively in the literature but is not totally understood as it relates to gait.

Gait deviations and balance

Evidence from studies with younger subjects shows clear differences in gait patterns on incline surfaces when data were compared to level ground walking. These adaptations are attributed to the overall increase in work requirements from various muscles and challenges to overall balance. These challenges to balance originate from changes in the relationship between the body's center of mass and position of subject's lower limbs (Lange & Hintermeister, 1996; Leroux et al., 1999). More specifically, there is a shift in the center of mass of the body anteriorly with inclines and slightly posterior with

declines to offset the effects of the sloped surface (Leroux et al., 1999). As this shift occurs the lower extremities are forced to adapt in order to maintain the most efficient gait pattern while continuing in a forward progression. Although no direct correlation between incline gait and balance has been established, a connection can be inferred from the significant findings documenting spatial and temporal changes in base of support (step width) (Woollacott, Shumway-Cook & Nashner, 1986). Base of support and percentage of stance time (single or double) are two variables often used as measures of stability when studying gait and balance (Brach et al., 2007). Both variables have been shown to change to maximize stability when walking on inclines or varying surfaces. Slight increases in step width or base of support on inclines were noted in an attempt to maintain subjects' center of mass, which demonstrates more lateral and forward deviation, over their base of support. The head, neck and trunk were found to be least stable while walking on inclines of greater slope angles (Cromwell, 2002). In addition, percentage of double support typically increases with age during level ground walking as adults shift their focus to a more stable pattern with decreased step lengths and velocity evident as a result (Kemoun, et al., 2002; Cromwell & Newton, 2004).

Further, variations in base of support and percentage of double support have been implicated as possible predictors of falls in older adults (Woollacott et al., 1986). Kemoun et al. (2002) demonstrated that increased double support time was significantly different in fallers and non-fallers in older adults (p = 0.024). In addition, when compared to younger populations, older adults

demonstrate a greater percentage of time in double support, an increased step width and a decrease in overall gait speed (Kemoun, et al., 2002). Pavol et al. (1999) suggest that these changes in otherwise healthy older adults develop as a balance strategy to compensate for age-related physiological changes.

Different balance strategies and an increased time latency of muscle responses are also more pronounced in older adults and different from those in younger adults. Older adults have earlier responses in their hip musculature before activation of ankle musculature during balance perturbations. This pattern is inverted in younger subjects who initially seek postural control through ankle strategies (Woollacott et al., 1986). Woollacott et al. (1986) also suggest that this deterioration of postural control is one reason for postural modifications in older adults. Often times it is this decline in postural control and variations in gait in older adults that lead to significant functional deficits. Although not established in the literature, intuitively it seems this decreased postural control in static postures and level ground scenarios may be more prevalent on incline surfaces due to the increased deviation of the body's center of pressure (Kawamura, Tokuhiro & Takechi, 1991).

To quantify balance during gait in older adults, Cromwell, Newton and Forrest (2001) developed a more sensitive technique to measure stability in terms of the amount of steps taken per unit of distance. The Gait-Stability Ratio (GSR) indicates a measure of walking stability and is measured as cadence divided by velocity. The higher GSR the more steps taken per unit of distance and the more unstable the gait pattern (Cromwell & Newton, 2004). The GSR

also provides a mechanism of normalizing cadence with respect to velocity (Cromwell, Newton, 2004). This measure has been shown to be a more sensitive balance measure than traditional techniques such as velocity and cadence measured alone. Therefore, the GSR is optimal when assessing balance in healthy adults as compared to other tests like the BBS which demonstrates a ceiling effect on higher functioning adults. Ultimately, the combination of different balance assessments provides a more accurate perspective on a person's balance as it relates to function and gait.

Dickerson and Fisher (1993) suggested that as people age they experience declines in function and activities of daily living (ADLs). Walking, bathing, dressing and moving from one chair to another are all activities that are often difficult for older adults to perform and determine an individual's level of independence (Shephard, 1990). The connection between balance or postural control and functional limitations in older adults is often measured by objective tests, with some of the more commonly used and practical tools such as the Berg Balance Scale (BBS) and Dynamic Gait Index (DGI). Both of these tests have shown sensitivity in identifying fall risk in community-dwelling older adults (Shumway-Cook, Brauer & Woollacott, 2000).

The BBS is well established in the literature as a valid predictor of falls (Berg, Wood-Dauphinee & Gayton, 1989; Berg, Wood-Dauphinee & Makin, 1992). It is a comprehensive test that evaluates fourteen different tasks including sitting balance, standing with gradual decreases in base of support, turning, reaching out of base of support and retrieving objects from the floor (Berg et al.,

1992). Based on the level of assistance required, a score from 0-4 in each category is given for a total possible score of 56. Supported by clinical experience and extensive use in research settings, Berg and colleagues (1989; 1992) established a cut off score of 45. Those that score at least 45 are deemed at less risk for falls while those that score less than 45 require further investigation and assessment for appropriate assistive devices. The BBS has also demonstrated high reliability and validity in evaluating older adults' level of function (Berg, 1992). Although highly reliable and useful, the BBS has limitations when assessing more dynamic aspects of gait in high functioning adults. All of the categories in the BBS are designed to assess static balance tests and everyday tasks such as sit to stand and picking up objects from the floor.

Strength deficits in the lower extremity, as seen in older adults, have also been linked to postural control and predictors of falls (Daubney & Culham, 1999). Decreased forces in dorsiflexion and eversion strength were found to be the only conclusive predictor of falls. One reason given for this relationship is the BBS includes several measures where stability needs to be maintained by the subject for extended periods of time (Daubney & Culham, 1999). Similarly, ankle strength was implicated in identifying older adult fallers and non-fallers when compared to knee strength. Further analysis revealed the greatest difference between fallers and non-fallers in the ankle muscles (Whipple, Wolffson & Amerman, 1987). These findings contribute to the body of evidence regarding age-related factors

such as decreased strength and subsequent decreased safety and independence.

The dynamic nature of gait and the variability of the different surfaces require an objective test that is designed to test such aspects of functional balance. The Dynamic Gait Index (DGI) is used extensively as a method to evaluate and document a patient's ability to modify gait in response to changing task demands in ambulatory patients with balance impairments (Shumway-Cook et al., 1997). It has been particularly effective at predicting falls in patients with vestibular dysfunction while also being used as a more dynamic gait assessment for community-living older adults. The DGI has shown high intra-and inter-rater reliability (0.76-0.98 and 0.98 respectively) while being validated during concurrent testing with the BBS and Timed Up and Go (TUG) (Whitney, Wrisley & Furman, 2003).

For the purposes of this study, the DGI may be more informative because of multiple items requiring attention-splitting task performance while walking. Tasks included on the DGI include changing speeds, turning, walking around objects and walking with head turns. These attention-splitting tasks may be more appropriate for an independent, older adult population due to the increased demands placed on multiple systems of the body such as visual, vestibular and cognitive centers. Another reason for the inclusion of the DGI is that most of the components on the test require the subject to modify their gait patterns as they walk (Marchetti & Whitney, 2006). Due to the many dynamic components of the DGI overall results are more generalizable to independent older adult populations

functioning at higher levels than the general population. Such dynamic components are not included in the BBS which limits its applicability to high functioning individuals (Cromwell & Newton, 2008). Therefore, by using clinical tools such as the BBS and DGI it is possible to establish correlations between function and balance related to walking on incline surfaces.

Objective tests are often helpful in objectifying change as well as providing clinicians and researchers with baseline information on overall function.   Often this level of function is associated with the ability to walk independently in order to complete daily tasks. As part of older adults' community interaction and independence, they often encounter more challenging surfaces and terrains which may be difficult to safely negotiate. Age related changes such as decreased strength, range of motion and propioception make it difficult to quickly adapt to alternating terrains. In addition, maintenance and recovery of balance is accomplished differently in older adults when compared to younger adults and resultant gait patterns in the elderly differ from those seen in younger populations (Hsiao-Wecksler & Robinovitch, 2007). For example, older adults exhibit greater double support time, increased step width (wider base of support) and decreased velocity on level surfaces when compared to younger subjects (Kemoun, 2002). Presumably, these variations are due to the numerous physiological and psychological (fear of falling) factors that occur with aging. Variability in these same gait parameters have been linked to factors that lead to falls in older adults (Woollacott, 1996).

In the next decade the greatest percentage of Americans will be 65 years and older. There is evidence that these older adults will be more active and generally more health conscious than ever before. Improvements in medications, the possibility of extending the retirement age and improved nutritional awareness are some of the factors that may contribute to older adults being more actively involved in society than ever before.

Research has demonstrated that with increasing age various idiopathic, physiological changes occur that alter the manner in which older adults walk and complete tasks throughout the day. Some of the more measurable physiological changes include decreased strength, range of motion and balance. As a result of these physiological changes many older adults become less independent with walking and overall daily function. Ultimately, the gait pattern of older adults becomes slower, more variable and less efficient. Therefore, with a less efficient gait pattern and slower balance reactions the frequency of falls older adults increases. Currently, much of the gait research on older adults has been collected while walking on level ground in a controlled environment. However, level ground walking is not the only surface that older adults encounter in the community and results from these studies cannot be generalized to walking on alternate surfaces such as inclines. These studies also do not address the increased balance requirements often required while walking on different surfaces or walking while performing other tasks. These are aspects of gait, particularly in older adults that may make them more susceptible to loss of balance and falls.

Research shows that older adults demonstrate different gait patterns on level ground when compared to younger subjects. The most common difference seen in older adults' patterns are decreased step lengths, increased cadence, increased bases of support, and decreased gait velocity. There is also evidence to suggest that the timing and amplitude of muscle contractions, as captured by EMG, varies in older adults. Kinematically, older adults demonstrate different trunk angles, decreased pelvic rotation and less dorsiflexion during level ground walking. The exact cause of the changes is still debated but most researchers agree that it is a multitude of factors that contribute to distinct older adults' gait pattern changes.

Walking on inclines is another scenario that induces distinctive gait pattern changes. More specifically, the gait patterns of healthy, young adults on incline and decline surfaces relative to level ground must be looked at more closely. When comparing incline walking to level walking the demands imposed on the human body are different and require various adaptations that have been outlined in the literature and summarized in this paper. Unfortunately, to date virtually no data have been compiled on older adults walking on inclines. It is logical to assume while walking on inclines resultant patterns adapted by older adults will be different than those seen in younger populations due to age-related physiological changes. Given the age related changes that occur in older adults, in order to maintain forward progression, maximize safety and assume an overall efficient gait pattern on inclines, a variety of adaptations occur. Further, these

adaptations will more than likely be different than the strategies used by younger adults.

Despite research frequently including aspects of strength and balance, the exact contribution of each is still debated. It is imperative to begin to narrow some of the gaps in the literature with regards to older adult walking patterns on inclines. During this process balance and strength correlations will be included to provide answers on why changes in spatiotemporal aspects of gait occur. Since research has not yet provided any normative data on older adults walking on inclines the information gained from this study will provide baseline data for future research. The onus is on clinicians and researchers to delve deeper into the walking patterns of older adults, particularly on surfaces other than level ground. Knowledge gained from such studies will assist in developing future clinical interventions and ultimately decrease the risk for falls in older adults.

CHAPTER III

Materials and Methods

Subjects

Before subject selection and data collection began permission and site approval from each community's board of directors were obtained (Appendix A). In addition, approval from the Seton Hall University Institutional Review Board (SHU-IRB) was also obtained before subject selection began.

Twenty seven healthy older adults $\geq$ 70 years old without disabilities or musculoskeletal impairments were recruited for this study. A sample of convenience from several local adult communities, community centers and hospitals in central New Jersey was recruited using flyer postings in shared locations throughout the community. The adult communities targeted in this study are 55 and older communities consisting of single family homes with various ethnic and religious backgrounds.

Inclusion criteria:

To be included in this study subjects were at least 70 years old. In addition, the University of Alabama Birmingham Life-Space Assessment Form (UAB-LSAF) (Appendix A) was used to screen subjects to determine level of independence before arrival at the data collection site. A minimum self-report score of a 32 on level four of the UAB-LSAF was required to be included in the study. After all testing protocols were described to each subject, those who met

the inclusion criteria were asked to complete a demographic information sheet

and sign an informed consent approved by the SHU-IRB.

Exclusion criteria:

Subjects were excluded if they: (1) were diagnosed with any neurological

or orthopedic condition that alters the normal observable gait sequence (2)

required assistance from another person or device during ambulation (3)

reported any visual or vestibular dysfunction that compromised balance during

ambulation.

All subjects who qualified for the study were assigned an alphanumeric

code before beginning data collection to maintain anonymity. All data collected

during the study were saved on a portable disk drive and stored securely at the

primary researcher's residence. Data were also stored on a designated laptop

computer with an encrypted password system which remained in a secure

location at Seton Hall University Graduate Health Science offices in Alfieri Hall.

Design and variables:

The study used a within-subjects repeated measures design with subjects

acting as their own controls. The independent variable used in this study was the

walking surface with two levels (1) level ground and (2) incline. The dependent

variables were spatiotemporal gait parameters including cadence, step length

and mean normalized gait velocity as measured by the GAITRite electronic

walkway system. In addition Berg Balance Scale (BBS), Dynamic Gait Index

(DGI) scores and a Gait-Stability Ratio (GSR) were calculated as objective

measures of function and balance.

Measurements:

During data collection the following measurements were included: demographic data, community independence level (University of Alabama-Birmingham Life-Space Assessment Form), leg length (cm), active ankle dorsiflexion and plantarflexion range of motion, ankle dorsiflexion, plantarflexion and hip extension muscle testing, functional mobility (BBS), dynamic balance (DGI) and GSR.

Demographic data:

All subjects were required to independently complete a demographic data sheet which provided age, gender, date of participation, location, comorbidities, and fall history (Appendix C). Completion of this data sheet served as a screening tool to confirm that the participant had adequate cognition and awareness to be included in this study.

Community independence level:

In order to confirm each subject's self-reported community independence level, and thus the ability to meet inclusion criteria, the University of Alabama-Birmingham Life-Space Assessment Form (UAB-LSAF) was used. The design of the questionnaire is intended to assess a person's "pattern of mobility in the prior month" among community-dwelling older adults (Baker, Bodner & Allman, 2003). Unlike other functional and physical measures that assess what subjects are able to do, the UAB-LSAF assesses what people actually do. The UAB-LSAF was originally designed to include one interviewer asking questions that evaluate how

frequently and in what capacity community-dwellers interact with their community.

Conceptually the form was designed using five concentric circles including different "life-spaces" beginning at the center with the most limited area: (1) other rooms other than the room you sleep in (2) areas outside your home but on your property including porch, driveway or hallway of apartment building (3) places in your neighborhood other than your house or apartment building (4) places outside your neighborhood but within your town and (5) places outside your town. In addition to the person's community mobility patterns the frequency and assistance required in those locations are also included. A value is given for life-space level and a total sum for all life-space levels is also obtained. For the purposes of this study inclusion requirements are that each participant must achieve a minimum score of 8 at life-space level four. A community-dweller that attains this score is described as being able to walk "outside their neighborhood but within their own town 4-6 times per week without assistance" (Peel et al., 2005).

UAB-LSAF has high test-retest reliability with Intraclass Correlation Coefficient (ICC) of 0.96 comparing in-home interview with a two week follow-up phone interview (Baker, Bodner and Allman, 2003). In addition, high correlations to physical performance and function measures were found between the UAB-LSAF and Short Physical Performance Battery, Instrumental Activities of Daily Living and Activities of Daily Living (Peel et al., 2005).

Strength testing:

Lower extremity strength of three different muscles was included as a covariate measurement. Based on evidence from electromyography studies hip extension (particularly the gluteus maximus), dorsiflexion and plantarflexion are all implicated as the primary power generators during incline gait, strength data, using a standardized hand-held dynamometer were collected (Lay, Hass, Nichols and Gregor, 2007). Each subject was positioned to maximally isolate and test each muscle group isometrically as described by previous literature (Bohannon, 1989; Ford-Smith, Wyman, Elswick and Fernandez, 2001).

Intrasession reliability of the hand-held dynamometer was found to be high ranging from 0.93 to 0.98 when testing was performed on various paretic and non-paretic muscles of the lower extremity (Bohannon, 1989). Rose et al., (2008) also found high intrarater reliability in all muscle groups around the ankle in young children with ICC = 0.94. In community-dwelling older adults hand held dynamometry also showed high ICCs ranging from 0.76 to 0.90 for individual lower limb scores (Ford-Smith et al., 2001). Overall this method of strength testing limits the subjectivity that is often associated with standard manual muscle testing particularly at the higher grades (> 4/5) (Knepler & Bohannon, 1998).

Ankle dorsiflexion range of motion (ROM):

Ankle dorsiflexion ROM was taken as a covariate measurement using the standardized testing positions as described by Norkin and White (2003). Perry (1991) described 10 degrees of dorsiflexion as the minimal requirement to walk

on level surfaces without compensatory motions incurred in the lower extremity.

On incline surfaces as much as 25-30 degrees of motion is required to walk uphill

in order to accommodate for the increase in slope angle (Leroux, Fung &

Barbeau, 2002).

Joint measurements using universal goniometers were shown to have

high intrarater reliability (ICC 0.92) when measuring active dorsiflexion with the

knee extended (Clapper & Wolf, 1988). Boone et al. (1977) found that intratester

reliability for selected motions around the ankle were higher than those found for

hip and wrist motions but not as high shoulder, elbow and knee motions. The

position of the knee did not have a significant effect on resultant active

dorsiflexion measurements among 27 subjects (mean age 26.1). Repeated

measures of ankle dorsiflexion with the knee flexed to 90 degrees ICCs were

0.97 while with the knee extended ICCs were 0.98 (McPoil, Cornwell & Wolfe,

1996).

To further explain possible changes in incline gait parameters, concurrent

tests of balance and functional mobility were conducted in the study. Functional

mobility was assessed using the Berg Balance Scale (BBS) (Appendix E), a 14-

item functional mobility scale originally designed to assess fall risk in older adults

was completed (Berg, et al., 1989). In order to ensure that each subject

understood each task, both verbal and visual instructions were given before

beginning each item as described by Berg, Wood-Dauphinee & Gayton, (1989).

Items on the BBS range from simple mobility tasks (i.e. transfers, standing

unsupported) to more difficult tasks (i.e. tandem stance and turning 360°). One

researcher provided supervision while providing assistance only when necessary during completion of the BBS as outlined by the developers of the test. Each item on the BBS is progressively more difficult and is scored from 0 to 4 on a Likert scale format. A zero represents an inability to perform a task while a score of four is achieved when the subject completes the task without assistance or compensatory strategies. Based on a maximum score of 56 research has shown that scores less than 45 are at greater risk for falls (Berg, Wood-Dauphinee & Gayton, 1989; Berg et al., 1992).

BBS has high interrater, intrarater and test-retest reliability with ICC of 0.98 for all measures of consistency (Berg, Wood-Dauphinee, Williams & Gayton, 1989). It also has been shown to have moderate to high concurrent validity when compared to other functional measurement tests such as Fugyl-Meyer, Dynamic Gait index, Timed Up and Go and the Tinnetti Balance Scale (Berg, Wood-Dauphinee, Williams & Gayton, 1989).

Dynamic balance was assessed by the Dynamic Gait Index (DGI) (Appendix F). The DGI was developed to document a patient's ability to modify gait in response to changing task demands in ambulatory patients with balance impairments (Shumway-Cook et al., 1997). The DGI tests eight different items ranging from level ground walking to walking with head turns (both vertical and horizontal), turning and stair negotiation. Each item is scored from 0 to 3 on a Likert scale based on the level of impairment while completing the task. A maximum score is a 24 with scores under 19 indicative of increased fall risk (Shumway-Cook et al., 1997).

McConvey and Bennett (2005) examined the reliability and validity of the total DGI and reported good interrater reliability (ICC for 0.98). Intrarater reliability ranged from 0.76 to 0.99 during that same study. The DGI compares favorably to the BBS with moderate but significant correlations (r = 0.71; p <.01) establishing its concurrent validity (Whitney, Wrisley & Furman, 2003).

A gait-stability ratio (GSR) was calculated indirectly by using values of cadence (steps/second) and dividing it by gait velocity (meters/second) to measure the changes in walking velocity and step length (Cromwell, Newton, Grisso & Edwards, 2001). The resultant unit of measurement for GSR is steps/meter and is found to be a more sensitive measure of dynamic balance than either cadence or walking velocity alone and describes individual's ability to adapt to balance changes (Rogers, Cromwell & Grady, 2008). An increase in GSR represents an increased number of steps taken per unit of distance indicating less stability during an activity. In addition, the GSR also provides a mechanism of normalizing cadence with respect to velocity (Cromwell & Newton, 2004). For this study the required numerical values to calculate GSR were obtained directly from the GAITRite software.

Correlations were found to exist between GSR and the more dynamic items on the BBS, specifically items 12, 13 and 14. Item 12 was found to have a strong inverse relationship with GSR calculations (r^2 = -0.54). Specifically, those who had fewer steps on to the stool during Item 12 of the BBS had higher GSR indexes. Based on these results, researchers suggested that the weightshifting and alternate leg movement required in Item 12 strongly correlates to dynamic

activities such as gait (Cromwell and Newton, 2004). The inclusion of this ratio

for higher functioning ambulators is a simple and easy way to measure stability of

balance during level and incline gait (Cromwell & Newton, 2004).

Procedures

Subjects who provided their contact information were pre-screened via

telephone call using the UAB-LSAF to ensure that they met all inclusion criteria.

If the subject qualified for the study, a location, time and date were given to the

subject to complete the battery of tests and gait trials. Data collection was

performed in one-40 minute session for each subject from July 2009 to October

2009 in a central location on the grounds of the local adult communities and

hospitals. Upon arrival subjects were asked to fill out a demographic information

sheet and provided consent for inclusion in the study (Appendix). If this

information was completed accurately and independently researchers concluded

that the subject's cognitive status was adequate for inclusion in the study.

Bilateral ankle dorsiflexion and plantarflexion and hip extension isometric

strength measurements were then taken using the Lafayette Manual Muscle Test

system model 01163™ (Appendix D) using standardized test methods as

described by Damiano and Abel (1998). To measure dorsiflexion the subject was

asked to lie supine with the knee extended with the ankle and foot in neutral and

the lower limb was stabilized. To ensure that the proper motion was performed

the researcher moved the ankle passively through the available dorsiflexion

range of motion. Verbal commands were given to the subjects to "pull their toes

up towards the ceiling as hard as they can on the command: ready, go". With

only the dynamometer's strain gauge pad in contact with the subject, resistance

was given on the dorsum of the foot at the level of the metatarsal heads.

Subjects performed an isometric contraction gradually increasing their force over

a three second time frame as described by Bohannon (1989). Approximately 15-

second rest breaks were given in between trials as the gauge was reset. In order

to increase intratester reliability and decrease fatigue, the examiner positioned

himself to maximize stability while maintaining proper body mechanics. Three

trials were performed and the average of the three trials was included in the data

for analysis.

To assess plantarflexion strength the subject remained in the supine

position with the foot and ankle in neutral and knee extended (appendix). To

ensure that the proper motion was performed the researcher moved the ankle

passively through the available plantarflexion range of motion. Instructions were

given to each subject to "push down towards the floor as hard as they can on the

command: ready, go." Subjects performed an isometric contraction gradually

increasing their force output over a three second time frame as described by

Bohannon (1989). Resistance was provided on the plantar aspect of the foot

directly on the metatarsal heads. Approximately 30-second rest breaks were

given in between trials as the gauge was reset. Three trials were performed and

the average of the trials was included in the data for analysis.

Hip extension strength was measured in prone with the hip extended and

knee flexed to isolate the gluteus maximus' contribution to hip extension

(appendix). This position was described by Taylor, Dodd and Graham (2004) and

is thought to be the most advantageous position for the researcher while isolating hip extensor strength. To ensure that the proper motion was performed the researcher moved the ankle passively through the available plantarflexion range of motion. Placement of the force pad was at the distal thigh just proximal to the popliteal fossa. Verbal instructions were given to the subject to "push upwards towards the ceiling as hard as they can on the command ready, go". The subject was asked to hold the contraction for three seconds to allow for appropriate muscle recruitment (Bohannon, 1989; Taylor, Dodd and Graham, 2004). Three trials were performed and the average of the trials was included for data analysis.

If a subject was unable to lie prone an alternate position in supine was used as described by (Ford-Smith et al., 2001) to measure hip extension strength. While supine the subject was positioned with their hip and knee flexed to 90 degrees. To ensure that the proper motion was performed the researcher moved the ankle passively through the available hip extension range of motion. The force pad was placed just proximal to the popliteal space at the distal hamstrings. Verbal instructions were given to the subject to push into the examiner's force "as if they wanted to lower their leg back down to the table" as hard as they can on the command: ready, go". As described in previous literature for each trial the subject was told to hold the contraction gradually increasing their force over a three second timeframe to allow for appropriate muscle recruitment (Bohannon, 1989; Taylor, Dodd and Graham, 2004). Rest breaks during strength testing were given between each trial and were the length of time it took the researcher to record the digital output and reset the

dynamometer to zero. Three trials were performed and the average of the trials was included for data analysis.

Active ankle dorsiflexion ROM was then performed by the subject and measured by the same researcher. Each subject was positioned in sitting with the knee flexed to approximately 90 degrees described by Norkin and White (2003) as the standard goniometric testing position for ankle dorsiflexion (appendix). Anatomical landmarks included the head of the fibula for the stationary arm of the goniometer, the lateral malleolus for the fulcrum while the movement arm of the goniometer was aligned parallel to the fifth metatarsal shaft of the foot. The subject was shown the desired motion passively by the investigator and then asked to raise their foot up without extending their knee while active dorsiflexion was measured and recorded. The average of three active measurements was used for data analysis as the subject's dorsiflexion range of motion.

While standing with comfortable footwear donned (walking shoe or sneaker without a heel or lift), bilateral leg length measurements (centimeters) were taken using a steel tape measure and recorded by the same researcher using the most superficial aspect of the greater trochanter as the proximal anatomical landmark measuring vertically to the floor bisecting the lateral malleolus as described by Cutlip et al., (2000). Data from the leg length measurements were then entered into the GAITRite software before gait trials began.

The Berg Balance Scale was then administered to each subject following the protocol established by Berg, Wood-Dauphinee, Williams and Gayton (1989). Following a one minute rest break subjects were then asked to complete the Dynamic Gait Index using the established protocol established in previous literature (Shumway-Cook et al., 1997). To increase intrarater reliability a copy of each standardized test (BBS and DGI) was used as described in previous literature by test designers.

The final component consisted of five walking trials each on level and incline surfaces. The initial subject was randomly assigned the walking surface to begin their walking trials by choosing from index cards labeled "I" for incline and "L" for level surface. After that selection subsequent subjects would start on the walking surface that the previous subject finished walking on. This method minimized the excessive movement of the GAITRite mat while guaranteeing the same amount of trials beginning on each surface. Ultimately, by randomizing the surface where each subject began practice and fatigue effects were controlled for during the gait trials. One practice trial on each walking surface was given to each subject in order to accommodate to the walking surface and angle of incline. Subjects were instructed to begin at a predetermined marker placed 2 meters before the electronic walkway. A second GAITRite was placed before each walking surface to determine if any preparatory stepping patterns existed prior to encountering the second GAITRite on the level and incline surfaces. Data collected from this GAITRite provided information on any preparatory patterns that developed as the subject approached the incline.

Before ambulating on either walking surface, verbal instructions remained consistent as subjects were asked to walk at a "comfortable" pace over the entire electronic walkway. Participants were asked to begin walking when they heard the command "start" and to continue until they heard the command "stop". This method of instruction ensured that each subject walked past the end of the GAITRite mat walkway before terminating each of their trials.

Placement of tape 2 meters before and after the electronic walkway served as starting and ending points respectively ensuring consistency between subjects and establishing a constant gait speed while the data were recorded (Cutlip, et al., 2000; Grabiner, et al., 2001). After each trial a 30-second rest period was given to the subjects while the computer was reset and a 1-minute rest period between each surface was provided to minimize potential fatigue and learning effects.

During the incline trials, as with level gait, each subject started 2 meters before recording began and was asked to walk at a "comfortable" self-selected pace over the electronic walkway and continue past the end of the mat towards a piece of tape placed on a 5 x 5 turn platform beyond the GAITRite recording surface. During the incline trials, the GAITRite mat was positioned approximately one meter after the incline began in an attempt to avoid recording the initial accommodating footfalls on the incline. This location was marked to ensure the same position of the GAITRite relative to the incline for all trials. As with the level gait trials, 30-second rest periods were given to the subjects between trials and a

one minute rest period was given to each subject after the series of five gait trials.

To increase inter-trial consistency and maximize safety handrails were provided the entire length of GAITRite during level and incline walking as well as the "turn platform". However, regardless of walking surface, subjects were asked not to use the handrails unless they perceived a loss of balance or unsteadiness. If the subjects grasped the handrail or disrupted a walking trial for any reason data from that particular trial was not used in the analysis and the trial was performed again. Additional safety measures were taken as the primary researcher walked alongside and slightly behind each subject while being careful not to come in contact with the GAITRite.

Instrumentation:

GAITRite computerized walkway

Spatial (distance) and temporal (time) parameters of gait were measured by GAITRite computerized walkway system. The GAITRite electronic walkway system is a pressure sensitive mat measuring 4 meters in length. An extensive series of imbedded pressure-sensitive sensors organized in a 48 x 288 grid pattern located between two layers of vinyl. The active recording area is 61 cm x 3.66 m, with 12.7 mm spaces in between adjacent switches (Cir systems, 2001). The mat is connected via serial port to a personal IBM computer using GAITRite Gold software running on a Windows 98 operating system. Gait trial data is captured at a sampling rate of 80 Hz. As the subject walks across the walkway, the system captures the geometry and relative arrangement of each footfall as a

measure of time. The application software controls the functionality of the walkway, processes the raw data into footfall patterns, and computes the temporal and spatial parameters of gait (GAITRite Systems, Inc., 2001). The resultant information is electronically stored in the software's data files.

The GAITRite electronic walkway is the gold standard when collecting spatiotemporal parameters of gait and has high test-retest reliability and high concurrent validity. ICC = 0.95 for spatial correlation between paper and pencil and the GAITRite (McDonough et al., 2001). Individual parameters measured by the GAITRite have also demonstrated high test-retest reliability between 0.82 and 0.92 (Menz, Tiedemann, Mun Sun Kwan & Lord, 2004). Cutlip et al., (2000) showed correlations between the GAITRite and video-based system of > 0.94. Concurrent validity of the system is also high (ICC = 0.99) when compared to another common gait analysis tool, the in-shoe Clinical Stride Analyzer (Bilney, Morris & Webster, 2003).

Modular ramp

During incline data collection subjects were asked to walk on a reinforced, modular ramp constructed out of commercial grade aluminum (Express Ramps Inc., 2005) (Appendix G). This portable ramp is commercially available and all components of this ramp reached or exceeded all American Disabilities Act (ADA) safety standards for wheelchair ramps. The ramp is designed for external and internal use and has been weight tested up to 850 pounds. Using explicit directions provided by Express Ramps, Inc. (2005), construction of the ramp resulted in a length of 18 feet and a width of 3 feet. An additional 5 x 5 level, foot

"turn platform" was added to allow subjects to comfortably turn around and

ensure a constant gait speed is maintained while on the ramp. Additional safety

considerations included an extruded, skid resistant surface and standard

handrails along the length of both sides of the ramp and turn platform.

In compliance with the ADA and the Occupational Health and Safety

Association all ramps must comply with a standard rise to run ratio. Standard

ramp ratios for business sectors are 1:12 while residential requirements are 3:12.

Expressed more simply, using the business parameters, one foot of height

requires 12 inches of run or slope length.

For this study, a 2:12 ratio was chosen to comply with the ADA standards

while increasing the angle to approximately 9.5 ° which is the approximate angle

that has been documented in previous incline gait studies to be the critical

threshold where gait parameters change. (Kawamura, Tokuhiro & Takechi, 1991;

McIntosh, Beatty, Dwan & Vickers, 2005). The 18-foot ramp that was

constructed was long enough to fulfill ratio requirements in order to rise to a

height of three feet. This length also accommodated for the entire length of the

GAITRite while maintaining a realistic angle that is encountered during

community ambulation.

Hand-held dynamometer

The Lafayette Manual Muscle Test system (model 01163)™ (Appendix D)

was used to gather objective data on force output of various muscle groups in the

lower extremity. This hand held device was designed to objectively quantify

isolated peak force muscle strength throughout the body. It measures peak

force, time to reach peak force and total test time. This system measures the

force produced when a muscle contracts by using the muscle to cause a force

against the force pad (Leavey, 2006). The system has the ability for immediate

digital readouts in pounds or kilograms and on-board data storage for up to 52

tests. The Lafayette system has the ability to measure up to 300 pounds of force

over a maximum of ten-second time frames. System features include portability

and versatility with the ability to comfortably fit in the examiner's hand. In

addition, the unit has adaptable stirrups to conform to various contours of the

body part being tested to maximize comfort and accuracy.

Data analysis

G*power software provided the software for calculations of the appropriate

sample size and subsequent power analysis (www.psycho.uni-

duesseldorf.de/aap/projects/gpower). It was determined that a minimum of 27

subjects was necessary to demonstrate significance with a 0.05 alpha

designation and medium effect size (.50 as defined by Cohen).

To control for extraneous variables, each group of subjects served as its

own control incorporating a repeated measures (within subjects) design for this

study. Data analysis was performed using The Statistical Package for the Social

Sciences (SPSS) software, version 16.0 for Windows (Narusis, 2007). For all

gait parameters a mean score was calculated across each of the five trials for

both level and incline surfaces. In order to compare differences between each

subject, paired t-tests were used to analyze means across conditions for the first

four hypotheses posed. This statistical test ensures that subjects are compared

only with themselves. Statistically, this reduces total error variance because the extraneous factors are the same across both treatment conditions (level and incline gait) (Portney & Watkins, 2000). A repeated measures design is the most appropriate statistical test for comparing differences when the same group is studied under two conditions (Portney & Watkins, 2003; Salkind, 2006).

A repeated measures analysis of variance (ANOVA) was run to determine the significance of the differences between mean GSR scores on level and incline surfaces (Hypotheses 5 & 6). This is the most appropriate and robust test when there is a small sample size with the same group is being tested under two different conditions (Portney & Watkins, 2003; Salkind, 2006; Field, 2009). Under such conditions when there is one independent variable being tested with two levels sphericity is assumed therefore, Mauchly's test for sphericity is not applicable.

CHAPTER IV

Results

Subjects and Demographics

Seventy-eight older adults participated in this study. Participant

demographics, UAB-LSAF scores, ROM and strength measures are presented

for all of the subjects in Table 1. All of the participants in this study were

recruited from four different and diverse older adult residential communities,

senior centers and assisted living facilities in central New Jersey.

The mean age of the sample population for this study was 77.8 $\pm$ 4.8,

ranging from 70-92 years. Mean age for males was 79.2 $\pm$ 5.6 while the mean

age for females in this study was 77.2 $\pm$ 4.5 (p = 0.08). Consistent with the

United States census data for this age range, females outnumbered males in this

study 52 (67%) to 26 (33%). Gender ratios nationwide for this age group are 82

males for every 100 females from 65-74 dropping to 42 for every 100 females

after the age of 85 (US Census Bureau, 2000). Factors that may contribute to the

gender difference in this population of older adults are a longer life expectancy

for females (79 to 72) in this country and an increased percentage of widows

(45%) than widowers (14%) after the age of 75 (US Census Bureau, 2000).

All of the subjects included in this study were healthy, active older adults.

Each subject was screened by the primary investigator to ensure that all subjects

scored a minimum level of 16 on level four of the University of Alabama-

Birmingham Life-Space Assessment Form (UAB-LSAF). This minimum score

indicated that subjects were able to ambulate independently outside their own neighborhood, but within their own town at least 1-3 times per week without aid from another person or assistive device (Baker, Bodner & Allman, 2003). Above this minimum level, analysis of UAB-LSAF scores, revealed similar differences between males (x = 24.9, SD ± 6.5) and females (x = 25.6, SD ± 5.4).

Functional balance scores were also collected for the Berg Balance Scale (BBS) and Dynamic Gait Index (DGI) and are presented in Table 2. Mean BBS score for the entire sample was 50 (± 3.5) while mean DGI score was 20 (± 2.1). On average males scored slightly lower on the BBS (49, ± 3.7) than females (51, ±3.3) while both males and females scored the same on the DGI (20, ± 2.4). When the subject sample was segregated by age, the younger subjects (70-79 years old) averaged slightly higher on both the BBS (51, + 3.1) and DGI (21, ± 2.0) than the older subjects (> 80) BBS (48, ± 3.3) and DGI (19, ± 2.0) respectively.

Table 1.

Study demographics and subject characteristics.

Variable	N (%)	Mean (± STD)	p level
Gender			
Male	26(33)		
Female	52(67)		
Age			
Male		79.2 (± 5.6)	
Female		77.2 (± 4.5)	0.08
Activity level (UAB-LSAF)		24*	
DF AROM (Left/Right)			
Male		11/11	
Female		11/12	0.32
Strength (Lbs. force) (Left/Right)			
Dorsiflexion			
Male		23/23	
Female		19/18	
Plantarflexion			
Male		25/23	
Female		19/20	
Hip extension			
Male		29/31	
Female		24/25	

* A score of 24 on Level 4 of the UAB-LSAF is a subject that at minimum walks in his/her community 1-3 times/week without any assistance from another person or assistive device.

Table 2.

Functional Balance Measures by gender and age.

Variable	Mean score (± STD)
Berg Balance Scale (BBS)	50 (± 3.5)
Dynamic Gait Index (DGI)	20 (± 2.1)
Gender:	
BBS:	
Males	49 (± 3.7)
Females	51 (± 3.3)
DGI:	
Males	20 (± 2.3)
Females	20 (± 2.0)
Age:	
BBS:	
70-79 years old	51 (± 3.1)
> 80	48 (± 3.3)
DGI:	
70-79 years old	21 (± 2.0)
> 80	19 (± 2.0)

Three different spatiotemporal parameters were compared between level ground and incline walking for each subject. Mean values were collected and potential differences between both walking surfaces were analyzed using paired t-tests for each spatiotemporal parameter: cadence, step length, mean normalized velocity (MNV). Gait-stability ratio (GSR), a measure of dynamic stability during gait, was also compared between surfaces using a t-test.

Cadence

Hypothesis 1: Cadence in older adults will decrease during incline gait when compared to level walking.

The mean cadence found on inclines (M = 111.57, SD = 8.96, n = 78) was significantly less than level ground, $t(77)$ = 7.19, one-tailed, p < .01. The 95% confidence interval for cadence scores on inclines ranged from 3.96 to 6.98. These findings support hypothesis one.

Step length

H2: Step length in older adults will increase during incline gait when compared to level walking.

The mean step length measured on inclines (M = 63.10, SD = 8.80, n = 78) decreased significantly when compared to level ground step length, $t(77)$ = 2.40, one-tailed, p = .01. The 95% confidence interval for cadence scores on inclines ranged from 0.20 to 2.12. Results on step length did not support hypothesis two.

Mean Normalized Velocity

H3: Gait velocity (MNV) in older adults will decrease during incline gait as compared to level walking.

Mean normalized velocity on inclines (M = 1.41, SD = 0.23, n = 78) was significantly less than level ground velocity, $t(77)$ = 6.44, one-tailed, p < .01. The 95% confidence interval for cadence scores on inclines ranged from 0.06 to 0.11. This finding supported hypothesis three.

4.5 Gait-Stability Ratio

H4: Gait-stability ratio (GSR) will increase during incline gait as compared to level ground walking.

Mean GSR values on inclines (M = 1.62, SD = 0.26, n = 78) were significantly greater than level ground GSR values, $t(77)$ = -2.73, one-tailed, p <

.01. The 95% confidence interval for cadence scores on inclines ranged from -0.07 to -0.01. This comparison assesses whether subjects increased the amount of steps per meter suggesting decreased balance. This finding supported hypothesis four.

Functional balance measures- Berg Balance Scale

H5: Older adults with lower BBS scores ($\leq$ 45) will demonstrate increased GSR while walking on inclines compared to level ground.

A repeated measures analysis of variance was performed to compare the difference between GSR on level and incline surfaces within subjects. In cases such as this when there are only two levels of an independent variable sphericity is automatically assumed, therefore, a Mauchly's test is not applicable (Field, 2009). The overall test for differences in means in the repeated-measures ANOVA was not significant (F1, 12) = 1.78, p = 0.20. This finding did not support hypothesis five.

Functional balance measures-Dynamic Gait Index

H6: Older adults with lower DGI scores ($\leq$ 19) will demonstrate increased GSR while walking on inclines compared to level ground.

A repeated measures analysis of variance was performed to compare the difference between GSR on level and incline surfaces within subjects. In cases such as this when there are only two levels of an independent variable sphericity is automatically assumed, therefore, a Mauchly's test is not applicable (Field, 2009). The overall test for differences in means in the repeated-measures

ANOVA was significant (F1, 12) = 4.12, p = 0.05. This finding supported

hypothesis six.

CHAPTER V

Discussion

It is well documented that healthy, young adults (ages 25-34) walk on inclines differently than level ground in order to progress the body's center of mass forward and upward against increased gravitational demands (Tokuhiro, Nagashima & Takechi, 1985; Leroux, Fung & Barbeau, 2002). Kinematic studies have supported this notion demonstrating increased hip extensor and plantarflexion power requirements in young healthy adults (Lay, Hass, Nichols & Gregor, 2007). When ambulating on inclined surfaces increased dorsiflexion has been observed in the swing phase of the gait cycle in order to safely clear the toes over a gradually increasing slope angle (Lange et al., 1996). Spatiotemporal aspects including cadence, step and stride length and velocity also change significantly on inclined surfaces (Sun, Walters, Svensson & Lloyd, 1996; McIntosh, Beatty, Dwan & Vickers, 2006). Despite clear evidence that walking patterns change in young adults none of these incline studies have analyzed the potential effects of inclines on gait adaptations in older adults. Therefore, efforts were made in this study to expand on the results of earlier research efforts specifically to healthy older adults who are living longer and maintaining active lifestyles into their eighth and ninth decades (Peel et al., 2005).

Several hypotheses were posed in order to assess the effect of an incline specifically on spatiotemporal aspects of older adult walking patterns. The first

hypothesis stated that cadence will decrease on inclines relative to level ground walking and this notion was supported by the results of this study. This decrease in cadence is similar to the adaptations seen in younger gait profiles on inclines (Kawamura, Tokuhiro & Takechi, 1991; McIntosh, Beatty, Dwan & Vickers, 2006).

The most likely explanation for the decrease in cadence was the increase in the temporal component. Specifically, the duration of each cycle increases during incline walking due to the increase in vertical displacement during each stride (Leroux, Fung & Barbeau, 2002; McIntosh, Beatty, Dwan & Vickers, 2006). Therefore, as the slope increases, the time necessary for each step increases thereby decreasing the number of steps taken in a fixed period of time. However, in this study, there was less effect on cadence due to the fact that step length (the spatial component), which typically inversely related to cadence, decreased as the older adults walked on inclines.

In this adult population step length did not increase on an incline walking surface. The most likely explanation is that this adaptation increases postural stability which is a primary concern of older adults. Results of this study lend additional support to the idea that when balance or safety is a concern, older adults alter their walking pattern to prioritize stability (Hausdorff et al., 1997; Owings & Grabiner, 2004). By increasing stability during walking older adults compensate for reductions in balance control (Cromwell & Newton, 2004). Although older adults tend to prioritize balance on level ground, this phenomenon is magnified on inclines (Rogers, Cromwell & Grady, 2008).

Previous research analyzing the step lengths of younger adults (age range 20-30 years old) on inclines have arrived at varying conclusions. Kawamura, Tokuhiro and Takechi (1991) found that there was an incremental increase in step length as slope angles increased plateauing at 10 degrees and then decreasing slightly at 12 degrees. Similarly, in a healthy younger population that does not perceive the incline as a hazard, stride length, which is comprised of two subsequent step lengths, increased with a gradually increasing slope angle leveling off at 10 degrees (McIntosh, Beatty, Dwan & Vickers, 2006). In younger populations it appears the goal while ambulating on an incline is to maintain a steady velocity while progressing uphill which is most efficiently accomplished by lengthening step and stride length.

These results differed, however, from those of a large outdoor observational study in which step length decreased in all age subgroups as the angle of the slope increased from level to 9 degrees (Sun, Walters, Svensson & Lloyd, 1996). Researchers concluded that the decrease in step length might have occurred due to the perception of a fall risk as condensation from weather conditions developed on the incline. Therefore, to offset the risk of slipping young and old pedestrians took smaller steps.

The third hypothesis proposed that mean normalized velocity (MNV) would decrease on inclines and was supported by the results of this study. Relative to level ground walking, velocity or MNV (normalized for leg length) decreased as older adults walked on inclines. Previous research findings have shown that walking speed generally decreases with age on level ground by

approximately 0.1 -0.7% per year after the age of 70 (Woo et al., 1995). The reasons for the decrease in speed associated with aging have been well documented in the literature. Slowing down as a gait adaptation may be associated with the general decrease in muscle strength due to a decrease in motor neurons, muscle fibers and aerobic capacity (Bendall, Bassey & Pearson, 1989; Trueblood & Rubenstein, 1991). Others have proposed that decreased velocity may also be due to various neurological systems becoming less efficient including declines in somatosensory feedback, vestibular and visual sensory systems (Peterka, Black & Schoenhoff, 1990; Stelmach & Worringham, 1985).

Previous findings from incline studies on young adults are not as robust. McIntosh, Beatty, Dwan and Vickers (2006) suggested that in healthy younger adults where there is no perception of a hazard (i.e. handrails and high friction floor) walking speed increased due primarily to longer stride lengths. However, at angles approaching 12 degrees, which is slightly more than in this study, velocity significantly decreased in younger subjects as they negotiated inclines (Kawamura, Tokuhiro and Takechi, 1991). Similarly, young and older subjects significantly decreased walking speed with increasing slope (Sun, Walters, Svensson & Walters, 1996).

In the present study (mean age = 78.2), decreases in both step length and cadence contributed to a decreased MNV during incline walking. Because of the vertical displacement of the body's center of mass and the increased work requirements older adults took longer to cover the same distance on inclines relative to level ground. While walking on level ground there is minimal active

joint motion required at the hip, knee and ankle required (Perry, 1992). However, while walking on inclines angular excursions of all three joints increase. It is likely that the increased physiological demand and the increased challenge to balance contribute to an increased duration of each gait cycle while walking on inclines (Leroux, Fung & Barbeau, 2002; McIntosh, Beatty, Dwan & Vickers, 2006). The resultant spatiotemporal effect is an overall decrease in MNV as measured by the GAITRite.

Results from this study also support the fourth hypothesis which states that Gait Stability Ratio (GSR) increases while walking on inclines relative to level ground. GSR is a ratio of cadence and velocity measured in units of steps per meter. The GSR which was initially designed to provide an indication of the amount an adaptation an individual makes to increase gait stability (Cromwell & Newton, 2004). An increase in the amount of steps per given distance suggests a subject's attempt at increasing the amount of time in double support and are indicative of an adaptation to increase stability (Rogers, Cromwell & Grady, 2008). An increase in GSR can also be viewed as a decrease in step length, slower forward progression of the body's center of mass and subsequent increased percentage of double-limb support time of the gait cycle (Rogers, Cromwell & Grady, 2008).

It is important to clarify that this ratio is derived directly from parameters collected by the GAITRite, cadence (steps/second) and velocity (cm/second). GSR is also considered a more sensitive measure of dynamic balance than either cadence or velocity alone making it useful for gait and balance analysis

ideal for healthy older adults and potentially one that may also be used as a predictor of balance loss or falls (Rogers, Cromwell & Grady, 2008). As hypothesized, GSR significantly increased on inclines when compared to the same subjects' GSR on level ground. In this particular study, both cadence and velocity decreased but not at an equal ratio. Therefore, the most logical explanation for the increase in GSR on inclines observed was a result of a larger decrease in velocity relative to cadence.

In this study the GSRs of all of the subjects were analyzed first comparing the changes between level ground and incline walking. Further analysis was then performed with GSR looking at those subjects who did not score over the established cutoffs for two standardized functional balance measures, the BBS and DGI. By separating the total sample of this study into those subjects who were challenged more and who scored lower on these two tests more information about incline negotiation could be realized.

By definition all subjects were deemed independent community ambulators using the University of Alabama-Birmingham Life Space Assessment Form and meeting all inclusion criteria. Despite all 78 subjects qualifying for this study as active and independent 12 of the subjects scored below the 45/56 cutoff established as increased fall risk on the Berg Balance Scale (Berg et al, 1992).

The fifth hypothesis stated that those subjects who scored lower than 45 on the BBS would have an increased GSR on inclines relative to level ground was supported by the results of this study. Although the differences did not reach significance, higher GSRs were found on inclines. A small sample size for

this subgroup and the nature of the items on the test most likely contributed to the lack of significance.

Previous studies have shown a ceiling effect for healthy older adults citing that the items are not difficult enough to challenge this population (Cromwell & Newton, 2004). Many of the items on the test are static balance activities such as sitting and standing unsupported and standing with feet together, tandem stance and single leg stance which may challenge a person's balance but not in a dynamic manner as gait does. Other items on the BBS are relatively simple tasks for healthy subjects. These include picking up an object from the ground, reaching forward, looking over one's shoulder, and transferring from one chair to another. In samples of healthy populations scores on these 11 items have been found to inflate the final scores and may not provide an accurate view of dynamic balance (Muir, Berg, Chesworth, Klar & Speechley, 2010). However, Item 12, alternate foot tapping on a stool for a fixed time period, is the only item on the BBS that has been highly correlated to aspects of gait such as walking velocity and GSR. The high correlation was attributed to the repetitive alternate leg movement and weightshifting which are similar to the dynamic aspects of walking (Cromwell & Newton, 2004). Although a frequently used and validated predictor of falls in the elderly, the BBS and its limited dynamic components, was not an ideal option for this study based on the high level of function of the population being tested.

Similar to the analysis for the low scoring BBS subgroup, a separate analysis was performed for the low scoring DGI subgroup. 22 subjects scored $\leq$

19 out of 24 designating them at a higher risk for falls (Shumway-Cook, 1996). The comparison of this subgroup's GSR on inclines relative to level ground was analyzed and results supported the hypothesis that GSR increases on inclines. However, unlike the results of BBS, the difference in this subgroup's GSR on inclines reached a level of significance. This finding supports the notion that those that scored lower on the more dynamic balance test did, in fact, change to a more stable pattern with more step per meter while walking on inclines.

It is likely that the significance that was attained in the DGI lies in the nature of the items on the DGI and the test's design. All of the items on the DGI are dynamic gait activities which are more applicable to comparisons with GSR and walking on inclines. Items such as walking with head turns both vertically and horizontally, stepping over an obstacle, changing speeds and walking around aspects require subjects to split their attention to more than one motor task. Walking on inclines requires similar planning and accommodation as the increasing slope induces greater strength, balance and ROM demands. Since none of the subjects fell or lost their balance it was clear that they successfully and safely altered their patterns to accommodate for the ramp as required to complete the items on the DGI.

Several limitations of this study have been identified. The first was that although the sample was large and culturally diverse, it was one of convenience. While this sample is certainly representative of a larger population of healthy and independent older adults caution should be used when generalizing these results. Second, because of the nature of the items on the BBS and the ceiling

effect for higher functioning healthy adults the BBS proved to be less than ideal for measuring aspects of dynamic balance in this study. Also once the total study sample was separated into subgroups of low and high scores on the BBS and the DGI the small sample sizes decreased the power of the results and may have contributed to the lack of significance among the sample. In addition, although the use of the GSR is gaining momentum as a highly specific and practical measure of gait stability it has not been validated against any gold standard. Lastly, spatiotemporal data collection is limited to conclusions based on what aspects of gait changed and not necessarily how these changes occurred as a kinematic or kinetic study may have provided.

CHAPTER VI

Conclusion

Despite the acknowledged limitations this study provided important

information on healthy adults over the age of 70. To date there has been virtually

no data collected on this rapidly growing population with regard to walking and

balance on any surface other than level ground. This study begins to look at a

normal, healthy older population without pathologies as they walk over a

common community barrier in order to provide baseline data for comparative

analyses in future studies. By using a repeated measures design this study

minimized external variables and was effective at determining the true treatment

effect of an incline walking surface. Clear differences in spatiotemporal

parameters of gait were established between level and incline walking surfaces.

In addition, although a small sample of lower score DGI subgroup was analyzed

there was some evidence to suggest that this test may be more appropriate for

assessing balance in healthy, older adults than the BBS.

The primary goal of this study was to examine the effects of an incline

walking surface on spatiotemporal gait parameters of healthy older adults. The

results of this study demonstrate that, in fact, healthy older adults do change their

gait pattern as they walk up an incline surface which is similar in slope to those

encountered in the community. Since none of the subjects fell or lost their

balance as they negotiated the incline surface there was an obvious conscious

adjustment in motor planning. This finding is important because it suggests that healthy older adults behave similarly to younger adults in that regard.

However, despite older adults adapting their gait pattern to the incline as younger subjects had done in previous studies the method of adaptation was different. This study clearly showed that, unlike younger adults who took longer steps and often increased velocities on inclines, older adults took smaller and slower steps with an overall subsequent decrease in velocity. This study's findings of decreased cadence, step length and increased GSR suggest that older adults prioritize stability on inclines. As suggested in previous work, older adults attempt to limit the amount of time that they are in single support and therefore, increase their stability while walking up inclines (Cromwell & Newton, 2004; Rogers, Cromwell & Grady 2007).

A secondary purpose of this study was to understand how older adults at a higher fall risk, as identified by two valid clinical balance measures, are affected by an incline slope relative to level ground. More specifically, this study looked more closely at the highly specific measure of GSR among those subjects who had lower BBS (< 45) and DGI (≤ 19) scores to determine the effect of an incline walking surface. Although the sample sizes were small after separating these two subgroups from the total sample results showed that GSR increased for both the low score BBS subgroup and the low score DGI subgroup.

Because only 12 of the 78 subjects in the sample did not score at least a 45/56 on the BBS it is interesting and clinically important as clinicians attempt to objectify balance and function. Results of this study suggest that the BBS itself

may not be an ideal test to use when studying this population and should be used with some hesitation in healthy older adults. This study does, however, provide some evidence that the DGI can be used with healthy older populations when assessing balance and gait. Results from this study suggest that those subjects who scored lower took more steps per meter (higher GSR) on inclines. From a clinical view, clinicians must be aware that when training those that have gait or balance dysfunctions, progression towards more challenging and dynamic surfaces should be included particularly for healthy, active older adults.

Gait patterns observed in healthy older adults are similar to younger adults except for step length. In this study, relative to level ground walking older adult subjects took smaller, slower steps as they walked up the inclined walkway. This finding is significant since multiple studies have shown an increased variability and fall risk with decreased velocity (Hausdorff et al., 1997; Barak, Wagenaar & Holt, 2006; Kang & Dingwell, 2007). These results suggest that clinicians should focus on maintaining adequate but safe velocities while training their patients as opposed to slower, more calculated steps.

As with all research this study adds to already existing knowledge on gait and balance in older adults while providing a framework for future studies on a population that is rapidly expanding. Based on these initial findings, future studies can begin to address the effect of inclines in compromised populations such as adults at risk for falls. These higher risk populations may include self reported fallers or patients with Parkinson's Disease, vestibular or Multiple Sclerosis where symptoms and function vary over time often while maintaining

an active lifestyle. Another interesting extension of this study may be to compare adults with higher versus lower fall risk as determined from DGI or BBS or other standardized balance test. While studying inclines it is extremely important to understand the influence of decline surfaces on walking patterns and balance is important as well. Lastly, a closer analysis of the GSR as a clinical tool measuring dynamic aspects of gait is also warranted. Finally, any data collected on this age group that contributes to knowledge of important daily functions such as walking and balance has far reaching economic and social benefits by minimizing incidence of falls and fall-related sequellae.

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Appendix A

How are you walking?

Looking for adults 70 and over to participate in a study that will analyze walking patterns on incline ramps and level ground. During this study valuable information will be gained about strength, joint motion, balance and their contributions to walking. Ultimately, this study will provide information on possible determinants of instability while walking, particularly on inclines.

- Research will be conducted here at this facility.
- Participation in this study will only require approximately 45 minutes of your time.
- Individual appointment times will be made for each participant to minimize waiting.
- Participation in this study is completely voluntary.
- A coding system will be assigned to each participant to assure anonymity.

Eligibility requirements:
- You must be 70 years or older.
- You must be able to walk independently in the community.
- Free from ankle sprains in the last six months.
- Free from major traumas to ankle and foot that required medical attention.

FOR MORE INFORMATION ON PARTICIPATION IN THIS STUDY AND TO SCHEDULE A TIME FOR TESTING PLEASE CONTACT:

Richard Ferraro, Physical Therapist, MS
School of Health and Medical Sciences, Dept. of Graduate Programs in Health Sciences, Seton Hall University
732-995-2300 or email @ Richard.Ferraro@shu.edu

Appendix B

Name:	Date:

These questions refer to your activities just within the past month.

LIFE-SPACE LEVEL							INDEPENDENCE	SCORE
During the past four weeks, have you been to . . .		How often did you get there?					Did you use aids or equipment? Did you need help from another person?	Level X Frequency X Independence
Life-Space Level 1... Other rooms of your home besides the room where you sleep?	Yes 1	No 0	Less than 1 /week 1	1-3 times /week 2	4-6 times /week 3	Daily 4	1 = Personal assistance 1.5 = Equipment only 2 = No equipment or personal assistance	
Score	1 x		4		x	1.5 =		**6** Level 1 Score
Life-Space Level 2... An area outside your home such as your porch, deck or patio, hallway (of an apartment building) or garage, in your own yard or driveway?	Yes 2	No 0	Less than 1 /week 1	1-3 times /week 2	4-6 times /week 3	Daily 4	1 = Personal assistance 1.5 = Equipment only 2 = No equipment or personal assistance	
Score	2 x		4		x	1.5 =		**12** Level 2 Score
Life-Space Level 3... Places in your neighborhood, other than your own yard or apartment building?	Yes 3	No 0	Less than 1 /week 1	1-3 times /week 2	4-6 times /week 3	Daily 4	1 = Personal assistance 1.5 = Equipment only 2 = No equipment or personal assistance	
Score	3 x		2		x	1.5 =		**9** Level 3 Score
Life-Space Level 4... Places outside your neighborhood, but within your town?	Yes 4	No 0	Less than 1 /week 1	1-3 times /week 2	4-6 times /week 3	Daily 4	1 = Personal assistance 1.5 = Equipment only 2 = No equipment or personal assistance	
Score	4 x		2		x	1 =		**8** Level 4 Score
Life-Space Level 5... Places outside your town?	Yes 5	No 0	Less than 1 /week 1	1-3 times /week 2	4-6 times /week 3	Daily 4	1 = Personal assistance 1.5 = Equipment only 2 = No equipment or personal assistance	
Score	0 x				x	=		**0** Level 5 Score
						TOTAL SCORE (ADD)		**35** Sum of Levels

Figure 2.
Example of scoring of the Life-Space Assessment. The subject traveled to all levels (levels 1–4) except for out of town (level 5); traveled daily to levels 1 and 2, and traveled 1 to 3 times each week to levels 3 and 4; uses a cane at all times and requires assistance with driving.

Appendix C

Demographics Questionnaire

Name:
Age:
Code:
Date of participation:
Location:

Gender: Male_____   Female_____

1. Have you had any sprains, fractures or surgeries in your leg over the past six months?
Yes_____No_____

2. Do you suffer from any medical conditions that may affect your movement or balance?
Yes_____No_____

3. Have you fallen in the past year?   Yes_____No_____

4. Do you have a fear of falling while walking? Yes_____No_____

5. Are you currently taking any prescribed medications? Yes_____ No_____ If yes, what
are they?_____

6. Are you taking any over-the-counter medications? Yes_____ No_____ If yes, what are
they?_____

_____

7. Do you require assistance from another person to walk? Yes_____No_____

8. Do you ever use a cane, walker or crutch to walk in the community? Yes_____No_____

10. Are you currently experiencing pain in either leg? Yes_____No_____

11. Do you have any visual or perceptual problems? Yes_____No_____ If so, do you wear
corrective lenses? Yes_____No_____

12. Do you ever experience shortness of breath? Yes_____No_____

13. How often do you exercise? Every day_____ 3-4 times/week_____ never_____

14. Do you ever get dizzy when you walk? Yes_____ No_____

15. If you turn your head fast do you feel dizzy or lightheaded? Yes_____ No_____

16. When you bend down to pick something up do you feel dizzy or lightheaded? Yes_____
No_____

Appendix D

The Lafayette Manual Muscle Test System (MMT)
3700 Sagamore Parkway North . PO Box 5729 . Lafayette, IN 47903 USA . Ph: 765-423-1505

Features and Specifications:

System Features:
· Designed for high inter-instrument reproducibility
· Three, easy to change molded plastic stirrups with pads
· Force measurement in pounds or kilograms (user selectable)
· Measures peak force, time to reach peak force and total test time
· Data storage for up to 52 tests in on-board memory (peak force and time to reach peak force)
· Manual or automatic storage of data
· Dual measurement range: 0-300lbs. (136.1kg) or 0-50lbs. (22.6kg)
· Selectable test time from 1-10 seconds
· Tone to indicate end of preset test time
· Microprocessor controlled
· Easy to read graphical LCD display
· Manual ON/OFF switch
· Manual or automatic reset
· Built-in stored data browsing capability
· Low battery detection indicated by tone and icon
· Automatic battery saving sleep mode
· Interactive menus which allow user to select device options
· Battery powered: (1) lithium battery
· Minimal measurement drift

System Specifications:
· **Size:** 3" x 4" x 1.5" (7.6cm x 10.2cm x 3.8cm)
· **Weight:** 10.6oz (300g)
· **Range:** 0-300lbs. (136.1kg) / 0-50 pounds (22.6kg)
· **Accuracy:** ± 1% over full scale (both ranges)
· **Resolution:** 0.4lb (0.2kg) high range / 0.1lb (0.1kg) low range
· **Battery Life:** 80-85 hours, 10-12 hours after low battery condition
· **Timing Accuracy:** ± 0.03%
· **Data Storage Capacity:** 52 tests
· **Calibration Points:** 0, 25 and 50lbs. (0,11.3 and 22.6kg)
· **Preset Test Length:** 1-10 seconds; in 1 second increments.

Appendix E

Subject code:_____

Berg Balance Scale

The Berg Balance Scale (BBS) was developed to measure balance among older people with impairment in balance function by assessing the performance of functional tasks. It is a valid instrument used for evaluation of the effectiveness of interventions and for quantitative descriptions of function in clinical practice and research. The BBS has been evaluated in several reliability studies. A recent study of the BBS, which was completed in Finland, indicates that a change of eight (8) BBS points is required to reveal a genuine change in function between two assessments among older people who are dependent in ADL and living in residential care facilities.

Description:
14-item scale designed to measure balance of the older adult in a clinical setting.
Equipment needed: Ruler, two standard chairs (one with arm rests, one without), footstool or step, stopwatch or wristwatch, 15 ft walkway
Completion:
Time: 15-20 minutes
Scoring: A five-point scale, ranging from 0-4. "0" indicates the lowest level of function and "4" the highest level of function. Total Score = 56
Interpretation: 41-56 = low fall risk
21-40 = medium fall risk
0 –20 = high fall risk
A change of 8 points is required to reveal a genuine change in function between 2 assessments.

ITEM DESCRIPTION SCORE (0-4)
Sitting to standing _____
Standing unsupported _____
Sitting unsupported _____
Standing to sitting _____
Transfers _____
Standing with eyes closed _____
Standing with feet together _____
Reaching forward with outstretched arm _____
Retrieving object from floor _____
Turning to look behind _____
Turning 360 degrees _____
Placing alternate foot on stool _____
Standing with one foot in front _____
Standing on one foot _____
Total _____

GENERAL INSTRUCTIONS
Please document each task and/or give instructions as written. When scoring, please record the
lowest response category that applies for each item.
In most items, the subject is asked to maintain a given position for a specific time. Progressively
more points are deducted if:
☐the time or distance requirements are not met
☐the subject's performance warrants supervision
☐the subject touches an external support or receives assistance from the examiner
Subject should understand that they must maintain their balance while attempting the tasks. The
choices of which leg to stand on or how far to reach are left to the subject. Poor judgment will
adversely influence the performance and the scoring.
Equipment required for testing is a stopwatch or watch with a second hand, and a ruler or other
indicator of 2, 5, and 10 inches. Chairs used during testing should be a reasonable height. Either
a step or a stool of average step height may be used for item # 12.

SITTING TO STANDING
INSTRUCTIONS: Please stand up. Try not to use your hand for support.
( ) 4 able to stand without using hands and stabilize independently
( ) 3 able to stand independently using hands
( ) 2 able to stand using hands after several tries
( ) 1 needs minimal aid to stand or stabilize
( ) 0 needs moderate or maximal assist to stand
STANDING UNSUPPORTED
INSTRUCTIONS: Please stand for two minutes without holding on.
( ) 4 able to stand safely for 2 minutes
( ) 3 able to stand 2 minutes with supervision
( ) 2 able to stand 30 seconds unsupported
( ) 1 needs several tries to stand 30 seconds unsupported
( ) 0 unable to stand 30 seconds unsupported
If a subject is able to stand 2 minutes unsupported, score full points for sitting unsupported.
Proceed to item #4.
SITTING WITH BACK UNSUPPORTED BUT FEET SUPPORTED ON FLOOR OR ON A STOOL
INSTRUCTIONS: Please sit with arms folded for 2 minutes.
( ) 4 able to sit safely and securely for 2 minutes
( ) 3 able to sit 2 minutes under supervision
( ) 2 able to able to sit 30 seconds
( ) 1 able to sit 10 seconds
( ) 0 unable to sit without support 10 seconds
STANDING TO SITTING
INSTRUCTIONS: Please sit down.
( ) 4 sits safely with minimal use of hands
( ) 3 controls descent by using hands
( ) 2 uses back of legs against chair to control descent
( ) 1 sits independently but has uncontrolled descent
( ) 0 needs assist to sit

TRANSFERS
INSTRUCTIONS: Arrange chair(s) for pivot transfer. Ask subject to transfer one way toward a seat with armrests and one way toward a seat without armrests. You may use two chairs (one with and one without armrests) or a bed and a chair.
( ) 4 able to transfer safely with minor use of hands
( ) 3 able to transfer safely definite need of hands
( ) 2 able to transfer with verbal cuing and/or supervision
( ) 1 needs one person to assist
( ) 0 needs two people to assist or supervise to be safe
STANDING UNSUPPORTED WITH EYES CLOSED
INSTRUCTIONS: Please close your eyes and stand still for 10 seconds.
( ) 4 able to stand 10 seconds safely
( ) 3 able to stand 10 seconds with supervision
( ) 2 able to stand 3 seconds
( ) 1 unable to keep eyes closed 3 seconds but stays safely
( ) 0 needs help to keep from falling

STANDING UNSUPPORTED WITH FEET TOGETHER
INSTRUCTIONS: Place your feet together and stand without holding on.
( ) 4 able to place feet together independently and stand 1 minute safely
( ) 3 able to place feet together independently and stand 1 minute with supervision
( ) 2 able to place feet together independently but unable to hold for 30 seconds
( ) 1 needs help to attain position but able to stand 15 seconds feet together
( ) 0 needs help to attain position and unable to hold for 15 seconds
REACHING FORWARD WITH OUTSTRETCHED ARM WHILE STANDING
INSTRUCTIONS: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can. (Examiner places a ruler at
the end of fingertips when arm is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is
the distance forward that the fingers reach while the subject is in the most forward lean position. When possible, ask subject to use
both arms when reaching to avoid rotation of the trunk.)
( ) 4 can reach forward confidently 25 cm (10 inches)
( ) 3 can reach forward 12 cm (5 inches)
( ) 2 can reach forward 5 cm (2 inches)
( ) 1 reaches forward but needs supervision
( ) 0 loses balance while trying/requires external support
PICK UP OBJECT FROM THE FLOOR FROM A STANDING POSITION
INSTRUCTIONS: Pick up the shoe/slipper, which is in front of your feet.
( ) 4 able to pick up slipper safely and easily
( ) 3 able to pick up slipper but needs supervision
( ) 2 unable to pick up but reaches 2-5 cm(1-2 inches) from slipper and keeps balance independently
( ) 1 unable to pick up and needs supervision while trying
( ) 0 unable to try/needs assist to keep from losing balance or falling
TURNING TO LOOK BEHIND OVER LEFT AND RIGHT SHOULDERS WHILE STANDING
INSTRUCTIONS: Turn to look directly behind you over toward the left shoulder. Repeat to the right. (Examiner may pick an object
to look at directly behind the subject to encourage a better twist turn.)
( ) 4 looks behind from both sides and weight shifts well
( ) 3 looks behind one side only other side shows less weight shift
( ) 2 turns sideways only but maintains balance
( ) 1 needs supervision when turning
( ) 0 needs assist to keep from losing balance or falling

TURN 360 DEGREES
INSTRUCTIONS: Turn completely around in a full circle. Pause. Then turn a full circle in the other direction.
( ) 4 able to turn 360 degrees safely in 4 seconds or less
( ) 3 able to turn 360 degrees safely one side only 4 seconds or less
( ) 2 able to turn 360 degrees safely but slowly
( ) 1 needs close supervision or verbal cuing
( ) 0 needs assistance while turning

PLACE ALTERNATE FOOT ON STEP OR STOOL WHILE STANDING UNSUPPORTED
INSTRUCTIONS: Place each foot alternately on the step/stool. Continue until each foot has touched the step/stool four times.
( ) 4 able to stand independently and safely and complete 8 steps in 20 seconds
( ) 3 able to stand independently and complete 8 steps in > 20 seconds
( ) 2 able to complete 4 steps without aid with supervision
( ) 1 able to complete > 2 steps needs minimal assist
( ) 0 needs assistance to keep from falling/unable to try
STANDING UNSUPPORTED ONE FOOT IN FRONT
INSTRUCTIONS: (DEMONSTRATE TO SUBJECT) Place one foot directly in front of the other. If you feel that you cannot place
your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot. (To
score 3 points, the length of the step should exceed the length of the other foot and the width of the stance should approximate the
subject's normal stride width.)
( ) 4 able to place foot tandem independently and hold 30 seconds
( ) 3 able to place foot ahead independently and hold 30 seconds
( ) 2 able to take small step independently and hold 30 seconds
( ) 1 needs help to step but can hold 15 seconds
( ) 0 loses balance while stepping or standing
STANDING ON ONE LEG
INSTRUCTIONS: Stand on one leg as long as you can without holding on.
( ) 4 able to lift leg independently and hold > 10 seconds
( ) 3 able to lift leg independently and hold 5-10 seconds
( ) 2 able to lift leg independently and hold L 3 seconds
( ) 1 tries to lift leg unable to hold 3 seconds but remains standing independently.
( ) 0 unable to try of needs assist to prevent fall

( ) TOTAL SCORE (Maximum = 56)

Appendix F

Dynamic Gait Index

Grading: Mark the lowest category which applies. Total individual scores (24 possible). Scores of 19 or less have been related to increase incidence of falls in the elderly.

1. Gait Level Surface _____
Instructions: Walk at your normal speed from here to the next mark (20').
Grading: Mark the lowest category that applies.

(3) Normal: Walks 20', no assistive devices, good speed, no evidence for imbalance, normal gait pattern.
(2) Mild impairment: Walks 20', uses assistive devices, slower speed, mild gait deviations.
(1) Moderate impairment: Walks 20', slow speed, abnormal gait pattern, evidence for imbalance.
(0) Severe impairment: Cannot walk 20' without assistance, severe gait deviations, or imbalance.

2. Change in gait speed _____
Instructions: Begin walking at your normal pace (for 5'), when I tell you "go," walk as _fast_ as you can (for 5'). When I tell you "slow," walk as _slowly_ as you can (for 5').

(3) Normal: Able to smoothly change walking speed without loss of balance or gait deviation. Shows a significant difference in walking speeds between normal, fast, and slow speeds.
(2) Mild impairment: Able to change speed but demonstrates mild gait deviations, or no gait deviations but unable to achieve a significant change in velocity, or uses and assistive device.
(1) Moderate impairment: Makes only minor adjustments to walking speed, or accomplishes a change in speed with significant gait deviations, or changes speed but has significant gait deviations, or changes speed but loses balance but is able to recover and continue walking.
(0) Severe impairment: Cannot change speeds, or loses balance and has to reach for wall or be caught.

3. Gait with horizontal head turns _____
Instructions: Begin walking at your normal pace. When I tell you to "look right," keep walking straight, but turn your head to the right. Keep looking to the right until I tell you "look left," then keep walking straight and turn your head to the left. Keep your head to the left until I tell you, "look straight," then keep walking straight but return your head to the center.

(3) Normal: Performs head turns smoothly with no change in gait.
(2) Mild impairment: Performs head turns smoothly with slight change in gait velocity (i.e., minor disruption to smooth gait path or uses walking aid).
(1) Moderate impairment: Performs head turns with moderate change in gait velocity, slows down, staggers but recovers, can continue to walk.

(0) Severe impairment: Performs task with severe disruptions of gait (i.e., staggers outside 15° path, loses balance, stops, reaches for wall).

4. Gait with vertical head turns _____
Instructions: Begin walking at your normal pace. When I tell you to "look up," keep walking straight, but tip your head and look up. Keep looking up until I tell you "look down," then keep walking straight and turn your head down. Keep looking down until I tell you, "look straight," then keep walking straight but return your head to the center.

(3) Normal: Performs head turns with no change in gait.

(2) Mild impairment: Performs task with slight change in gait velocity (i.e., minor disruption to smooth gait path or uses walking aid).

(1) Moderate impairment: Performs tasks with moderate change in gait velocity, slows down, staggers but recovers, can continue to walk.

(0) Severe impairment: Performs task with severe disruption or gait (i.e., staggers outside 15° path, loses balance, stops reaches for wall).

5. Gait and pivot turn _____
Instructions: Begin walking at your normal pace. When I tell you to "stop and turn," turn as quickly as you can to face the opposite direction and stop.

(3) Normal: Pivot and turns safely within 3 seconds and stops quickly with no loss of balance.

(2) Mild impairment: Pivot turns safely in >3 seconds and stops with no loss of balance.

(1) Moderate impairment: Turns slowly, requires verbal cueing, requires several small steps to catch balance following turn and stop.

(0) Severe impairment: Cannot turn safely, requires assistance to turn and stop.

6. Step over obstacle _____
Instructions: Begin walking at your normal speed. When you come to the shoe box, step over it, not around it, and keep walking.

(3) Normal: Able to step over box without changing gait speed; no evidence for imbalance.

(2) Mild impairment: Able to step over box, but must slow down and adjust steps to clear box safely.

(1) Moderate impairment: Able to step over box but must stop, then step over. May require verbal cueing.

(0) Severe impairment: Cannot perform without assistance.

7. Step around obstacles _____
Instructions: Begin walking at your normal speed. When you come to the first cone (about 6' away), walk around the right side of it. When you come to the second cone (6' past first cone), walk around it to the left.

(3) Normal: Able to walk around cones safely without changing gait speed; no evidence of imbalance.

(2) Mild impairment: Able to step around both cones, but must slow down and adjust steps to clear cones.

(1) Moderate impairment: Able to clear cones but must significantly slow speed to accomplish task, or requires verbal cueing.

(0) Severe impairment: Unable to clear cones, walks into one or both cones, or requires physical assistance.
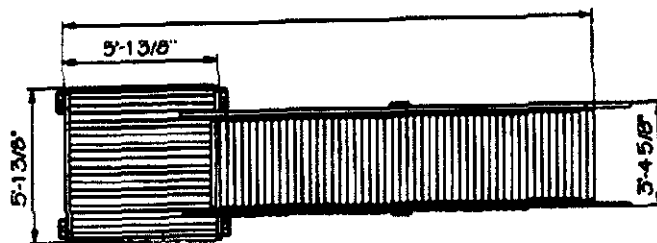
8. Stairs _____

Instructions: Walk up these stairs as you would at home (i.e., using the rail if necessary). At the top, turn around and walk down.
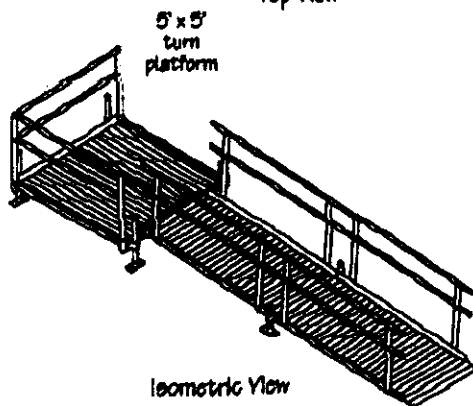
(3) Normal: Alternating feet, no rail.

(2) Mild impairment: Alternating feet, must use rail.

(1) Moderate impairment: Two feet to stair, must use rail.

(0) Severe impairment: Cannot perform safely.

Appendix G

Express Ramps Modular Ramp

Top View

Isometric View

- Meets or exceeds all ADA ramp specifications, section 4.8
- Easy installation. Install a 24' system, including handrails, in less than 30 minutes!
- Configure a ramp system for almost any site
- Ramps and platforms made of lightweight aluminum
- Ramps, handrails, and platforms are pre-assembled
- Our exclusive "patent pending" connecting system eliminates drilling and riveting handrails
- No anchoring into concrete footings or pads is required
- Architecturally designed for a clean, modern appearance
- 36" width
- 850 pound weight capacity

Appendix H

Telephone Screening Protocol

Hello Mr./Mrs._____:

My name is Rich Ferraro and I am the physical therapist that is conducting the research study on walking. I would like to take a few minutes of your time before you come to the research site to ask you a few questions to make sure that you are eligible for this study.

The first set of questions pertains to your level of interaction in the community and the level of assistance you require to walk in and around the community. If at any time you don't understand anything please stop me and ask that explain further.

At this time the primary investigator would ask questions as outlined on the following UAB-LSAF (see attached page 2) while a score is obtained.

If a minimum score of 16 is obtained at level four of the UAB-LSAF and there is no assistance required to ambulate the screening would continue.

The next set of questions pertains to some personal information related to medication use, any current diagnoses and exercise frequency. Do you have any questions thus far?

At this time the questions from the demographic questionnaire (see attached page 3) would be asked. If the primary investigator feels for any reason based on medication use or comorbidities that may affect the subject's ability to walk and move the subject would be told he/she does not qualify as follows:

Mrs. Jones based on your diagnoses of diabetes with peripheral neuropathy I cannot include you in this study for fear that the results would be effected adversely but I appreciate you taking time out to volunteer.

If at this time in the screening the requirements for the UAB-LSAF and the demographic questionnaire have been met the following statement would ensure:

Based on the questions you have answered Mrs. Jones you do indeed qualify for this study so I would like to set up an appointment that is convenient for you to come and participate in this study. How does September 30 at 11 AM at the Freehold Senior Center sound. If you have any other questions please feel free to contact me via email or phone with the number found on the flyer.

When a time has been confirmed the primary investigator would inform the subject to wear comfortable shoes without a heel that he/she is comfortable walking in.

Appendix I

Data Assessment Sheet

The scores obtained on the standardized Berg Balance Test _____ and the

Dynamic Gait Index _____ on (DATE)_____, suggest that

(NAME)_____ has met or exceeded the minimum balance

score requirements and is at lower risk for falls.

Richard Ferraro, PT MS

Appendix J

Data Assessment Sheet

The scores obtained on the standardized Berg Balance Test _____ and the

Dynamic Gait Index _____ on (DATE)_____, suggest that

(NAME)_____ has not met the minimum balance score

requirements and is considered at a higher risk for falls. Based on these findings it is

recommended that (NAME)_____ follow up with his/her physician

and seeks further intervention to address these apparent balance deficits.

Richard Ferraro, PT MS