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One of the most common neurological issues in the elderly is a stroke event, affecting nearly 800,000 adults in the U.S. alone every year. Since falls occur at a rate of 73% per year with people who are more than six months past the stroke event compared to a 30% fall rate in aged-matched healthy elderly, the potential consequences for injury are devastating. Current literature does not completely address the specific deficits in gait and balance after a stroke. To resolve this problem, the purpose of this investigation was to compare gait mechanics to clinical tests that indicate fall risks in 20 healthy elderly adults (63.4±8.9 years) and 7 non-cerebellar/non-brain stem stroke survivors (57.6±7.7 years). The dependent variables for gait were step length, step width, step time, and stride time for both the affected and unaffected sides. The metrics of mean, standard deviation (SD), coefficient of variation (CoV), detrended fluctuation analysis alpha (DFA α) and sample entropy (SampEn) were calculated for each dependent variable. Further, the Timed Up and Go (TUG), Berg balance assessment (Berg), Functional Gait Assessment (FGA), Activities-Specific Balance Confidence Scale (ABC), lower extremity strength, and lower extremity flexibility were taken as clinical assessments of fall risk. The data showed that most dependent variables for mean, SD, and CoV were different between groups, whereas DFA α and SampEn generally were not. The TUG, Berg, FGA, and ABC showed group differences. No differences in strength or flexibility were observed between the unaffected limbs of the stroke survivor group and matched limbs of the healthy elderly group. However, significant differences were observed in

strength and flexibility between the affected and matched limbs between groups. Sixty-four out of a possible 200 correlations between the gait and clinical metrics were found to be significant, indicating some relation between traditional laboratory tests and clinical assessments. These data suggest that summary metrics (mean, SD, and CoV) may be the strongest indicators of gait dysfunction after a stroke.

GAIT AND BALANCE CHARACTERISTICS AFTER A NON-CEREBELLAR STROKE

by

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APPROVAL PAGE

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CHAPTER I

INTRODUCTION

Gait is a fundamental movement skill that will be acquired early in life for most humans. From youth to older ages, gait evolves over time as a part of natural development and aging (Kesler et al, 2005; Hausdorff, Zemany, Peng & Goldberger, 1999). Changes to gait can also occur with injuries, from causes such as orthopedic or neurological changes, which typically are corrected through traditional physical therapy or other rehabilitation. Of the 45.5 million adults enrolled in Medicare part B services in 2006, 8.5% received outpatient physical therapy at over $3 billion in cost, with cost related specifically to gait rehabilitation reported to be $223 million (Clolek & Hwang, 2008; Fritz, Tracy & Brennan, 2011). In their longitudinal study, Fritz et al (2011) explored the usage of part B Medicare (outpatient physical therapy) for mobility issues. The authors discovered in post-treatment questionnaires that only 63.9% of adults over the age of 65 receiving therapy for musculoskeletal pain in eight Utah outpatient therapy clinics reported any benefit in mobility from their treatment. Similarly, in a study by Harada, Chiu, Fowler, Lee & Reuben (1995), no increases in gait speed were achieved for 27 patients, in spite of a month of "individualized" physical therapy training programs created to increase balance and mobility. Clearly more improvement in traditional

therapy practice s required, but the challenge to traditional rehabilitation is to determine the reasons that effective outcomes are not being consistently achieved. Certainly one such limitation is the lack of quantifiable metrics to objectively evaluate a patient's progress with gait in physical therapy. Despite the decades of study, the details of what constitutes normal, "healthy" gait and how it is controlled is not well understood. Without defining functional and dysfunctional gait for various adult populations, the faulty portions of gait may not be effectively corrected and rehabilitation programs might be less successful as prescribed. Therapeutic intervention is less than optimal without understanding exactly what parts of gait are functional and what parts are pathological. To accomplish this end, gait mechanics for normal, uninjured adults have been studied to discover what constitutes healthy gait patterns (Hausdorff et al, 2001; Wrisley, 2004). The gait patterns for frail older adults (Steffen, 2002; Newell, VanSwearingen, Hile & Brach, 2012; Halliday, Winter, Frank, Patla & Prince, 1998) and clinical populations of older adults (Krasovsky & Levin, 2010; Kempen et al, 2011; Rhea, Wutzke & Lewek, 2012; Halliday et al, 1998) have also been analyzed to identify how gait unfolds with normal life development, after injury, or after illness. In all, the majority of literature in this field has focused on the differences in gait speed, timing and coordination between healthy and clinical populations.

Healthy gait has specific characteristics of sequence, and a gait stride is usually defined by the distance from contact on one foot through to the contact of the same foot (Winter, 1991). The sequence of a gait stride is heel contact, then mid-stance, then toe-off, and then swing-through while the opposite leg conducts the same process in anti-

phase coordination. The gait cycle also has the elements of double leg stance (with both legs on the ground simultaneously) and single stance (with one foot in contact with the ground at a time) (Vaughan, Davis & O'Connor, 1999). The percentage of time on each limb is nearly equal in healthy gait during both double and single stance phases (Winter, 1991). The coordinated sequence of gait for healthy adults defines what traditionally is "normal" (Winter, 1991; Hausdorff, Peng, Laden, Wei & Goldberger, 1995).

When medical pathology occurs, the gait cycle can become asymmetrical (Vaughan et al, 1999) but also can be altered in other parameters of measure, such as joint range of motion, length of a stride, step width, gait speed, trunk movement and the excursion of the center of mass in three planes (Vaughan et al, 1999). The specific muscle groups assisting control for each movement are also important, as muscular control provides the force for acceleration and deceleration depending on the phase of the gait cycle (Winter, 1992). Accurate measures are possible for all these variables, which would likely allow specific diagnosis and quantification of asymmetry, refined treatment focus, and assessment of treatment success.

Of all pathology that can lead to asymmetry in gait, one of the most potentially devastating is a stroke. With rates approaching 800,000 incidents a year in the United States as of 2008, expectations of stroke incidence climbing to 21.9% by 2030 as relative to 2013, and the impact of higher rates among minorities and the poorly educated, this condition is poised to be a major concern for the medical and rehabilitation communities (Go et al, 2013). The survival rates of stroke are improving according to these authors, especially among men. This potentially could increase the need for rehabilitation

services in our aging population, highlighting the need for better quantitative measures of gait. For physical therapy practice to keep pace with the needs of greater numbers of stroke patients, evidence-based practice must be the standard.

In keeping with this standard, an examination of the gait variables from literature should be considered. The portions of gait impacted by stroke include reduced/asymmetrical step length and peak knee flexion (Lewek, Feasel, Wentz, Brooks & Whitton, 2012; Hwang et al, 2010; Jonsdottir et al, 2010). Loss of gait speed after a stroke is also noted in multiple studies (Bowden, Balasubrumanian, Behrman & Kautz, 2008; Dickstein, 2008) likely stemming from muscular changes from tone or strength losses (Salzman, 2010). One limitation in previous research is the use of relatively short duration trials to examine gait in stroke survivors. Short duration trials make the use of metrics examining underlying gait patterns less accurate (Damouras, Chang, Sejdic & Chau, 2009). There is a paucity of literature comparing quantitative measures of gait analysis between stroke survivors and healthy age-matched adults (Cruz, Lewek & Dhaher, 2009). Without this analysis of gait mechanics after a stroke, which is virtually impossible in clinical practice, best practices for prescribing treatment in physical therapy may not be adequately informed. The use of motion capture analysis on a treadmill could provide the evidence to bridge quantitative gait variability analyses (now commonly used in motor behavior research) with clinical practice.

The study of variability in gait (or gait dynamics) has been conducted for nearly two decades to explore the presence and meaning of gait variability between strides (Hausdorff et al, 1995). Hausdorff concluded that long-range gait patterns recur (persist)

in young healthy adults. These patterns have been postulated to reflect the presence of adaptive or maladaptive gait (West, 2007). This postulate is supported by findings of deteriorated gait variability patterns in frail or pathological elderly, such as those with Parkinson's and Huntington's disease (Hausdorff et al, 1997; Hove et al, 2012; Lamoth et al, 2011). In most cases, the study results show a loss of healthy variations of gait, even when the mechanism for the loss isn't clear.

Research examining gait dynamics in clinical populations of stroke survivors is limited (Rhea, Wutzke & Lewek, 2012; Roerdink & Beek, 2011; Roerdink et al, 2009). There also is currently no established connection among the clinical measures of motor function (Berg balance test, Timed Up and Go, Functional Gait Assessment, lower limb strength, lower limb flexibility), an assessment of fear of falling (Activities-Specific Balance Confidence Scale) and gait variability. A comprehensive comparison among these measurements would help researchers and clinicians understand how gait control is altered after a stroke. This could also drive physical therapy evidence-based practice and assist with pinpointing sources of falls post-stroke due to balance and gait dysfunction (Lord, Sherrington & Menz, 2001).

The issue for differentiating what is healthy normal gait versus pathological gait lies in knowing what is normal for the elderly population. A means of describing the details of healthy gait needs to be conducted with same-age comparison stroke populations to provide accurate comparison. Standardized balance and functional testing must also be done, since these tests are the field-based system in place for physical therapy practice. The details of step length, step width, step time, and stride time

compared with standardized clinical tests should overlap in a meaningful way to differentiate pathological gait and balance findings from those of healthy older people. A process of comparing the standard balance tests with motion capture camera data for gait would be helpful to see if the standard of practice employed by physical therapists is really a best practice.

The purpose of this study is two-fold: (1) to conduct a detailed analysis on gait characteristics collected with motion capture cameras to compare stroke survivors to healthy elderly when walking for a long duration (10 minutes) in order to accurately measure gait variability and (2) to explore the potential relationship between motion capture gait metrics and standardized clinical testing metrics. Based on the previous literature, the following five hypotheses were made:

Hypothesis 1: Stroke survivors would exhibit different mean values in the gait variables of interest (greater step width, and otherwise shorter step time, step length and stride time of affected versus unaffected limbs) relative to healthy elderly adults.

Hypothesis 2: The magnitude of variability of the gait variables (assessed via the SD and CoV) would be greater for the stroke survivors relative to healthy elderly adults.

Hypothesis 3: The structure of variability of the gait variables (assessed with DFA α and SampEn) would be different between the groups, with the expectation stroke survivors would have lower SampEn and DFA α values when walking at comfortable self-selected speeds relative to healthy elderly adults.

Hypothesis 4: Differences would be observed between groups in the clinical metrics (assessed via the TUG, Berg balance, FGA, ABC, strength and flexibility of

affected and unaffected limbs). Specifically, the stroke survivors will have higher TUG scores, lower FGA and Berg balance scores, lower ABC scores and lower measures of strength and flexibility on affected and unaffected limbs relative to healthy elderly adults.

Hypothesis 5: An exploratory hypothesis was made to examine the relationship between the gait and clinical variables. It was suggested lower values of the structure of variability (lower DFA α and SampEn) of the gait variables would be negatively correlated with the TUG and positively correlated with the Berg, FGA, and ABC.

CHAPTER II

REVIEW OF THE LITERATURE

Overview

As our population ages, the onset and advancement of physical and cognitive changes is inevitable. The incidence of pathological gait changes for the oldest segment of the population, from age 87 to 97, is about 80% (Bloem, Gussekloo, Lagaay, Remarque, Haan & Westendorp, 2000; Kesler et al, 2005). What constitutes normal aging gait is less defined in literature than the qualities that describe abnormal gait. The changes observed for the elderly include increased stance time on both legs, increased stance width, decreased gait speed, weaker toe-off strength and less definition to the heel to toe sequence, all of which has been termed "senile gait disorder" (Salzman, 2010). Many experts discuss the changes of gait and balance related to fall risk, with findings at age 65 defining the beginning of greater risk of a fall (Salzman, 2010; Westlake, 2007; Dhital, Pey & Stanford, 2010).

Many possible sources that contribute to changes in gait mechanics and control of balance are examined in literature. For example, a decrease in vestibular acuity resulted in greater postural shifts for 65 to 75 year old adults compared to those under age 40, and changes in visual acuity from age-related sources such as cataracts and glaucoma

are related to increased fall risk (Dhital et al, 2010). In the Dhital et al study, the risk of falls from vision changes due to glaucoma was related to increased postural sway on all surfaces, whether on firmer or softer surfaces. In a study of peripheral neuropathy affecting the feet and ankles with its' effect on control of postural stability, both muscle spindle proprioceptive information and plantar surface sensory input were compared for order of importance in controlling balance (van Deursen & Simoneau, (1999). It was concluded that the joint and muscle information of dorsiflexors and plantarflexors was reduced for subjects with diabetic peripheral neuropathy, even when controlling the sensory information received from the plantar surface of the foot. This demonstrates that proprioceptive information from the joints can impact balance aside from the peripheral sensory portion of the loss; the authors also note the nervous system's use of proprioceptive joint information more than the cutaneous portion of information during active movement. In a study examining gait dynamics by Gates and Dingwell (2007), a population with peripheral neuropathy and a control group were compared in walking trials lasting 10 minutes. The result of the Gates and Dingwell study concluded that while stride timing of the peripheral neuropathy group was more varied than the control group, the overall long-range structure of gait was not different between groups. These studies collectively show that alterations and use of sensory information can impact the control of gait.

Similarly, changes in strength with the normal aging process have been shown to alter balance and gait in a predictable manner in older adults. Strength losses related to aging resulted in 22% less isometric torque for hip flexion and 31% less for extension

among women ages 69 to 82 years (Dean, Kuo & Alexander, 2004) compared to 21 to 25 year old adults. Further, Dean and his co-authors noted a gait velocity loss related to hip movement with the older subjects. The authors note that the ability to recover from a trip event may be impacted by these changes, based on fall statistics for older adults.

These changes of gait may be better understood by creating a picture of normal healthy gait mechanics and documenting their relation to standard clinical metrics. By first describing the three-dimensional aspects of gait, we can better understand how the elements of this motor behavior coordinate in the gait cycle. The portions of gait that are altered by illness, injury, and aging are incompletely outlined in literature, making comparison among different conditions challenging.

One of the most commonly occurring neurological injuries is a non-cerebellar stroke, affecting up to 800,000 people in just the United States each year (Go et al, 2013). Further, up to 60% of stroke survivors experience gait changes that render them non-ambulatory or in need of assistance to walk initially (Lin, Hsu, Hsu, Wu & Hsieh, 2010). The specific changes in gait from a non-cerebellar stroke are partially illustrated in published literature, but altered gait mechanics resulting from stroke events need to be better defined to improve rehabilitation outcomes. With falls occurring at higher rates within the first year after a stroke, more refined gait and balance testing is crucial for determining patient safety. However, comparing clinical gait and balance assessments with a quantitative measurement of post-stroke gait mechanics has not been done. Additionally, the traditional clinical assessments may not identify gait and balance dysfunction characteristics that are better established using motion capture to get detailed

analysis. Validation of the balance testing with the "gold standard" of motion capture information may create more specific interventions clinically. The analysis of gait in this method allows examination of gait variability, which may help identify healthy, adaptive gait. This chapter will outline the literature in the following areas: (1) the gait cycle (2) how the gait cycle changes with aging and pathology (3) the role of gait variability (4) gait mechanics following a stroke and (5) standard clinical assessment tools for gait and balance after a stroke. The chapter summary details the gaps in literature relating to this thesis proposal and how project addressed the gaps.

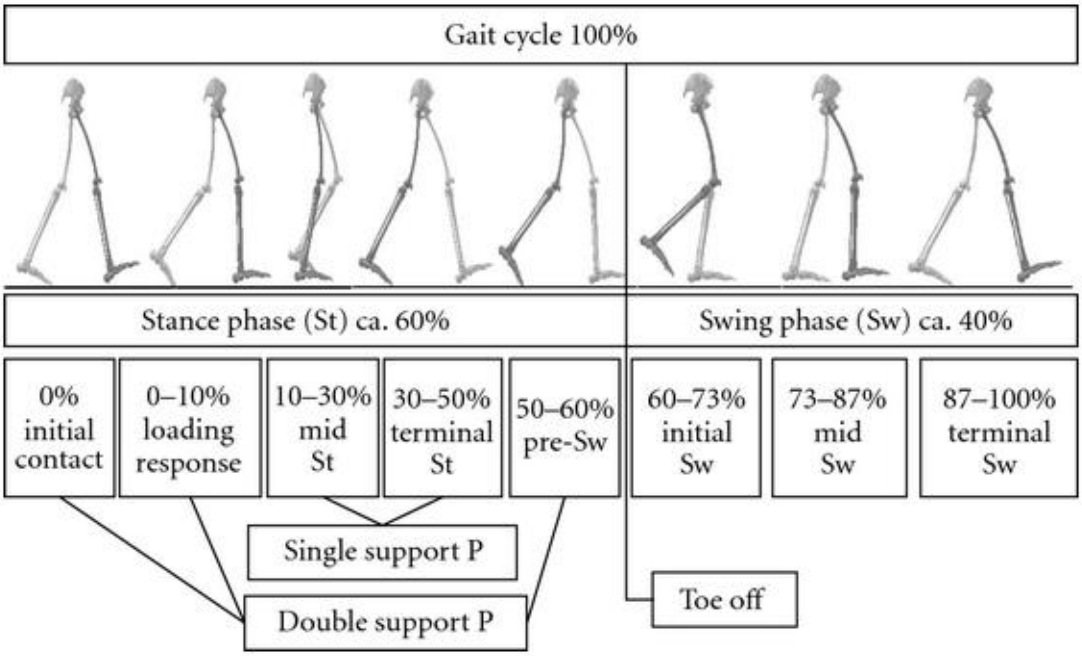
The Gait Cycle

Gait occurs in three planes of motion, with joint flexion/extension in the sagittal plane, rotation in the transverse plane and abduction/adduction in the frontal plane (Vaughan et al, 1999). As the sagittal plane is the focus for most two-dimension biomechanical measures (Winter, 1991; Robertson et al, 2004), the importance of the other two planes of normal gait merit further investigation. The characteristics of gait measured in the other planes include step width, displacement of the center of mass vertically and laterally, and symmetry of the limbs during gait phases (Rubino, 2002).

Gait is described as cyclical and periodic, alternating stance and swing elements of the lower extremities through a complete gait cycle (Vaughan et al, 1999). A cycle of gait conventionally is heel strike to heel strike on the same limb, with a repeating sequence with subsequent steps (Robertson et al, 2004). The stance portion of gait alternates from double to single leg stance, with single stance occurring in even time

frames for both legs with normal healthy gait (Vaughan et al, 1999). Double leg stance occurs for 20% of the gait cycle and 40% on each leg during single leg stance (Winter, 2009). The gait cycle is illustrated in the following figure (Hartmann, Kreuzpointner, Haefner, Michels, Schwirtz & Hass, 2010).

Figure 1. The Gait Cycle with Proportions for each Phase with Healthy Adult Gait



These proportions change with increasing speed, specifically with decreasing stance times and increasing swing times, reflecting the decreased time of feet on the ground (Winter, 1991). Winter reports an increase in stride length that accounts for these changes in the gait cycle relevant to speed increases. Speed also decreases step width with increased stride length, increased cadence, or steps per minute (Winter, 1991; Robertson, 2004). Conversely, slower gait speed is characterized by broader step width,

shorter stride, decreased cadence and increased stance time (Salzman, 2010; Winter, 1991).

The muscular control of gait involves several key groups: plantarflexors for propulsion, quadriceps for stability in stance and deceleration, gluteals for propulsion and hamstrings for deceleration and swing control (Vaughan et al, 1999; Winter, 1991). The plantarflexors (he gastrocnemius/soleus group) controls the ankle for the initiation of heel contact with peak force for the gastrocnemius at mid-stance, and soleus peaking forces at toe-off with an aggressive effort (Winter, 1991). The quadriceps muscle, comprised of the rectus femoris, vastus lateralis, vastus medialis and vastus intermedius, are primarily controllers for extending the knee just before heel strike. The quadriceps also control knee flexion at heel strike, peak control at mid-stance, and limits flexion of the knee to decelerate the swing phase as the leg moves backward (Winter, 1991; Hausdorff & Alexander, 2005). The gluteal muscles assist gait with late swing control of the hip and during heel strike to control hip flexion and forward rotation of the thigh (Vaughan et al, 1999; Winter, 1991). The hamstrings are comprised of a lateral head, biceps femoris with long and short segments, and medially, semimembranosus and semitendinosus. The function of hamstrings is to assist swing-through with maximal effort at the end and into deceleration at heel strike (Winter, 1991; Vaughan et al, 1999). While a number of trunk and lower extremity muscles contribute to the controlled adaptability of healthy gait, these key muscle groups are very important to synchronizing and coordinating balance as a contribution to gait (Trueblood, 2011; McGibbon, Krebs & Scarborough, 2003). Hip flexion is often tested for its contribution to gait (Bohannon, 1997; Wang, Olson &

Protas, 2002), but it is controlled by a weak muscle group relative to the others, as are the muscles that control dorsiflexion and eversion.

Gait Changes Due to Aging and Pathology

While the mechanics of healthy adult gait are fairly uniform in non-clinical populations, the presentation of gait with pathology or normal aging can be complicated, leading to challenges in defining and prescribing treatment to correct gait mechanics. Medical professionals have difficulty determining the causes of gait changes, partly because many factors can be at play (Salzman, 2010). Gait patterns are reflective of the underlying pathologies, with changes of speed being quite often noted in neurological populations (Rubino, 2002; Kluding & Gajewski, 2009) as well as fragile elderly populations (Wrisley & Kumar, 2010). In fact, several studies note the similarities of fragile elderly and neurological groups: the shuffling gait of Parkinson's disease is similar to fearful walking (Rubino, 2002) and dementia decreases overall activity and gait stability similarly as well (Rubino, 2002; Salzman, 2010). These authors suggest that gait changes in the elderly are subclinical symptoms of potential impending medical changes related to cardiovascular health and central nervous system alterations. The gait changes noted in the fragile elderly specifically are wider step width, shorter step length, increased step times, and greater stance times of both double and single leg stance postures (Pavol, Owings, Foley & Grabiner, 1999).

Strength changes are noted in both neurological and elderly populations, specifically changes in plantarflexors, hip extensors/abductors, quadriceps and

hamstrings, as the lower limb is weak along with hip extension and abduction (Horlings, van Engelen, Allum & Bloem, 2008; Bird et al, 2012). The Bird et al (2012) study specifically discusses weakness as a function of activity that fluctuates in the elderly populations as a factor of the time of the year. In the Dean et al (2004) study, the observation of older subjects having both strength and velocity losses in hip flexors and extensors was noted regarding that group's incidence of falls. The reduced age-related ability to both generate muscle force and produce quick movement reliably was deemed by these authors to present a greater fall risk for the older subjects.

The changes in gait that occur in unhealthy or elderly populations do not just create differences of measured step length or timing of individual steps. The potentially greater change would be the underlying patterns of the time series. These patterns represent a tendency toward either very self-similar or dissimilar steps. Healthy gait can be said to be self-similar in these metrics of variability, but not too extreme (Herman et al, 2005). In the Herman et al (2005) study, the subjects with high level gait disorder were noted to have very dissimilar patterns along with higher fall risk. Interestingly, the dissimilarity in gait patterns coincided with none of the balance tests, but related to a higher fear of falling. This suggests that the sensitivity of balance testing doesn't match up with functional changes in mobility, and that perception of loss of abilities may be an important screen for fall risk. This finding merits further investigation to see what other predictive factors for fall risk potentially could be identified.

Role of Gait Variability

Over the last twenty years, the number of scientific publications focusing on the inherent variability in the biological systems of healthy adult populations has increased (West, 2007). This variability was first observed as a normal fluctuation in systems such as heart rates (Peng, 1993). The theory behind these fluctuations is that the system must be in a state of preparedness for a sudden need to respond to a perturbation. These fluctuations have been recorded in many systems, including postural sway data (Petit, 2012), body temperatures (West, 2007), respiration rates and volumes (Sammer, 2010), and gait dynamics (Hausdorff et al, 2001). The fractal characteristics, meaning patterns that exhibit self-similar traits at many time scales, have been identified with a variety of metrics to derive meaning from these complex phenomena (Lamoth, 2011). The traditional metrics for measuring variation of human motor behavior have been statistical: mean, standard deviation, and median values. While these measures can be extremely useful, the calculations can be limiting to explain the complexities of abnormal and normal gait. For example, if ten measures of stride time are taken and all are 1.2 seconds in length, the mean is 1.2 seconds. In another example, if half the measures are one second and half are 1.4 seconds, the mean is still 1.2 seconds. The difference in the two calculations is that one is quite regular and the other has a large discrepancy between the first and second half of the behavior, yet the overall behavior of both groups appears the same with this metric. A more enlightening analysis would be a metric that examines the distribution of scores to see how similar or dissimilar the stride times actually are

throughout the behavior. More importantly, the measures of healthy biological systems fluctuate in ranges based on whether the system is diseased or healthy (Herman et al, 2005; Gates and Dingwell, 2006). For measures of gait, observations of stride length and stride time can be quantified and analyzed to see how the patterns correlate over a time series. These correlations are the nature of fractal gait, meaning the healthy patterns are unfolding over longer time periods (Hausdorff, 1995). As numerous metrics have been utilized to explain these phenomena in biological data, particularly in gait and balance (see Bravi et al, 2011 for a review), this literature review will cover two that are most frequently utilized for gait: detrended fluctuation analysis (DFA) and sample entropy (SampEn).

DFA

Physiological systems demonstrate fluctuations that occur in dynamic patterns of self-similarity, termed fractal patterns. These patterns have been shown to be structured in a predictable way, which can be described with detrended fluctuation analysis (DFA), a metric that compares patterns over many time scales to calculate the degree of self-similarity (Bravi et al, 2011; Herman, Giladi, Gurevich and Hausdorff, 2005; Hausdorff et al, 1995). Hausdorff et al (1995) demonstrated that the fluctuations in gait are not random, but reflect healthy control of a physiological system. The discovery of this underlying structure in young healthy adults (Hausdorff, 2007; Hausdorff, 1995) illuminates a new means of measuring and recording both decline and improvement in the control of gait for physical therapy and other branches of rehabilitation care. The

measurement of the patterns with DFA, along with other measures, may create a new standard for evaluating clinical populations for progress of recovery from injury and illness.

DFA quantifies self-similarity by quantifying the variability details of a physiological time-series. DFA was originally used to describe the structure of DNA, with the discovery that the exact sequence of a strand of DNA was deliberate and not random (Peng et al, 1994). The discovery that the ordering of thousands of DNA nucleotides was meaningful led to investigation of other types of physiological phenomena. In 1995, Hausdorff et al discovered that random shuffling of stride intervals created very different and unrelated patterns as compared to self-similar stride intervals: the original data display long range correlations (patterns over multiple time scales), which are present with young, healthy adults .

DFA is calculated by first demeaning the data, which is a subtraction of the average step measurement (S_{ave}) from each individual step measure (S_i).

Figure 2. Calculation of Demeaning of Data

$$y(k) = \sum_{i=1}^{k} \left[S(i) - S_{ave} \right]$$

The equation is the summation of each data point with subtraction of the mean values, leaving the remaining values of the points as the time series y(k). The y(k) time series is then portioned into boxes, starting with a few points (n=4), and then a trend line is created in each box. The trend values are subtracted from each data point, with

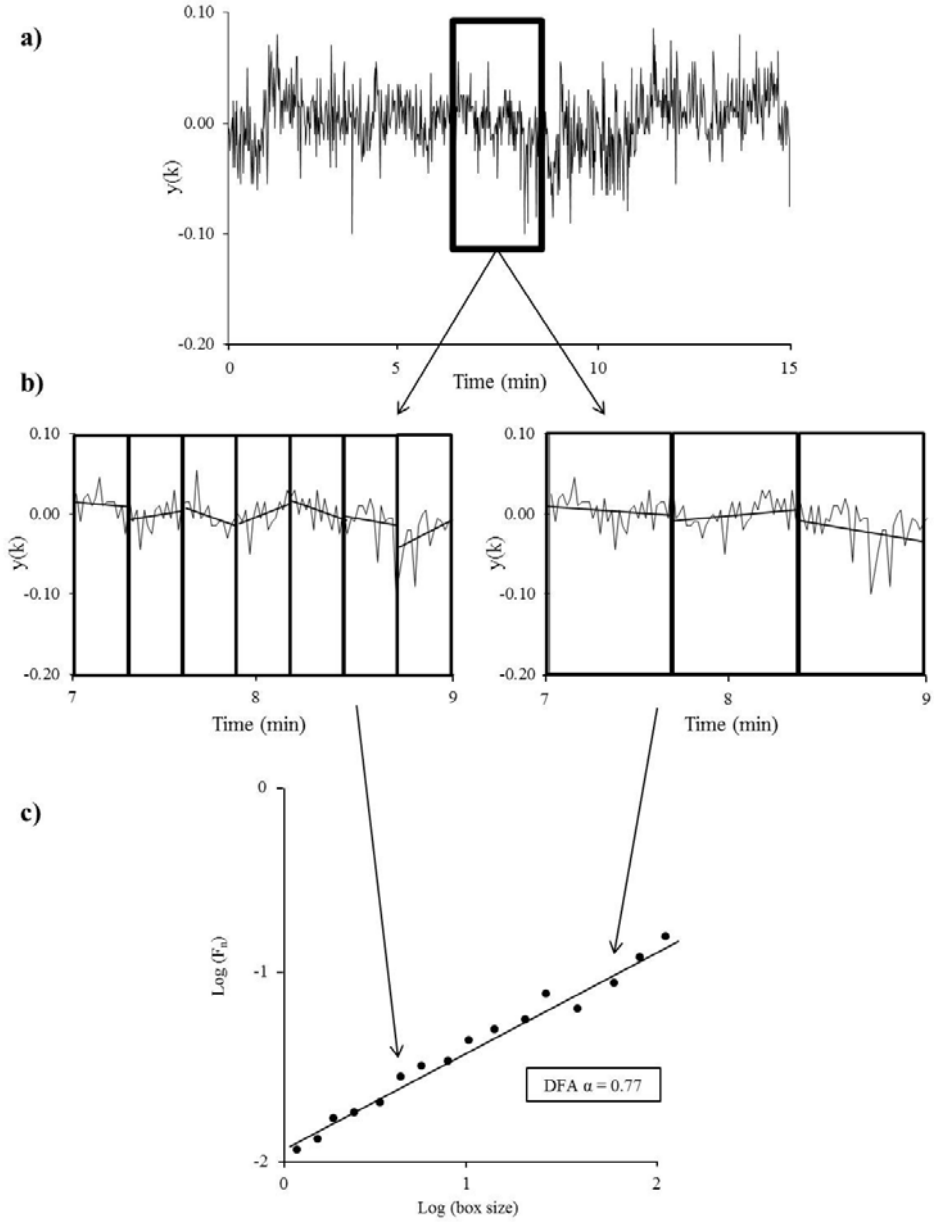
remaining values calculated as absolute numbers. The remaining detrended values are summarized in a Root Mean Square equation as follows.

Figure 3. Root Mean Square Calculation for DFA

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}[y(k) - y_n\,(k)]^2}$$

Root Mean Square is the amount of fluctuation in the integrated, detrended time series for that set of boxes. The log of the RMS values is plotted against the log of the box size to create a log-log plot. The process repeats by increasing box sizes by one more point (from n=4 to n=1/4 of time series length), and the process is reiterated. A line is then fit to the log-log plot and the slope of the line corresponds to the DFA alpha (α) metric. The DFA process is illustrated in Figure 4 taken from Rhea, Kiefer, and Warren (2014).

Figure 4. Log-Log Graph of Data Demonstrating Power Law Scaling

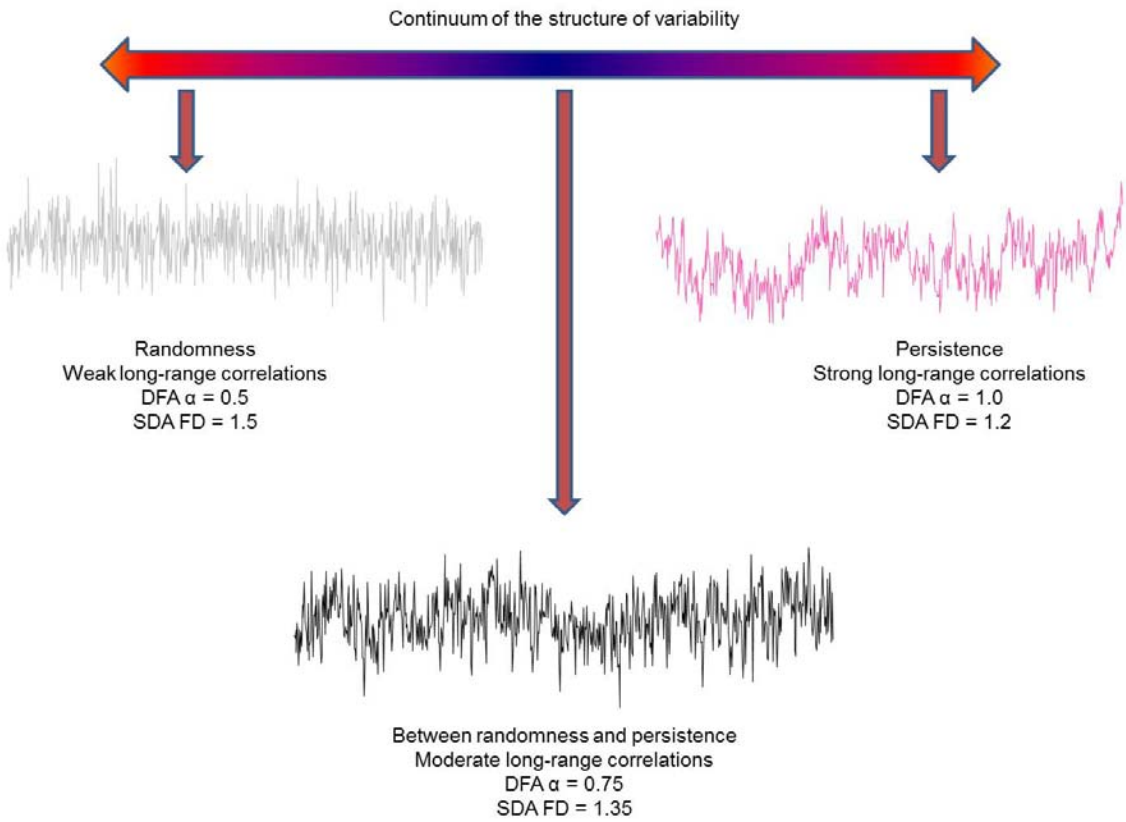


The data are said to be self-similar if the fluctuations scale as a power-law, meaning the integrated time series value increases with the increase in number of strides

or time scale. As Riley and Van Orden observe (2005), the measures of motor behavior in biological systems are not regular and stationary, and require special metrics to see the small details of patterns that may not repeat very often. The DFA α values for gait normally run around .75 (Hausdorff et al, 1995; Hausdorff, 2007). These numbers represent a balance between very random data with a DFA value of .50, and very regular data with a DFA slope of 1.0. When healthy adult stride values are shuffled, the DFA tends to run to .50 slope values. The same DFA values are seen in older adults with balance issues, and this loss of healthy adult variability is postulated to reflect lack of adaptability, possibly leading to falls (Hausdorff, 2007).

As complex systems operate, there are also interactions with other parts of the system, and all operate on varied time scales (West, 2007). West uses the analogy of a farm community delivering food to a city, with distribution and utilization occurring at a level that is not readily apparent and not centrally coordinated in an obvious way. In human systems, heart rates are usually 60 to 80 beats a minute, respiration rates are 15 to 20 breaths a minute and gait cycles take 1.0 to 1.5 seconds to complete. While the coordination of systems takes place in healthy individuals, even fluctuating over time based on age (Hausdorff, 2007), the source of the control remains somewhat a mystery. What is known is that when a change occurs in one portion of the system, such as an alteration of neurological information (e.g., Parkinson's disease), the end result is a change in variability of gait (Hausdorff, 2007). The following images are stride time variability as pink noise (DFA = .5) and white noise (DFA = 1.0) as depicted in a study by Rhea, Kiefer, D'Andrea, Warren, and Aaron (2014).

Figure 5. Examples of Time Series Depicting Random (DFA α = 0.50), Between Random and Persistent (DFA α = 0.75), and Persistent (DFA α = 1.0) Patterns.



SampEn

Another metric that has become more commonly used for evaluating complex time series data is Sample Entropy (SampEn). The primary usage for SampEn has been to evaluate cardiac data (Maestri et al, 2007) and EEG data (Song, Crowcroft and Zhang, 2012), but now it has been expanded to include Gait & Posture in a number of newer pieces of literature (Rhea, Wutzke and Lewek, 2012; Yentes, Hunt and Schmid, 2013; Rhea et al, 2011).

Sample entropy measures self-similarity by using a series of data points (*n* data points in length) and comparing this template to the successive strings of data in the time series. This process is repeated for time series of (*n*+1) without comparison to itself for each series. The string has an established tolerance of matches in the series to count, resulting in scoring of zero for insufficient matches to scoring one for matches that have met the minimum tolerance level (Bravi, Longtin, and Seely, 2011). The process continues with (*n*+2) and higher until all possible data strings have been compared to all templates.

Figure 6. Sample Entropy Calculation

$$\text{Samp En} = -\log \frac{A}{B}$$

The equation value A= number of template matches divided by B which is the number of attempted matches. SampEn values range from 0 (highly similar) to 2 (highly complex).

Gait Mechanics and Dynamics Post-Stroke

The changes of gait related specifically to stroke have been reported by many authors to describe numerous areas of deficit. Speed deficits are often reported and can be improved by compensatory strategies (Krasovsky and Levin, 2010). These authors make the point that compensation does not improve the functional reason behind the speed loss. Speed of gait is the utmost priority to those who are ambulating in the community, since

slow gait speed makes crossing the street more unsafe and difficult. Slow gait speed is identified as a common gait issue for post-stroke patients in a study conducted at Rancho Los Amigos Rehabilitation Hospital in California (Mulroy, Gronley, Weiss, Newsam and Perry, 2003). Individuals who were post-stroke at six months demonstrated varying gait patterns and were captured within five days of admission if they could walk with just moderate assistance and no orthosis, and within five days of being able to walk without an orthosis when possible. Gait kinematics were collected for 10 feet with motion capture cameras, along with EMG data of hamstrings, gluteus maximus, hamstrings, adductors, quadriceps and plantarflexors. The data were examined for gait velocity, step cadence and stride length. The results indicated four distinct groups: (1) a fast group with reduced knee extension mid stance, (2) a moderately fast group with greater mid stance knee flexion, (3) a slow velocity group with excessive knee flexion mid stance, and (4) a very slow velocity group with knee hyperextension mid stance and inadequate dorsiflexion. In addition, the strength of the hip extensors, knee extensors, and plantarflexors were reduced for all groups compared to the fastest group. The net result for the two slowest groups was that knee control was reduced either to buckling or hyperextension.

The loss of strength in plantarflexors has been linked to knee hyperextension, which is knee extension beyond neutral in a weight-bearing position, rather than being in slight flexion (Cooper et al, 2012). The strength changes, measured by a hand-held dynamometer, were also noted in quadriceps and hamstring muscles. The incidence of

the hyperextension is between 40- 60% of stroke patients, contributing to pain and laxity in the affected knee over time (Cooper et al, 2012).

The control of the knee has also been examined by comparing the progression of hip positions while controlling for head position (Lewek, Schmit, Hornby and Dhaher, 2006). This protocol was used to test quadriceps strength to see if the muscle was either inhibited or facilitated by hip position. The subjects were at least twelve months post-stroke, had hemiplegic symptoms from the stroke event, and were of ages 42 to70 years old. The results showed that hip proprioception and vestibular information of the head and trunk impact the force generation of the quadriceps at the knee.

In addition to strength changes, the deficiencies in muscle tone associated with a stroke event can affect gait dynamics. Muscle tone is involuntarily controlled by the central nervous system injury, which includes the brain and spinal cord, and is usually hypertonic from a hyperactive stretch reflex (Somerfeld, 2004). The resulting gait patterns can include circumduction of the leg on swing-through, scissoring of the legs or crossing in front of the other leg, dragging of the foot, and the tendency to hold the foot in an inverted or plantarflexed position (Alexander and Goldberg, 2005). The hypertonic state creates a resistance to movement, a state of velocity-dependent resistance (spasticity), or a spasmed resistance to movement (clonus).

Coordination of the involved and uninvolved sides of the body during postural control in standing and gait can be impaired. The clinical observations are reduced weight through the affected side, presented as postural shift to the uninvolved side, and therefore uneven and asymmetrical walking patterns are present (Cooper et al, 2012).

Cooper et al acknowledged that symmetry and coordination are not interchangeable terms: symmetry is one factor in coordination. In a an EMG training study with stroke survivors, it was shown that the training led to lower peak knee flexion and a decrease in stride length on the involved side (Jonsdottir et al, 2010). This asymmetrical movement is at least in part attributed to low plantarflexor power for push off.

Traditionally, changes are often examined by recording gait quality, but gait speed is often the functional indicator of success with therapy (Dickstein, 2008). Dickstein reported gait speed as low as .53 meters/second for stroke populations as compared with 1.34 meters/second for non-impaired control subjects of the same age. The ability to transition from being ambulatory in the home to the community, according to the author, rests on making the transition to the higher gait speed.

With so many variables in function and gait dynamics, some structured testing often is done in physical therapy to try to quantify the causes of these changes. The testing processes are an attempt to establish measures relating the changes to fall risk (Powell and Myers, 1995). Falls occur with at least 30% of adult females of the age of 65 or older (Lord, Sherrington and Menz, 2001). When these authors compared adults in post-stroke groups of the same ages, 73% were reporting falls within six months of discharge home from the hospital. Fall risk is substantial in the post-stroke populations, potentially leading to fractures and other debilitating injuries.

Traditional assessment of gait in stroke survivors clearly needs more in-depth evaluation with novel strategies. Limited research has focused on the altered variability inherent in post-stroke gait, including strategies to correct the underlying dynamic

patterns. As an example of this research direction, Rhea, Wutzke and Lewek (2012) studied how a gait speed training on a treadmill in a stroke survivor group influenced gait dynamics. While the study findings did not support for increasing complexity of the hemiplegic limb joint movement with only a single session of training variable speed, it does illustrate how motion capture and gait dynamics can be used to examine the efficacy of potential gait rehabilitation programs for stoke survivors. Motion capture information is not readily available clinically and is just emerging as a resource for gait analysis, especially for clinical stroke populations. The details of motion capture are needed to bridge the gap between traditional clinical measures and rehabilitation techniques to better address the difference in fall statistics for clinical and healthy adult populations.

Testing of Balance

The physical limitations that result from a non-cerebellar stroke event are multifactorial, difficult to index, and can lead to significant limitations of physical activity afterward (Delbaere et al, 2004). In order to effectively focus rehabilitation, an appropriate test battery must be employed. The focus of this thesis was to examine commonly used balance and postural control assessment tools to see how this test battery compares to motion capture information. The tests include the (1) Timed Up and Go, (2) Berg balance assessment, (3) Functional Gait Assessment, (4) Activities-Specific Test of Balance and Confidence, (5) lower extremity strength tests, and (6) lower extremity flexibility tests.

The Timed Up and Go is a fast test that was derived from the "Get Up and Go" test, and proposed in 1991 by Podsiadlo and Richardson. The authors found that the TUG is correlated with the Berg balance assessment (r=.81) and predictive of the ability to go outside alone. The TUG instructions are to cue the subject to stand from a chair, to walk three meters and turn to come back and sit down. The cut-off score for fall risk has been determined to be 14 seconds, although some authors propose ten to 12 seconds (Alexander, 2005). The TUG is an indirect measure of gait speed, which is a predictor of falls in multiple studies (Harada et al; Dickstein, 2008; Kempen et al, 2011). The test is appropriate for older adults (65-95 years), community dwellers, and stroke survivors, but not for cognitively impaired elderly (Hayes and Johnson, 2003). Therefore, a test establishing a cognitive threshold would need to be included with this measure to ensure validity.

The Berg balance assessment is a fourteen item test that includes many challenging tasks, including a functional reach, a single leg balance task, a step-up task, and rotational movements to both look over the shoulders and turn 360 degrees in each direction. The test was proposed in 1992 by Berg, Wood-Dauphinee, Williams and Maki, and was used to follow 70 stroke patients for a year. The study results showed that the test moderately correlated to self-review, caregiver ratings, and laboratory measures. These items are considered to have good inter- and intra-rater reliability, but takes ten to twenty minutes to complete (Mancini and Horak, 2010; Salzman, 2010). The protocol instructions are descriptive to help with the most accurate choice of scores for each item

(Berg, Wood-Dauphine, Williams and Maki, 1992). The Berg test is valid for post-stroke use, as well as for older community dwelling adults (Hayes and Johnson, 2003).

The Functional Gait Assessment (FGA) is derived from the Dynamic Gait Index (DGI). The FGA offers ten items that reflect a higher physical challenge than the DGI, including gait with eyes closed and backward gait. The intra-rater reliability is not as good as inter-rater, due to suspicions that the performance of the test items varies within the same subject (Wrisley, Marchetti, Kuharsky and Whitney, 2004). However, Wrisley et al have validated the test for internal consistency at .79. Limited feedback is provided in the instructions for those administering the test. The instructions for subjects are concise, directing scoring of the items (Wrisley and Kumar, 2010) and the total score of the FGA is 30 versus only 24 points for the DGI. This greater range and depth of difficulty of test items provides a valid and reliable test, according to Wrisley and Kumar, which relates to fall risk. Additionally the authors report a correlation between the Berg and Activities-Specific Test of Balance and Confidence (ABC) scale for prediction of falls. The ABC scale is considered to be a good predictor of falls in the elderly, with a correlation to fall risk, as fear of falling is a strong fall predictor (Herman et al, 2005).

The ABC scale is self-rating of confidence in maintaining balance with progressively more challenging daily skills; from walking in the house to icy outdoor terrain (Powell and Myers, 1995). The rating is from 0% (no confidence) to 100% (completely confident) that the challenge is manageable. The ABC scale is used for determining whether an individual is fearful of certain activities, since self-limiting can lead to physical disuse decline (Boulgarides, McGinty, Willet and Barnes, 2003).

Additionally the testing is valid for community-dwelling older adults and has excellent reliability for test/retesting (Westlake, 2007).

Since many of the aforementioned tests are not valid for cognitively impaired elderly, it is important that minimum performance in a cognitive test be used as inclusion criteria for balance testing studies. The Mini Mental Status Examination is designed to be a test of cognitive function for establishing a baseline for memory, orientation and praxis (Trueblood, 2010). While the test doesn't measure gait sequencing, it does offer a measure of quantifying recall and sequencing of a non-novel skill (Trueblood, 2010; Salzman, 2010). The Mini Mental has been used extensively in elderly populations to quantify cognitive function and correlates well with the Minimum Data Set, which is the Medicare standard assessment tool for residents in skilled nursing care (Hartmaier et al, 1995).

While balance skills and measures of confidence, could paint a vivid picture of potential areas of fall risk for an older adult, certain physical limitations must also be accounted for in the significance of these measures. Both strength and flexibility in the legs will alter or enhance movement (Dean et al, 2004; Horlings et al, 2008) depending on the values as compared to normal populations. Compensation for deficits will alter gait mechanics, such as the specific dropping of the hip with gluteus medius weakness on the opposite leg (Hoppenfeld, 1976). Accurate testing for gait changes should include the physical assessment of strength and flexibility.

Lower Extremity Strength and Flexibility

The standard for strength testing historically has been manual muscle testing, which originated with two orthopedic surgeons in 1912 (Hislop and Montgomery, 2007). The doctors, Wilhelmina Wright and Robert W. Lovett, developed the first gravity-based testing system that graded from zero (no discernible muscle contraction) to six (normal strength). These tests were developed at that time for use on post-polio populations first, mainly by the physicians, as the field of physical therapy did not exist until around 1913 (American Physical Therapy Association, 2013). Doctor Wright served as the first physical therapist at that time and the testing processes she used are quite similar to the testing protocols in current books (Hislop and Montgomery, 2007). The grading system taught by physical therapy programs in the United States utilizes a zero to five rating, with zero being no visible or palpable contraction of a muscle to five being a full effort against maximal resistance and through full joint movement against gravity (Clarkson, 2000). The challenge presented by muscle testing lies in the rater reliability: while the testing from zero to three involves only a volitional effort, the testing from three to five involves a judgment by the therapist as to the degree of resistance offered (Hislop and Montgomery, 2007). The test limitation of maximal effort also implies the tester can exert a greater force than the tested patient. This means the level judged as five or normal maximal strength doesn't compete with the tester's maximum strength (Lunsford and Perry, 1995), which calls into question the validity of "five". Further complicating the process is that male and female patients have strength capabilities influenced by

anthropometrics, as well as age influences on strength (Clarkson, 2000). The original testing processes were created to quantify strength losses in clinical populations, and therefore alternate positioning for various levels of weakness were necessary (Hislop and Montgomery, 2007). All these factors make uniform quantification of absolute strength of any muscle group difficult. The need for a more accurate and valid system to test strength in a repeatable way has led to other systems of strength measurement.

Numerous studies of strength testing using hand-held dynamometers in comparison with standard manual muscle testing are available in the literature. The use of standard muscle testing positions with both types of testing doesn't always occur (Bohannon, 2007; Dean, Kuo and Alexander, 2004) and may involve alternate positions, such as a supine hip flexion test versus the standard seated positioning with the hip and knee bent to 90 degrees (Clarkson, 2000). The observation is made that the rater experience is helpful for testing, to be sure the positioning for testing eliminates the substitution of stronger muscles (Hislop and Montgomery, 2007). Stabilization of the start position insures that consistent testing can be done (Bohannon, 2007; Orqvist et al, 2007) and the ability to reliably test a post-stroke population with hand-held dynamometers to look at the affected and unaffected sides has been demonstrated (Kluding and Gajewski, 2009). When comparing strictly manual muscle testing to dynamometers, the observation has been made that dynamometers are useful for the grades of strength above three, but three and below are better tested manually in children (Fosang and Baker, 2006). In adults however, the two systems compare well with accuracy evident in repetitive testing of adults with normal function (Bohannon, 2007)

and adults post-stroke (Svantesson et al, 2007), providing that joint position is stable for all testing interventions. Cooper et al (2011) showed that a 50% loss of strength was present in the affected side relative to the unaffected side in stroke survivors.

Normative values for adults vary with decades, with aging leading to a loss of strength. Bohannon's study of six decades of life (1997) covers a range of strength numbers with respect to age. Non-side specific numbers for knee flexion were noted in Newtons for Danneskiold-Samsoe et al (2009). The numbers for these studies are as follows:

Table 1. Dynamometer Data in Newtons with Dominant/Non-Dominant Side Values Listed Where Available

	Age 40-49	Age 50-59	Age 60-69	Age 70-79
Hip abduction				
male	311/321	308.9/303.6	261.4/258.9	250.8/246.0
female	218.4/201.5	214.8/207.4	172.3/164.2	152.7/147.1
Knee flexion				
male	129.0	134.0	108.0	93.0
female	80.7	72.8	62.7	56.7
Knee extension				
male	583.0/588.9	470.9/467.7	386.9/376.5	360.3/365.9
female	363/380.6	334.7/318.7	273.6/265.9	210.1/204.7

While the table data is helpful for determining normal dynamometer values, minimal literature is available with normal plantarflexion and hip extension values for any ages. The test positions for muscle strength are standardized in existing literature to obtain dorsiflexion in a neutral extended position for the hip, knee and ankle in neutral dorsiflexion; for hip abduction in neutral hip and knee extension, for knee flexion and

extension with hip, and knee in 90 degrees flexion position with arms in the lap (Andrews, Thomas and Bohannon, 1996; Bohannon, 1997). Logically, plantarflexion could be obtained in the same position as dorsiflexion, but hip extension requires more consideration. The standard test for fair or greater strength is in prone with the knee extended (Hislop and Montgomery, 2007; Clarkson, 2000; Kendall, 1993) and also the position for aligning the goniometer axes on the axillary line of the trunk and on the femur pointing at the lateral epicondyle (Clarkson, 2000). The alternate position may be needed for the elderly, as prone may be difficult to attain with stroke survivors who have challenges in the upper extremity. Clarkson's (2000) alternate method is to have the individual lean forward and prop on the table, and extend the hip from the standing posture.

The muscle testing positions are important to measure strength with hypertonic muscles from stroke events. In a study by Gregson, Leathley, Moore, Smith, Sharma and Watkins (2000) inter-rater reliability was assessed for strength testing of 35 stroke patients in multiple care settings. The authors used a fixed position of sitting in a chair and with flexed lower extremity joints for testing of the hip, knee, and ankle. The result of the study was that the standard testing position was reliable for assessing power, even for those subjects with muscle tone issues, although some difficulty in assessing plantarflexion tone did occur between raters. In a study of dynamometer strength testing by Bohannon (1996), one subject who had sustained a known stroke event just prior to testing the fixed postures showed that strength data was reliable. In a classic reference on stroke rehabilitation, Bobath (1990) reports that positioning to reduce muscle tone in a

limb helps reveal underlying function. With the concept of inhibition, a limb is positioned to reduce excessive tone interference, such as flexing the hip and knee to observe hypertonic plantarflexion. The breaking up of a motor extensor or flexor pattern is possible with combined flexed and extended joints in standard muscle testing positions.

The standard test for goniometric position of dorsiflexion is supine with the hip extended and knee flexed 20 degrees with a towel roll, and would allow for strength testing also of plantarflexion (Clarkson, 2000). The arms of the goniometer per Clarkson would line up along the sole of the foot and posterior to the lateral malleolus.

Comparison of the stroke and healthy older adults regarding lower extremity strength and flexibility would help identify which deficits might impact the motion capture data. The loss of flexibility necessary to take a full step or strength losses that decrease the propulsion forces will help explain the differences in gait. If the clinical gait measures are fairly close, yet gait appears quite different, more in-depth metrics from motion capture might be needed.

Summary

The literature describing gait and balance deficits in stroke survivors has limitations that need to be addressed if evidence-based practice is to be achieved. Most importantly, a complete motion-capture based description of gait for both healthy older adults and a post-stroke population should be described and compared with non-linear analyses. This provides a quantitative base for comparison of specific improvements from rehabilitation, since the underlying patterns are the "gold standard" for determining

the health or adaptability of gait. Current literature contains both shorter trials of gait with motion capture information, and longer trials with limited analysis. The complete analysis of both hemiplegic and healthy lower extremities, in comparison with healthy adaptable adults, is also lacking in literature. Additionally, there are no complete comparisons of motion capture information to clinical assessments. Since confidence of falling is related to fall risk, it would be advisable to use the ABC scale to search for common threads with motion capture and other clinical test. In short, the basic science of gait deficits of stroke survivors is not fully informed and needs to be better defined to increase the effectiveness of rehabilitation. Physical therapy cannot fix what has not been defined as a change in normal function, but with specific parameters for measurement, the treatment provided after a stroke can be more effective, easier to assess progress, and more "answerable" to insurance providers for enhanced reimbursement.

CHAPTER III

OUTLINE OF PROCEDURES

Participants

Twenty healthy elderly (63.4±8.9 years, 10 male and 10 female, 173.9±9.3 cm, 81.0 ±15.9 kg) and 7 non-cerebellar/non-brain stem stroke survivors (57.6±7.7 years, 5 male and 2 female, 170.1±6.4 cm, 84.2±13.2 kg) were recruited to participate. The average time since the stroke event in the stroke survivors was (36±25.5) months. Four of the stroke survivors were right side involved. The UNCG IRB approved all study procedures and all participants signed a consent form. Exclusion criteria included: anyone taking narcotic medication or anti-seizure medication, blood pressure measures above 150/100 or 90/50, and lower than 90% oxygen saturation.

Instrumentation

The gait dynamics data were collected using Qualysis motion capture cameras (Gothenburg, Sweden) while participants walked on a Simbex Active Step treadmill (Lebanon, NH). The data collected in Qualysis software was resolved for landmark labeling, and then Visual 3D software (C-Motion, Germantown, MD) was used to import the data to organize data sets that include measures of step length, step width, step time, stride time of the affected limb and stride time of the unaffected limb. The final calculations of data were done in Matlab (MathWorks, Natick, MA) to compute

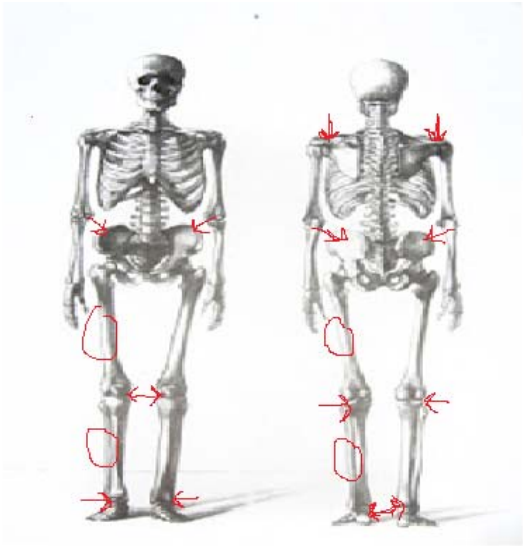
detrended fluctuation analysis and sample entropy. Excel software were used to calculate the mean, SD, and CoV of the variables. The strength testing for hip extension, hip abduction, plantarflexion, quadriceps and hamstrings muscle groups were collected with a hand-held dynamometer (Lafayette Industries, Lafayette, IN). Lower limb flexibility was measured with a clinical standard goniometer (Elite Medical Instruments, Fullerton, CA).

Procedure

The 27 participants provided a healthy and stroke (if applicable) medical history. All clinical tests were assessed first in the following order: (1) TUG, (2) Berg, (3) FGA, and (4) ABC. Next, the Mini Mental Status Examination was given to assess cognitive performance. Strength measurements were then made in kilograms with a hand-held dynamometer to measure the gastrocnemius/soleus group, hamstrings, hip abductors, hip extensors and quadriceps group. The ankle and hip abduction strength measures were done from supine and sitting for the knee strength, and either supine or standing (alternate) for hip extension. Active assisted range of motion of the ankles and hip extension were then measured. Next, the anthropometric data were collected and then retro-reflective markers were applied to the participants' body. A total of 36 markers were used and placed on the shoulders, anterior superior and posterior superior iliac crests, the thigh and shank segment panels, medial and lateral knee and ankles, medial and lateral metatarsophalangeal joints, and calcaneus laterally. The marker locations are

illustrated with red arrows on the modified skeletal designs in Figure 7 (altered from JoBSPapa.com).

Figure 7. The Locations of Panels and Individual Retro-Reflective Markers for Data Collection



The subjects wore a non-weight supporting treadmill harness attached to an overhead support for safety. The harness was loose enough to avoid interference with normal gait dynamics, yet tight enough to catch the subject if a trip were to occur. A trial of static in information of 10 seconds was collected for creating the Visual 3D model, followed by a 20 second trial of walking to assist the Qualysis in identifying the markers. Next, the participants selected their walking speed by telling the researchers to increase or decrease the treadmill speed until they were comfortable. Next, the gait trial lasted for 10 minutes while the participants walked at their self-selected speed while the retro-reflective markers were recorded at 200 Hz. The gait was performed without the use of a handrail or an assistive device of any kind.

Data Collection and Analysis

Data from the TUG, Berg, FGA, and ABC were recorded in the units denoted by each test. For the strength measures, a global measure of lower extremity strength for each side was created by taking the average of the strength values from the five muscle groups on each limb. Similarly, the range of motion scores were averaged for the ankle and hip joints on each limb to produce a side-specific global measure of lower extremity flexibility. The data from QTM was exported to Visual3D to create the following five time series: (1) step length, step width, step time, stride time of the affected limb, stride time of the unaffected limb. Matlab was then used to calculate DFA and SampEn, and Excel was used to calculate the mean, SD, and CoV.

To address hypotheses 1-3, a separate MANOVA for each analysis (mean, SD, CoV, DFAα and SampEn) was run, with group (stroke or healthy) as the independent variable and gait metrics (step length, step width, step time, affected stride time, and unaffected stride time) as the dependent variable. To address hypothesis 4, one MANOVA was run, with group (stroke or healthy) as the independent variable and the clinical metrics (TUG, Berg balance, FGA, ABC, affected side strength, unaffected side strength, affected side flexibility and unaffected side flexibility) as the dependent variables. To address hypothesis 5, Spearman's rho was used to examine the correlation of each gait metric to each clinical metric. For significant MANOVA tests, follow-up ANOVA's were run to examine group differences within each dependent variable. Statistical significance was set at $p \leq .05$ for all tests.

CHAPTER IV

RESULTS

Group Differences in Gait Metrics

Self-selected walking speed was significantly different between groups, $t(25)=5.36$, $p<.001$, with the healthy elderly adults walking significantly faster (0.88 ± 0.22 m/s) compared to the stroke survivor group (0.36 ± 0.22 m/s). Although time was controlled for during the walking test (10 minutes), the faster walking speed of the healthy elderly led to a significantly greater number of strides taken (490.6 ± 44.8) compared to the stroke survivor group (361.9 ± 104.4), $t(25)=4.55$, $p<.001$.

The MANOVA for mean gait metrics revealed significant differences between groups, $F(5,21)=6.79$, $p=.001$. Follow up ANOVAs revealed that the mean of all dependent variables was different between groups (Table 2). Specifically, the stroke survivors exhibited a shorter mean step length, a greater mean step width, a longer mean step time and a longer stride time for both the affected and unaffected limbs (Figure 8).

The MANOVA for SD also revealed significant group differences, $F(5,21)=12.4$, $p= <.001$. Follow-up ANOVAs revealed that the SD of all dependent variables was different between the groups except for mean step width (Table 2). Specifically, the stroke survivors had a higher SD in step length, step time, affected stride time, and unaffected stride time (Figure 9).

Figure 8. Mean Values for each of the Gait Metrics with Standard Error Bars

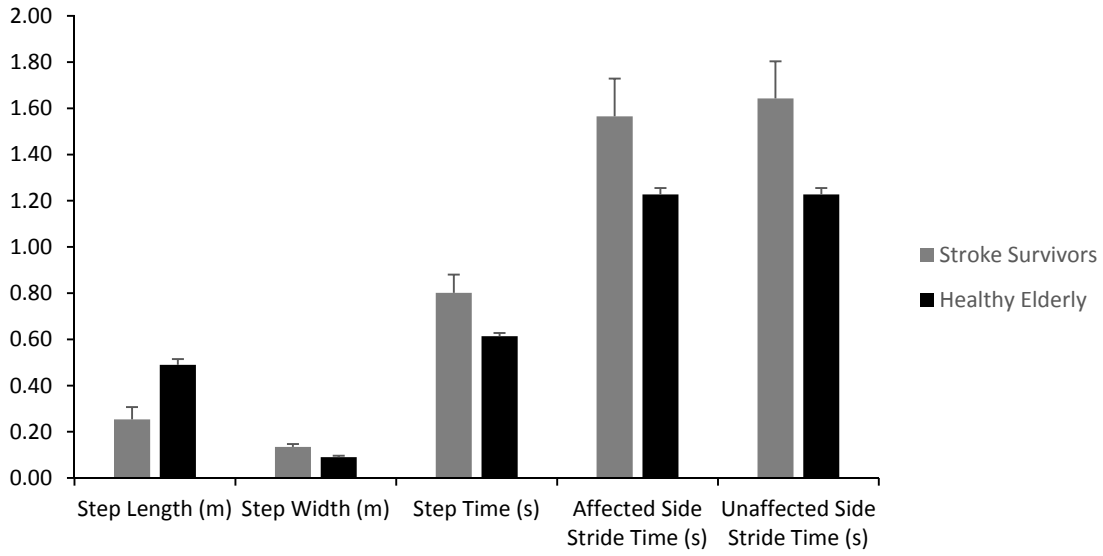


Figure 9. SD Values for each of the Gait Metrics with Standard Error Bars

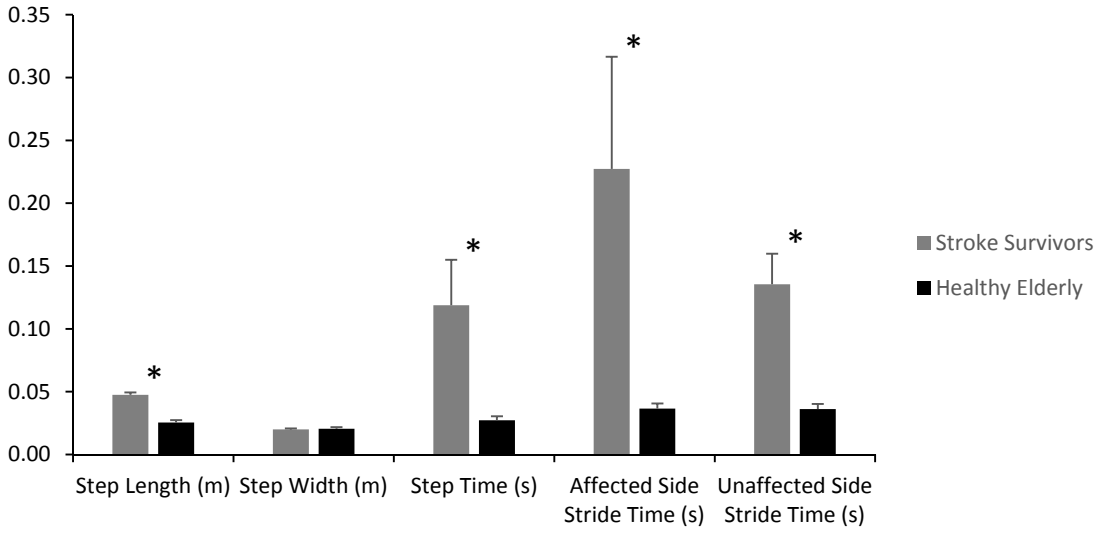
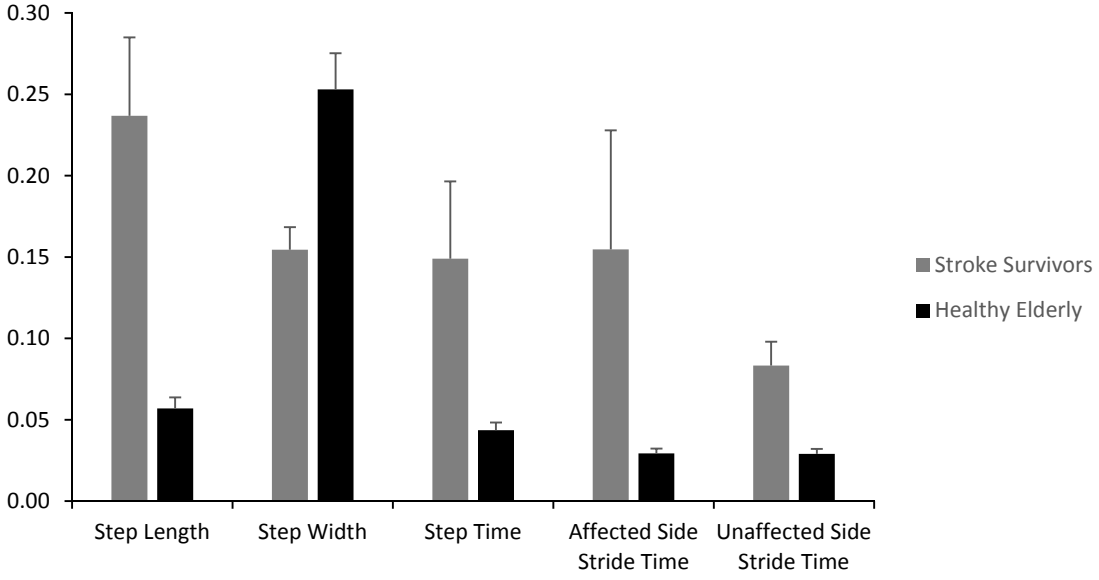


Figure 10. CoV Values for each of the Gait Metrics with Standard Error Bars



The MANOVA for CoV showed that the groups were different, $F(5,21)=7.73$, $p=<.001$. Follow-up ANOVAs demonstrated significant differences for all dependent variables (Table 2). Specifically, the stroke survivors had a higher CoV in step length, step time, affected stride time, and unaffected stride time, along with a lower CoV in step width (Figure 10).

The MANOVA for SampEn did not reveal any group differences, $F(5,21)=1.43$, $p=.25$ (Table 2, Figure 11). However, MANOVA for DFA α did show group differences, $F(5,21)=3.66$, $p=.015$. Follow-up ANOVAs showed significant difference only in step length (Table 2), with stroke survivors exhibiting a lower DFA α (more random) step length (Figure 12).

Figure 11. SampEn Values for each of the Gait Metrics with Standard Error Bars

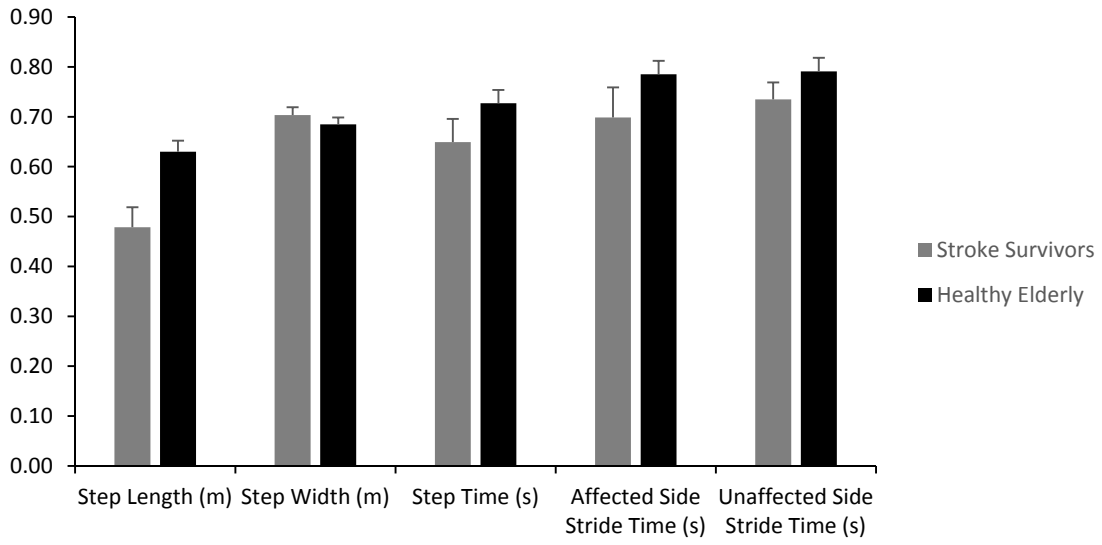
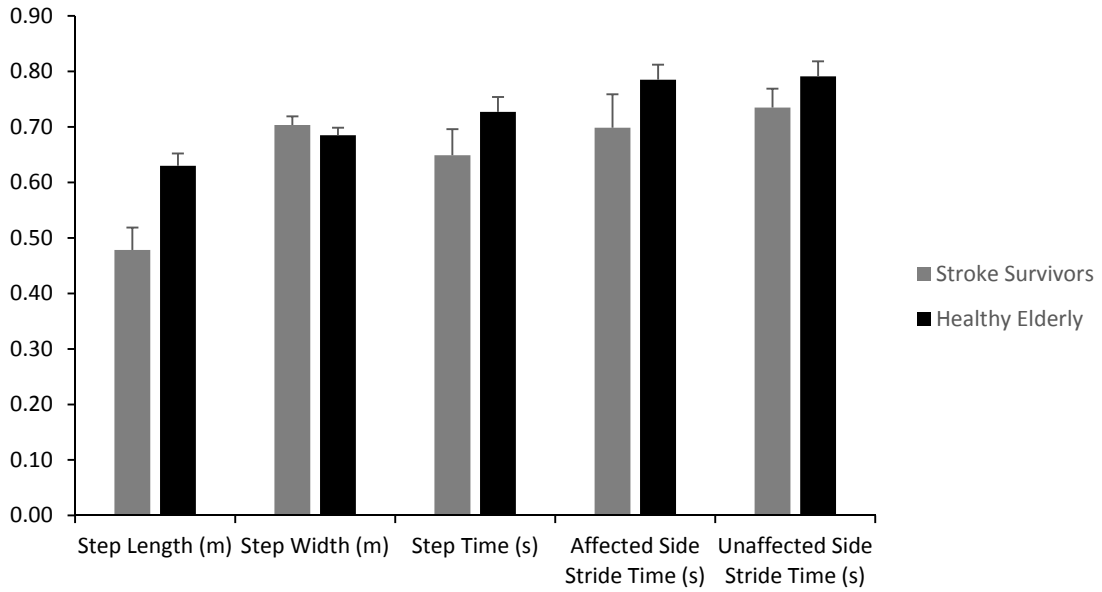


Table 2. Between Subjects Statistics for each Dependent Variable Within each Gait Metric

Metric	Variable name	df	F	p value	partial eta squared
Mean	Step Length	1,25	21.55	< .001	0.463
	Step Width	1,25	12.16	0.002	0.327
	Step Time	1,25	14.80	0.001	0.372
	Affected Stride Time	1,25	11.30	0.003	0.311
	Unaffected Stride Time	1,25	18.10	<.001	0.420
Standard Deviation	Step Length	1,25	40.20	< .001	0.616
	Step Width	1,25	0.070	0.800	0.003
	Step Time	1,25	21.90	< .001	0.467
	Affected Stride Time	1,25	16.50	< .001	0.397
	Unaffected Stride Time	1,25	47.50	< .001	0.655
Coefficient of Variation	Step Length	1,25	42.36	<.001	0.629
	Step Width	1,25	6.209	0.020	0.199
	Step Time	1,25	16.322	<.001	0.395
	Affected Stride Time	1,25	10.546	0.003	0.297
	Unaffected Stride Time	1,25	35.038	<.001	0.584
SampEn	Step Length	1,25	0.132	0.720	0.005
	Step Width	1,25	5.226	0.031	0.173
	Step Time	1,25	0.026	0.872	0.001
	Affected Stride Time	1,25	2.024	0.167	0.075
	Unaffected Stride Time	1,25	0.002	0.965	0.000
DFA α	Step Length	1,25	12.30	0.002	0.330
	Step Width	1,25	0.552	0.464	0.022
	Step Time	1,25	2.233	0.148	0.082
	Affected Stride Time	1,25	2.275	0.144	0.083
	Unaffected Stride Time	1,25	1.258	0.273	0.048

Figure 12. DFA α Values for each of the Gait Metrics with Standard Error Bars



Group Differences in Clinical Metrics

The MANOVA for the clinical metrics did show significant group differences, F (8,18)=5.49, p=0.001. Follow-up ANOVAs showed group differences in all clinical metrics except for all variables except unaffected side strength and unaffected side flexibility (Table 3). Specifically, the stroke survivors had longer TUG times, lower Berg scores, lower FGA scores, lower ABC scores lower affected side strength, and lower affected side flexibility (Figure 13).

Relationship between Gait and Clinical Metrics

All significant correlations are presented in Table 4. Of the 200 possible correlations, only 64 were significant. Interestingly, SampEn and DFA α, two metrics that are purported to measure functional ability, showed little or no correlation with the clinical metrics.

Figure 13. Mean Values for each of the Clinical Metrics with Standard Error Bars

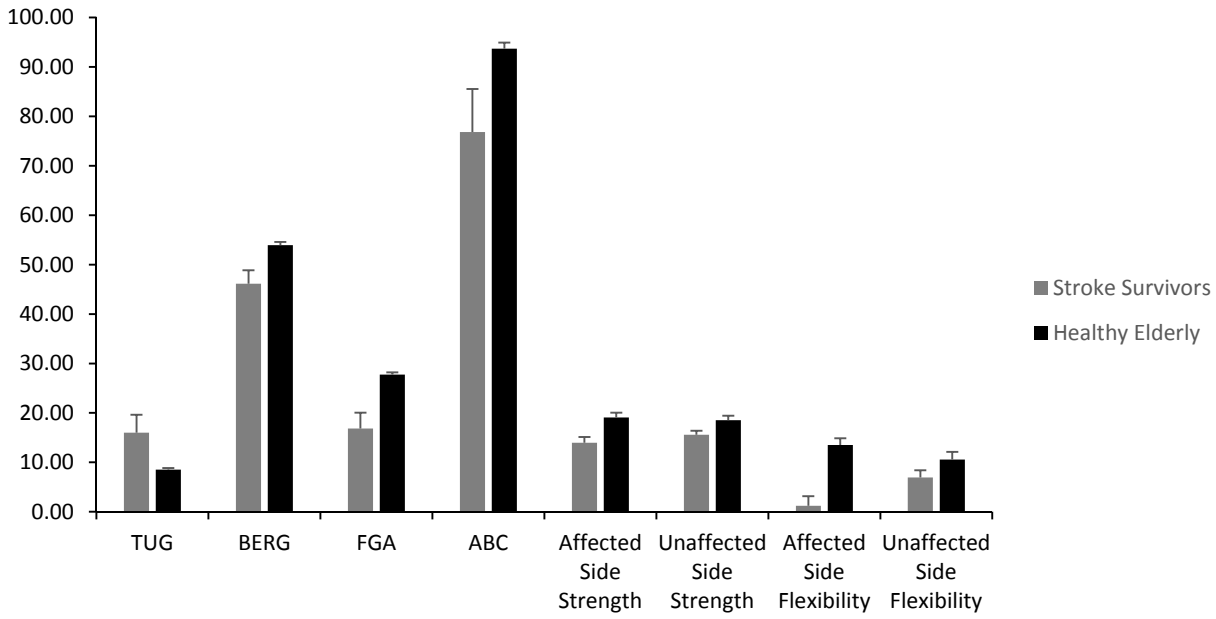


Table 3. Between Subjects Statistics for each Clinical Metric

Variable Name	df	F	p-value	part eta squ
TUG	1,25	13.752	0.001	0.355
BERG	1,25	18.732	0	0.428
FGA	1,25	29.889	0	0.545
ABC	1,25	11.365	0.002	0.313
Aff_strength	1,25	8.261	0.008	0.248
Unaff_strength	1,25	3.465	0.074	0.122
Aff_flex	1,25	20.267	0	0.448
Unaff_flex	1,25	1.76	0.197	0.066

Table 4. Correlations Among Gait Metrics and Clinical Variables.

metric	Dependent variable	TUG	Berg	FGA	ABC	Strength Affected limb	Strength Unaffected limb	Flexibility Affected limb	Flexibility Unaffected limb
mean	step length	-0.604	0.757	0.741	0.616	0.445		0.556	
	step width							-0.505	
	step time		-0.498	-0.383				-0.588	-0.438
	affec stride time		-0.395					-0.507	-0.477
	unaffec stride time		-0.522	-0.404				-0.612	-0.399
Stdev	step length	0.460	-0.59	-0.634	-0.643			-0.726	
	step width								
	step time	0.574	-0.726	-0.728	-0.611			-0.601	
	affec stride time	0.576	-0.701	-0.717	-0.714			-0.683	
	unaffec stride time	0.596	-0.709	-0.735	-0.701			-0.645	
CoV	step length	0.55	-0.702	-0.684	-0.686			-0.675	
	step width								
	step time	0.563	-0.694	-0.743	-0.62			-0.543	
	affec stride time	0.625	-0.706	-0.727	-0.744			-0.657	
	unaffec stride time	0.608	-0.677	-0.731	-0.741			-0.638	
SamEn	step length								
	step width								
	step time								
	affec stride time								
	unaffec stride time								
DFA	step length		0.466			0.382			
	step width								
	step time		-0.427	-0.456					
	affec stride time		-0.481						
	unaffec stride time		-0.533						

CHAPTER V

DISCUSSION

It is important to note that the groups were both functioning in a community-dwelling situation, and both were able to walk without any assistive device for the purpose of being on a treadmill (challenging without a handrail). The stroke survivors were mostly employed people who had resumed driving and are out in the community for activities and personal interests, despite the clearly identified deficits remaining of the stroke events. The participants who were stroke survivors had sustained only one event, and were all demonstrating cognitive competency for participation, had acceptable blood pressure and oxygen saturation measures, and were competent for making decisions. This makes the argument that the high functional level of the stroke survivors might make the lack of differences between groups a reasonable expectation due to recovery.

Findings from Gait Data

Hypothesis 1 stated that the stroke survivors would exhibit greater mean values in the gait variables of interest. In general, this hypothesis was supported. As expected, the mean step length was shorter, the mean step width was greater, the mean step time was longer, and the mean affected and unaffected stride time was longer for the stroke survivors. Findings from previous literature support these results (Hausdorff & Alexander, 2005; Winter & Eng, 1995; Herman et al, 2005), with the expectation that

neurological change (degrading of the neurological input) influences the difference in mean gait metrics in stroke survivors. Since less neural information is available to guide the motion of stroke survivors, this strategy is likely adopted to increase stability during ambulation. Herman et al (2005) use the term "cautious gait" to describe this strategy in patients with high-level gait disorder, which is also coupled with reduced gait velocity. Although this gait strategy is adopted to increase stability, stroke survivors fall at a rate roughly 2.3 times higher than older adults who haven't suffered a stroke, highlighting the extraordinary challenge of gait control following a stroke (Wrisley & Kumar, 2010; Lord, Sherrington & Menz, 2001). From the stroke survivor's perspective, the change in their gait that ultimately results in a reduction in gait velocity is a significant contributor to the reduction in their quality of life (Bowden et al, 2008; Dickstein, 2008).

Since the neural pathways following a stroke are interrupted, it was also hypothesized that the magnitude of variability (assessed via the SD and CoV) would be greater for the stroke survivor group (hypothesis 2). This finding would indicate less consistent control of the gait cycle during treadmill walking. The SD of step length, step time, affected and unaffected stride times were all greater in the stroke survivor group. Previous study findings for gait variability are in agreement with this study (Winter and Eng, 1994; Herman et al, 2005). Herman et al (2005) controlled for co-morbidities and reported increased timing variability in the gait cycle in patients with high-level gait disorder. Mizuike, Ohgi, & Morita (2009) suggested that increased variability in the gait patterns of stroke survivors was due to a reduction in the degree of freedom available to complete the task, leading to a more rigid and variable pattern.

SD of step width, surprisingly, was not different between groups. This could be due to the fact that the motion capture space used for collecting the gait metrics was located on a standard width treadmill belt using a fixed belt speed (albeit self-selected). The fairly confined parameters of the collection may have contributed to less variation in the gait metrics relative to overground gait. This is especially true for the step width metric, as there is less room for error in the direction associated with step width (medial-lateral) compared to the direction of all the other gait metrics (anterior-posterior). This is an important finding for the clinical community. Step width is typically described as a measure of stability, and has been linked to fall risk (Vaughan et al, 1999; Hausdorff & Alexander, 2005). Thus, measuring step width on a treadmill may confine the patient's gait motion so that the researcher or practitioner does not get a valid measure of the patient's step width, which could lead to a mischaracterization of their gait ability. Future research should compare the gait metrics of this study to overground walking with a stroke survivor group to better identify the potentially constraining effects of treadmill walking.

Hypothesis 3 stated that the structure of variability (assessed through SampEn and DFA α) would be different between the groups. Specifically, it was expected that the stroke survivor group would have lower SampEn and DFA α values compared to the healthy group. In general, this hypothesis was not supported. No differences in SampEn were observed and only DFA α of step length was different between groups, with the stroke survivors exhibiting a lower DFA α (i.e., more random structure of variability). The lack of support for this hypothesis could be a result of the length of data for the gait

collection (Yentes et al, 2012), in which the authors caution against using shorter data sets, especially in relation to gait variables. The motivation of this study was to extend the data collection time to 10 minutes so that an appropriate number of data points could be recorded for both groups. Previous work has suggested that a minimum number of strides to accurately characterize DFA α is 600 (Damouras, Chang, Sejdic & Chau, 2009), while no such guidelines have been suggested for SampEn. The healthy elderly group in the current study walked at a greater speed (0.88 ± 0.22 m/s) compared to the stroke survivor group (0.36 ± 0.22 m/s). This led to a greater number of strides taken during the 10 minute walking test by the healthy elderly group (490.6 ± 44.8) compared to the stroke survivor group (361.9 ± 104.4). However, both of these mean stride numbers are less than the guideline for an accurate characterization of DFA α. Although our DFA α values are near the previously reported values (~0.75), the lack of group differences could be attributed to not having enough strides in each group to fully characterize their gait behavior with this metric. Further, the use of a treadmill may have created a constrained environment, forcing both groups to produce a similar structure of variability. Previous work showed no significant differences in the structure of variability in stride time in young healthy adults when comparing treadmill walking to overground walking (Chang, Shaikh & Chau, 2008). However, an older adult population, regardless of clinical pathology, may walk more cautiously on a treadmill compared to overground, which may be a plausible explanation for our lack of group differences in the structure of variability metrics. This postulate warrants further empirical examination. It is noted that in overground gait, adults with high-level gait disorder had a significantly lower DFA α of

stride time relative to controls (Herman et al, 2005), suggesting that a less constraining environment may be more appropriate to characterize structure of variability metrics in elderly and clinical populations.

Finally, the treadmill used for this study did not have a handrail for support of the participants in the study. According to Chang et al (2008), the use of a handrail provides sufficient assistance as to elevate the DFA α to a higher, less random value. This potential interference with valid results was avoided, lending credibility to the study method and therefore was considered to be a strength of the study protocol.

Findings from Clinical Variables

Hypothesis 4 states that differences would be observed between groups in the clinical metrics (assessed via the TUG, Berg balance, FGA, ABC, strength and flexibility). Specifically, the stroke survivors were expected to have higher TUG scores and lower Berg balance, FGA, ABC, strength, and flexibility scores. This hypothesis was supported for all of the variables except strength and flexibility of the unaffected side of the stroke survivors compared to matched limbs of the healthy older adults. The results and interpretation for each clinical metric are outlines below.

Timed Up and Go (TUG)

The group difference observed for the TUG supports the findings in previous literature showing that clinical populations display lower scores (Boulgarides et al, 2003; Hayes & Johnson, 2003). A score higher than 10 seconds in the TUG has been related to

a relatively higher risk of a fall. The healthy elderly group in my study scored (8.5 ± 1.4 sec), indicative of a low risk of falling. However, the stroke survivors had a significantly higher TUG (16.0 ± 9.0 sec), indicating a functional difference in their gait patterns. This finding may help explain previous research that shows stroke survivors fall at a higher rate than healthy elderly adults (Powell & Myers, 1995).

Berg Balance Assessment

The Berg balance assessment (Berg et al, 1995) has been extensively cited in literature as useful and valid for stroke populations and community-dwelling elderly (Hayes & Johnson, 2003; Boulgarides et al, 2003; Steffen et al, 2002), although the original study tool was used for screening nursing home populations. The findings in the current study showed that stroke survivors scored lower (46.1 ± 7.2) on the Berg balance assessment relative to healthy elderly adults (54.0 ± 2.8), supporting previous findings (Schmid et al, 2012). A score of 43 or less has been shown to be indicative of higher fall risk (Wrisley and Kumar, 2010), which is why the Berg balance assessment is commonly used in clinical settings. Although the stroke survivor group was above the cut-off scores, they were significantly closer to the cut-off score for fall-risk relative to the healthy elderly participants, supporting our TUG findings. It should be noted that the Berg balance assessment measures balance control in primarily static postures, although most falls occur during dynamic activities (i.e., very few falls occur when standing still). The relation between the Berg static balance assessment and gait function in this study will be discussed in the "Correlation Between Gait and Clinical Metrics" section.

Functional Gait Assessment (FGA)

The FGA, which is derived from the dynamic gait index (DGI), is a clinical assessment designed to measure a patient's adaptive gait ability. The FGA has been shown to have good intra and inter-rater reliability (Wrisley et al, 2004) and a score of 22 or less of 30 points indicates an increased risk of falling (Wrisley & Kumar, 2010). The FGA has mainly been used to assess gait differences in clinical populations with vestibular challenges (Wrisley et al, 2004), but has also been used for stroke populations (Thieme, Ritschel & Zange, 2009). The FGA findings of the current study showed that the stroke patients were again in a fall-risk category, with the stroke survivors scoring 16.9 ± 8.5 and our healthy elderly group scoring 27.8 ±2.1. These results support previous results by showing that the stroke survivors scored significantly lower (Thieme, Ritschel & Zange, 2009). This was likely due to the rather complex challenges within the FGA (i.e., walking with eyes closed, walking backward, vertical and horizontal head turns with gait), some of which the stroke survivors found difficult to complete. Thieme et al (2009) note the inability to use the FGA with stroke survivors who rely on assistive devices to walk, reinforcing the use for community-dwelling adults. The degraded neurological input for the stroke survivors may account for the lower FGA scores in this study and the higher fall rates observed in previous research (Wrisley & Kumar, 2010; Lord, Sherrington & Menz, 2001). Thus, the FGA data shows that our stroke survivor population had difficulty ambulating in challenging environments.

Activities-Specific Balance Confidence (ABC) Scale

As predicted, stroke survivors scored lower (76.8 ± 23.0) on the ABC scale relative to the healthy elderly adults (93.7 ± 5.5). The ABC scale indicates a person's confidence in maintaining balance (i.e., avoiding a fall) in a series of challenging tasks (Powell and Myers, 1995). For example, test takers are asked about being able to navigate an escalator, successfully walk to the car, and safely reach overhead on tiptoes. The ABC scale has been shown to be a patient reported outcome that predicts fall risk (Herman et al, 2009; Wrisley,2004), with Herman et al (2009) showing that the item for stair climbing confidence on the ABC correlates to limiting of oneself to the performance of this task. The authors also note that 10% of fatal falls for the elderly occur on stairs.

Of importance to the current study, it has been validated in community-dwelling populations and been shown to have good test-retest reliability for assessing self-limiting behavior (Westlake, 2007). The lower values observed in the stroke survivor population provide important patient-reported outcome data that is congruent with our other gait and clinical metrics. That is, not only do the stroke survivors show biomechanical and clinical differences between groups, they also perceive their functional limitations. This finding is congruent with previous work showing that "fallers" score lower on the ABC scale relative to healthy elderly adults (Herman et al, 2009).

Strength

Lower extremity strength was measured in this study by taking the average of the strength measures from the following lower extremity muscle groups: hip extensors, plantarflexors, hip abductors, quadriceps and hamstrings. Multiple studies confirm that single-session testing of lower body strength with hand-held dynamometers is valid (Wang, Olson & Protas, 2002; Bohannon, 1997). Three consecutive measurements were made for each group of muscles on each limb, and then separated into affected limb and unaffected limb strength. It was hypothesized that the stroke survivor group would have lower strength values relative to the healthy elderly adults. The hypothesis was partially supported. The strength of the affected side (14.0 ± 3.0 kg) of the stroke survivor group was significantly lower than the matched limb of the healthy elderly group (19.1 ± 4.3 kg). However, no difference was observed between the unaffected side strength for the stroke survivor group (15.6 ± 2.1) compared to the matched limb of the healthy elderly adults (18.5 ± 3.9). These data support the finding of Horstman et al (2008) showing that there were no differences in intra-limb strength between healthy controls and stroke survivors when measuring quadriceps and hamstring strength. While no statistical difference between the unaffected limb of the stoke survivor group and the matched limb of the healthy group was observed, it is important to note that functional differences do not always reach statistical significance. It is also important to note that the measure of strength in the current study was a global lower extremity measure, as the strength of five muscle groups were a combined strength metric. Future research should focus on the

relation between weakness in specific muscle groups and gait metrics (e.g., abductors and step width).

Flexibility

A global measure of lower limb flexibility was recorded by averaging the range of motion scores of the following measures: hip extension and dorsiflexion. It was hypothesized that the stroke survivor group would exhibit a lower flexibility score relative to the healthy elderly adults. This hypothesis was partially supported. The stroke survivors had significantly lower flexibility in their affected limb (1.2 ± 5.1 deg) compared to the matched limb in the healthy older adults (13.5 ± 6.1 deg). Interestingly, no differences were observed between the unaffected limb flexibility in the stoke survivors (6.9 ± 3.9 deg) compared to the matched limb of the healthy older adults (10.6 ± 6.9). Flexibility measures of the lower extremity were important in the context of this study because previous work has shown that flexibility measures of the hip and ankle correlate with increased fall risk (Christiansen, 2007; Kerrigan et al, 2001; Dibenedetto et al, 2005). Thus, the asymmetrical flexibility exhibited by the stroke survivor group may partially account for the higher fall rate observed in this clinical population (Kerrigan et al, 2001; Dibenedetto et al, 2005). The reduced flexibility, in conjunction with the reduced strength of the affected limb may lead to a less adaptable limb when confronted with a perturbation, which could ultimately lead to a fall.

Correlation between Gait and Clinical Group Differences

Hypothesis 5 was an exploratory hypothesis to examine the relationship between the gait and clinical metrics. The motivation behind this analysis was to determine if gait metrics that are commonly used to objectively measure functional gait ability relate to clinical metrics that subjectively index functional gait ability. The twenty-five gait metrics (mean, SD, CoV, SampEn and DFA α of step length, step time, step width, affected limb and unaffected limb stride time) were compared to the eight clinical metrics (TUG, Berg balance, FGA, ABC, affected side strength, unaffected side strength, affected side flexibility and unaffected side flexibility), leading to a total of 200 matched variables (Table 4). Sixty-four matched variables were found to be significantly correlated.

For the variables of gait of mean step width, mean step time, and mean stride time of both limbs of stroke group participants, there was a negative correlation with affected limb flexibility. A greater mean step width, mean step time, and affected/unaffected stride times were associated with lower flexibility in the affected limb. This is congruent with previous literature that connects gait changes to flexibility of both hip extension and ankle dorsiflexion (Kerrigan et al, 2005; Christiansen, 2007). The correlation is positive for mean step length and affected limb flexibility: greater flexibility was related to greater mean step length. These findings mesh with the aforementioned studies (Christiansen, 2007; Kerrigan et al, 2001; Dibenedetto et al, 2005). Stride length was found to increase in these studies as hip extension increased, but the results were all

obtained on short walkways overground. Collectively, these studies show that a common clinical metric (lower extremity flexibility) influences gait control.

The measure of strength for the affected side of the stroke survivor group had positive correlation to mean step length and DFA α of step length. In a study by Mulroy et al (2002), the observation was made that overground gait in a 6 month post-stroke population had hip extension strength losses that were related to slow walking speeds. Gaviria et al (1995) related reduced stride length and reduced gait speed of overground gait to strength loss at the ankle in plantarflexion. Gait speed was controlled in the current study (i.e., it was set as a constant throughout the trial) and therefore was not compared with lower extremity strength. However, the findings of this study support previous research showing that a loss of strength can affect the control of gait.

The SD and CoV of the gait variables of step length, step time, and stride time of unaffected and affected limbs were positively correlated to all the balance test measures of TUG, Berg, FGA, and ABC. This is an interesting finding, as a higher magnitude of variability in gait has been traditionally been considered a marker of dysfunctional gait control (Lipsitz, 2002; Hausdorff, 2007). Clinically, dysfunctional gait would be indicated by a higher TUG and lower Berg, FGA, and ABC scores. Thus, the only positive correlation that would be expected is between the magnitude of variability of the gait variables and the TUG, while a negative correlation would be expected between the magnitude of variability of the gait variables and the clinical metrics. Surprisingly, none of the balance tests were correlated to SD, CoV or mean of step width. This was not expected as the mean step width was significantly different between the groups, as were

the balance tests between groups. As was previously discussed, the testing format was not conducive to unlimited freedom for step width on a standard belt treadmill. A test format with overground walking may result in greater measures of both

The clinical testing of Berg balance did positively correlate to the step length finding for DFA α. This result implies that the challenge of a narrower base of support relates to a shorter step length in the context of lower stride variability, possibly to maintain step to step balance. This is congruent with findings of Hausdorff (2007) and Lipsitz (2009) showing that being a higher fall rate is related to a lower DFA α

The stroke survivor group was found to have significantly less confidence in avoiding a fall in the current study, and the ABC was found to correlate with mean step length, SD step length, SD step time, SD affected side stride time, CoV step length, CoV step time, CoV affected side stride time, and CoV unaffected side stride time. Fear of falls is considered to be a significant predictor of fall risk in the literature (Herman et al, 2005; Boulgarides, McGinty, Willet and Barnes, 2003). The findings of Herman et al (2005) showed that when comparing individuals who were older and had a high-level gait disorder (HLGD), defined as an undiagnosed condition that was linked to stride timing maladaptive fluctuations, the HLGD population had slower gait speed, changes in cadence, muscle weakness, and slower TUG scores as compared to controls subjects. Especially noteworthy was that ABC scores were significantly correlated to stride timing variability, as were found in the current study. However, Herman et al (2005) did not find a correlation between stride variability and strength or TUG, but mainly to the fear of falls and depression measures. The study concluded that fall risk may be mainly due to

self-limiting mobility over the concerns, leading to further debility. In the current study, we show that the ABC is correlated to the TUG, Berg, and the FGA. It is interesting to note that the clinical metrics and patient-reported outcomes indicated that the stroke survivor group was at a higher risk of falling due to dysfunctional gait control. However, the gait metrics that have been reported to measure a person's functional ability during gait (DFA α and SampEn) did not pick up on any differences between the groups (sans DFA α of step length).

DFA α was significantly correlated to only a few clinical metrics. However, the mean number of total strides taken by the healthy elderly group (490.6 ± 44.8) and stroke survivor group (361.9 ± 104.4) violates the guideline established by Damouras, Chang, Sejdic and Chau (2009) that suggested a minimum of 600 strides for accurate characterization of DFA α. No such guideline for SampEn exists in literature. The use of a treadmill for data capture was previously established as a comparable task to overground walking for young healthy subjects with no neurological conditions (Chang, Shaikh & Chau, 2009). The study did not have a representation of older subjects, nor did it include clinical populations. This limitation of the treadmill study may mean that the task is genuinely confining the older populations who may be more self-limited in the task given the less than perfect balance test scores for both study groups. The ABC was highly negatively correlated to the TUG (-.61) and highly positively correlated to the Berg balance and FGA (.62 and .69 respectively), showing that balance confidence was related to functional ability

The SampEn measure was not significantly correlated to any gait metrics. The finding that SD of step width was not different between the study groups when mean step width was significantly different is interesting in consideration of the lack of SampEn findings. This highlights how the magnitude of variability in gait can fluctuate without a concurrent change in the structure of variability. This finding is possibly related to the geometrical limitations of treadmill walking. Further study of overground gait in comparison to treadmill gait with a stroke survivor group would help identify how gait is controlled in each environment.

Strength measures were not significantly related with most of the variables of this study, other than being correlated to mean step length and DFA α of step length. Dean et al (2004) suggested that strength was related to fall risk by demonstrating torque and velocity losses for both hip flexion and extension over each decade of life, although no comparative balance testing results were obtained. Additionally, Dean et al (2004) studied only the right leg as the dominant limb, overlooking the left leg as a potential factor for fall risk. Further, Dean et al studied a kicking task, which might not be the most relevant to an elderly population that might potentially fall in a functional task.

A Comment on Gait Speed

Speed of gait in the current study was significantly different between the groups, as represented by self-selected walking speed in the treadmill task, the TUG times, and with the timed portions of the FGA. Speed of gait has been related to changes in gait control in the literature, with decreased speed related to weak plantarflexors, increased

stance, and increased step width from normal aging (Salzman, 2010). Winter (1991) reports increased speed being attached to decreased mean step width, increased mean stride length, and increased cadence (steps per minute). Both Salzman and Winter report decreased speed being related to shorter stride, greater mean step width, and decreased cadence. The results of this study support the findings of all the previous literature listed, although cadence was not formally calculated for the scope of this study.

Of further note in the study, the TUG timing was proportionately closer between groups in comparison to the treadmill speeds. Both times were selected by the participants, who as a group chose to walk at 0.88 m/s versus 0.36 m/s for stroke survivors, with 2.4 times faster speed on average for the treadmill. By contrast, the TUG times were 8.5 seconds on average versus 16.0 for stroke survivors, a 53% faster score for the healthy older participants. As was previously mentioned, the much slower speeds of the treadmill task may have influenced the underlying patterns, potentially affecting the metrics for structure of gait. According to Jordan, Challis and Newell (2007), the expected lower DFA values from the neurologically impaired population might be increased by the slower walking speed. This might be avoided in an overground walking task as the stroke survivors would be able to control for speed in a less constrained task walking overground. A normal walking task would eliminate the confound of potentially artificially slowing speed to create some control for the confined treadmill task.

While gait speed is attached to recovery after a stroke for functional reasons (Krasovsky & Levin, 2010; Dickstein, 2008), Krasovsky and Levin do note that increasing speed does not necessarily mean recovery of underlying deficits. Speed

increases, according to the authors, may be accomplished by sacrificing coordination of gait and incorporating abnormal patterns (circumduction of the hip, for example). Clearly speed for the sake of itself is not a goal, but rather another indicator of recovery within an appropriate context.

Final Observations for Further Study

The primary purpose for this study was to develop an understanding of the differences and similarities in healthy older adults and adults who have sustained a single stroke event affecting one side of the body. The difficulties in recruiting people who have sustained a stroke have made the numbers of participants quite uneven, and may wash out the significance of some of the details of the gait and clinical variables. Collection of the study information should continue since the hope is to contribute to evidence-based practice for physical therapy. Specifically, clinical practice lacks an understanding of which variables of gait are really involved with fall risk, what clinical tests are the most meaningful, and which therapy interventions will be the most beneficial.

The possibility exists that the lack of significant inter-group findings, such as with strength, may be a reflection of true age-related changes and not just lack of clinical subjects. Since strength and flexibility are the same for the stroke unaffected sides and healthy older adults for both limbs, the case can be made for aging as the reason. This does pinpoint areas for further study with healthy older adults to see if flexibility corrections can influence the measures of balance and fear of falls, for example.

Further, it is important to note that the groups were both functioning in a community-dwelling situation, and both were able to walk without any assistive device for the purpose of being on a treadmill (challenging without a handrail). The stroke survivors were mostly employed people who had resumed driving and are out in the community for activities and personal interests, despite the clearly identified deficits remaining of the stroke events. The participants who were stroke survivors had sustained only one event, and were all demonstrating cognitive competency for participation, had acceptable blood pressure and oxygen saturation measures, and were competent for making decisions. This makes the argument that the

Finally, the format of testing using motion capture space and a treadmill make a very limiting medium to get a true representation of neuromotor control in an unconstrained setting. Further study of post-stroke gait should include a long overground gait collection (Lewek, 2009). While this might be difficult and involve a track set-up rather than just one direction of movement, the nuances of speed fluctuations and variability of gait not hindered by a space constraint will be more evident. The collection process could involve electronic goniometers, accelerometers, EMG, which would allow for the collection of more gait variables to provide greater insight into gait control. The element of overground gait may provide a more realistic representation of the details and deficits of the stroke survivor population.

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APPENDIX A

TESTING FORMS

The Activities-specific Balance Confidence (ABC) Scale*

Instructions to Participants:

For each of the following, please indicate your level of confidence in doing the activity without losing your balance or becoming unsteady from choosing one of the percentage points on the scale from 0% to 100%. If you do not currently do the activity in question, try and imagine how confident you would be if you had to do the activity. If you normally use a walking aid to do the activity or hold onto someone, rate your confidence as it you were using these supports. If you have any questions about answering any of these items, please ask the administrator.

The Activities-specific Balance Confidence (ABC) Scale*
For each of the following activities, please indicate your level of self-confidence by choosing a corresponding number from the following rating scale:

0% 10 20 30 40 50 60 70 80 90 100%

no confidence completely confident

"How confident are you that you will not lose your balance or become unsteady

when you…

1. …walk around the house? ____%
2. …walk up or down stairs? ____%
3. …bend over and pick up a slipper from the front of a closet floor ____%
4. …reach for a small can off a shelf at eye level? ____%
5. …stand on your tiptoes and reach for something above your head? ____%
6. …stand on a chair and reach for something? ____%
7. …sweep the floor? ____%
8. …walk outside the house to a car parked in the driveway? ____%
9. …get into or out of a car? ____%
10. …walk across a parking lot to the mall? ____%
11. …walk up or down a ramp? ____%
12. …walk in a crowded mall where people rapidly walk past you? ____%
13. …are bumped into by people as you walk through the mall?____%
14. step onto or off an escalator while you are holding onto a railing? ____%
15. … step onto or off an escalator while holding onto parcels such that you cannot hold onto the railing? ____%
16. …walk outside on icy sidewalks? ____%

Berg Balance Scale
Name: Date of Test:
.

1. Sit to Stand

- Instructions: “Please stand up. Try not to use your hands for support”
- Grading: Please mark the lowest category that applies

( ) 0: Needs moderate or maximal assistance to stand
( ) 1: Needs minimal assistance to stand or to stabilize
( ) 2: Able to stand using hands after several tries
( ) 3: Able to stand independently using hands
( ) 4: Able to stand with no hands and stabilize independently

2. Standing unsupported

- Instructions: “Please stand for 2 minutes without holding onto anything”
- Grading: Please mark the lowest category that applies

( ) 0: Unable to stand 30 seconds unassisted
( ) 1: Needs several tries to stand 30 seconds unsupported
( ) 2: Able to stand 30 seconds unsupported
( ) 3: Able to stand 2 minutes without supervision
( ) 4: Able to stand safely for 2 minutes

If person is able to stand 2 minutes safely, score full points for sitting unsupported (item 3). Proceed to item 4.

3. Sitting with back unsupported with feet on floor or on a stool

- Instructions: “Sit with arms folded for 2 minutes”
- Grading: Please mark the lowest category that applies

( ) 0: Unable to sit without support for 10 seconds
( ) 1: Able to sit for 10 seconds
( ) 2: Able to sit for 30 seconds
( ) 3: Able to sit for 2 minutes under supervision
( ) 4: Able to sit safely and securely for 2 minutes

4. Stand to sit

- Instructions: “Please sit down”
- Grading: Please mark the lowest category that applies

( ) 0: Needs assistance to sit
( ) 1: Sits independently but had uncontrolled descent
( ) 2: Uses back of legs against chair to control descent
( ) 3: Controls descent by using hands
( ) 4: Sits safely with minimal use of hands

5. Transfers

- Instructions: “Please move from chair to chair and back again” (Person moves one way toward a seat with armrests and one way toward a seat without armrests) Arrange chairs for pivot transfer
- Grading: Please mark the lowest category that applies

( ) 0: Needs two people to assist or supervise to be safe
( ) 1: Needs one person to assist
(.) 2**:** Able to transfer with verbal cueing and/or supervision
(.) 3: Able to transfer safely with definite use of hands
(.) 4: Able to transfer safely with minor use of hands

6. *Standing unsupported with eyes closed

- Instructions: “Close your eyes and stand still for 10 seconds”
- Grading: Please mark the lowest category that applies

( ) 0: Needs help to keep from falling
( ) 1: Unable to keep eyes closed for 3 seconds but remains steady
( ) 2: Able to stand for 3 seconds
( ) 3: Able to stand for 10 seconds without supervision
( ) 4: Able to stand for 10 seconds safely

7. *Stand unsupported with feet together

- Instructions: “Place your feet together and stand without holding on to anything”
- Grading: Please mark the lowest category that applies

( ) 0: Needs help to attain position and unable to hold for 15 seconds
( ) 1: Needs help to attain position but able to stand for 15 seconds with feet together
( ) 2: Able to place feet together independently but unable to hold for 30 seconds
( ) 3: Able to place feet together independently and stand for 1 minute without supervision
( ) 4: Able to place feet together independently and stand for 1 minute safely

The following items are to be performed while standing unsupported

8. *Reaching forward with outstretched arm

- Instructions: “Lift your arm to 90□. Stretch out your fingers and reach forward as far as you can” (Examiner places a ruler and end of fingertips when arm is at 90□. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance toward that the fingers reach while the person is in the most forward lean position.)
- Grading: Please mark the lowest category that applies.

( ) 0: Needs help to keep from falling
( ) 1: Reaches forward but needs supervision
( ) 2: Can reach forward more than 2 inches safely
( ) 3: Can reach forward more than 5 inches safely
( ) 4: Can reach forward confidently more than 10 inches

9. ***Pick up object from the floor from a standing position**

- Instructions: “Please pick up the shoe/slipper that is placed in front of your feet”
- Grading: Please mark the lowest category

( ) 0: Unable to try/needs assistance to keep from losing balance or falling
( ) 1: Unable to pick up shoe and needs supervision while trying
( ) 2: Unable to pick up shoe but comes within 1-2 inches and maintains balance
(.) 3: Able to pick up shoe but needs supervision
(.) 4: Able to pick up shoe safely and easily

10. ***Turn to look behind over left and right shoulders while standing**

❖ Instructions: “Turn you upper body to look directly over your left shoulder. Now try turning to look over you right shoulder”

❖ Grading: Please mark the lowest category that applies

( ) 0: Needs assistance to keep from falling
( ) 1: Needs supervision when turning
( ) 2: Turns sideways only but maintains balance
( ) 3: Looks behind one side only; other side shows less weight shift
( ) 4: Looks behind from both sides and weight shifts well

11. *Turn 360□

❖ Instructions: “Turn completely in a full circle. Pause, then turn in a full circle in the other direction”

❖ Grading: Please mark the lowest category that applies

( ) 0: Needs assistance while turning
( ) 1: Needs close supervision or verbal cueing
( ) 2: Able to turn 360□ safely but slowly
( ) 3: Able to turn 360□ safely to one side only in less than 4 seconds
( ) 4: Able to turn 360□ in less than 4 seconds to each side

12. *Place alternate foot on bench or stool while standing unsupported

❖ Instructions: “Place each foot alternately on the bench (or stool). Continue until each foot has touched the bench (or stool) four times”. (Recommended use of 6-inch-high-bench.)

❖ Grading: Please mark the lowest category that applies.

( ) 0: Needs assistance to keep from falling/unable to try
( ) 1: Able to complete fewer than two steps; needs minimal assistance
( ) 2: Able to complete four steps without assistance but with supervision
( ) 3: Able to stand independently and complete eight steps in more than 20 seconds
( ) 4: Able to stand independently and safely and complete eight steps in less than 20 seconds

13. *Stand unsupported with one foot in front

❖ Instructions: “Place one foot directly in front of the other. If you feel that you can’t 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” (Demonstrate this test item)

❖ Grading: Please mark the lowest category that applies

( ) 0: Loses balance while stepping or standing
( ) 1: Needs help to step but can hold for 15 seconds
( ) 2: Able to take small step independently and hold for 30 seconds
( ) 3: Able to place one foot ahead of the other independently and hold for 30 seconds
(.) 4: Able to place feet in tandem position independently and hold for 30 seconds

14. *Standing on one leg

❖ Instructions: “Please stand on one leg as long as you can without holding onto anything”

❖ Grading: Please mark the lowest category that applies

( ) 0: Unable to try or needs assistance to prevent fall
( ) 1: Tries to lift leg, unable to hold 3 seconds but remains standing independently
( ) 2: Able to lift leg independently and hold up to 3 seconds
( ) 3: Able to lift leg independently and holds for 5 to 10 seconds
( ) 4: Able to lift leg independently and hold more than 10 seconds

Total Score /56

Appendix.
Functional Gait Assessment[a]

Requirements: A marked 6-m (20-ft) walkway that is marked with a 30.48-cm (12-in) width.

_____1. GAIT LEVEL SURFACE

Instructions: *Walk at your normal speed from here to the next mark (6 m [20 ft]).*

Grading: Mark the highest category that applies.

(3) Normal—Walks 6 m (20 ft) in less than 5.5 seconds, no assistive devices, good speed, no evidence for imbalance, normal gait pattern, deviates no more than 15.24 cm (6 in) outside of the 30.48-cm (12-in) walkway width.

(2) Mild impairment—Walks 6 m (20 ft) in less than 7 seconds but greater than 5.5 seconds, uses assistive device, slower speed, mild gait deviations, or deviates 15.24–25.4 cm (6–10 in) outside of the 30.48-cm (12-in) walkway width.

(1) Moderate impairment—Walks 6 m (20 ft), slow speed, abnormal gait pattern, evidence for imbalance, or deviates 25.4–38.1 cm (10–15 in) outside of the 30.48-cm (12-in) walkway width. Requires more than 7 seconds to ambulate 6 m (20 ft).

(0) Severe impairment—Cannot walk 6 m (20 ft) without assistance, severe gait deviations or imbalance, deviates greater than 38.1 cm (15 in) outside of the 30.48-cm (12-in) walkway width or reaches and touches the wall.

_____2. CHANGE IN GAIT SPEED

Instructions: *Begin walking at your normal pace (for 1.5 m [5 ft]). When I tell you "go," walk as fast as you can (for 1.5 m [5 ft]). When I tell you "slow," walk as slowly as you can (for 1.5 m [5 ft]).*

Grading: Mark the highest category that applies.

(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. Deviates no more than 15.24 cm (6 in) outside of the 30.48-cm (12-in) walkway width.

(2) Mild impairment—Is able to change speed but demonstrates mild gait deviations, deviates 15.24–25.4 cm (6–10 in) outside of the 30.48-cm (12-in) walkway width, or no gait deviations but unable to achieve a significant change in velocity, or uses an assistive device.

(1) Moderate impairment—Makes only minor adjustments to walking speed, or accomplishes a change in speed with significant gait deviations, deviates 25.4–38.1 cm (10–15 in) outside the 30.48-cm (12-in) walkway width, or changes speed but loses balance but is able to recover and continue walking.

(0) Severe impairment—Cannot change speeds, deviates greater than 38.1 cm (15 in) outside 30.48-cm (12-in) walkway width, or loses balance and has to reach for wall or be caught.

_____3. GAIT WITH HORIZONTAL HEAD TURNS

Instructions: *Walk from here to the next mark 6 m (20 ft) away. Begin walking at your normal pace. Keep walking straight; after 3 steps, turn your head to the right and keep walking straight while looking to the right. After 3 more steps, turn your head to the left and keep walking straight while looking left. Continue alternating looking right and left every 3 steps until you have completed 2 repetitions in each direction.*

Grading: Mark the highest category that applies.

(3) Normal—Performs head turns smoothly with no change in gait. Deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width.

(2) Mild impairment—Performs head turns smoothly with slight change in gait velocity (eg, minor disruption to smooth gait path), deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width, or uses an assistive device.

(1) Moderate impairment—Performs head turns with moderate change in gait velocity, slows down, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width but recovers, can continue to walk.

(0) Severe impairment—Performs task with severe disruption of gait (eg, staggers 38.1 cm [15 in] outside 30.48-cm (12-in) walkway width, loses balance, stops, or reaches for wall).

_____4. GAIT WITH VERTICAL HEAD TURNS

Instructions: *Walk from here to the next mark (6 m [20 ft]). Begin walking at your normal pace. Keep walking straight; after 3 steps, tip your head up and keep walking straight while looking up. After 3 more steps, tip your head down, keep walking straight while looking down. Continue alternating looking up and down every 3 steps until you have completed 2 repetitions in each direction.*

Grading: Mark the highest category that applies.

(3) Normal—Performs head turns with no change in gait. Deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width.

(2) Mild impairment—Performs task with slight change in gait velocity (eg, minor disruption to smooth gait path), deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width or uses assistive device.

(1) Moderate impairment—Performs task with moderate change in gait velocity, slows down, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width but recovers, can continue to walk.

(0) Severe impairment—Performs task with severe disruption of gait (eg, staggers 38.1 cm [15 in] outside 30.48-cm (12-in) walkway width, loses balance, stops, reaches for wall).

_____5. GAIT AND PIVOT TURN

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

Grading: Mark the highest category that applies.

(3) Normal—Pivot 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, or pivot turns safely within 3 seconds and stops with mild imbalance, requires small steps to catch balance.

(1) Moderate impairment—Turns slowly, requires verbal cueing, or 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.*

Grading: Mark the highest category that applies.

(3) Normal—Is able to step over 2 stacked shoe boxes taped together (22.86 cm [9 in] total height) without changing gait speed; no evidence of imbalance.

(2) Mild impairment—Is able to step over one shoe box (11.43 cm [4.5 in] total height) without changing gait speed; no evidence of imbalance.

(1) Moderate impairment—Is able to step over one shoe box (11.43 cm [4.5 in] total height) but must slow down and adjust steps to clear box safely. May require verbal cueing.

(0) Severe impairment—Cannot perform without assistance.

(Continued)

Appendix.
Continued

_____7. GAIT WITH NARROW BASE OF SUPPORT

Instructions: *Walk on the floor with arms folded across the chest, feet aligned heel to toe in tandem for a distance of 3.6 m [12 ft]. The number of steps taken in a straight line are counted for a maximum of 10 steps.*
Grading: Mark the highest category that applies.

(3) Normal—Is able to ambulate for 10 steps heel to toe with no staggering.
(2) Mild impairment—Ambulates 7–9 steps.
(1) Moderate impairment—Ambulates 4–7 steps.
(0) Severe impairment—Ambulates less than 4 steps heel to toe or cannot perform without assistance.

_____8. GAIT WITH EYES CLOSED

Instructions: *Walk at your normal speed from here to the next mark (6 m [20 ft]) with your eyes closed.*
Grading: Mark the highest category that applies.

(3) Normal—Walks 6 m (20 ft), no assistive devices, good speed, no evidence of imbalance, normal gait pattern, deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width. Ambulates 6 m (20 ft) in less than 7 seconds.
(2) Mild impairment—Walks 6 m (20 ft), uses assistive device, slower speed, mild gait deviations, deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width. Ambulates 6 m (20 ft) in less than 9 seconds but greater than 7 seconds.
(1) Moderate impairment—Walks 6 m (20 ft), slow speed, abnormal gait pattern, evidence for imbalance, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width. Requires more than 9 seconds to ambulate 6 m (20 ft).
(0) Severe impairment—Cannot walk 6 m (20 ft) without assistance, severe gait deviations or imbalance, deviates greater than 38.1 cm (15 in) outside 30.48-cm (12-in) walkway width or will not attempt task.

_____9. AMBULATING BACKWARDS

Instructions: *Walk backwards until I tell you to stop.*
Grading: Mark the highest category that applies.

(3) Normal—Walks 6 m (20 ft), no assistive devices, good speed, no evidence for imbalance, normal gait pattern, deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width.
(2) Mild impairment—Walks 6 m (20 ft), uses assistive device, slower speed, mild gait deviations, deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width.
(1) Moderate impairment—Walks 6 m (20 ft), slow speed, abnormal gait pattern, evidence for imbalance, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width.
(0) Severe impairment—Cannot walk 6 m (20 ft) without assistance, severe gait deviations or imbalance, deviates greater than 38.1 cm (15 in) outside 30.48-cm (12-in) walkway width or will not attempt task.

_____10. STEPS

Instructions: *Walk up these stairs as you would at home (ie, using the rail if necessary). At the top turn around and walk down.*
Grading: Mark the highest category that applies.

(3) Normal—Alternating feet, no rail.
(2) Mild impairment—Alternating feet, must use rail.
(1) Moderate impairment—Two feet to a stair; must use rail.
(0) Severe impairment—Cannot do safely.

TOTAL SCORE: _____ MAXIMUM SCORE 30

[a] Adapted from Dynamic Gait Index.[1] Modified and reprinted with permission of authors and Lippincott Williams & Wilkins (http://lww.com).

STANDARDIZED MINI-MENTAL STATE EXAMINATION (SMMSE)

	QUESTION	TIME ALLOWED	SCORE
1	a. ***What year is this?***	10 seconds	/1
	b. ***Which season is this?***	10 seconds	/1
	c. ***What month is this?***	10 seconds	/1
	d. ***What is today's date?***	10 seconds	/1
	e. ***What day of the week is this?***	10 seconds	/1
2	a. ***What country are we in?***	10 seconds	/1
	b. ***What province are we in?***	10 seconds	/1
	c. ***What city/town are we in?***	10 seconds	/1
	d. ***IN HOME – What is the street address of this house?*** ***IN FACILITY – What is the name of this building?***	10 seconds	/1
	e. ***IN HOME – What room are we in? IN FACILITY – What floor are we on?***	10 seconds	/1
3	SAY: ***I am going to name three objects. When I am finished, I want you to repeat them. Remember what they are because I am going to ask you to name them again in a few minutes.*** Say the following words slowly at 1-second intervals - bell/ car/ man	20 seconds	/3
4	***Spell the word WORLD. Now spell it backwards.***	30 seconds	/5
5	***Now what were the three objects I asked you to remember?***	10 seconds	/3
6	SHOW wristwatch. ASK: ***What is this called?***	10 seconds	/1
7	SHOW pencil. ASK: ***What is this called?***	10 seconds	/1
8	SAY: ***I would like you to repeat this phrase after me: No ifs, ands or buts.***	10 seconds	/1
9	SAY: ***Read the words on the page and then do what it says.*** Then hand the person the sheet with CLOSE YOUR EYES on it. If the subject reads and does not close their eyes, repeat up to three times. Score only if subject closes eyes	10 seconds	/1
10	HAND the person a pencil and paper. SAY: ***Write any complete sentence on that piece of paper.*** (Note: The sentence must make sense. Ignore spelling errors)	30 seconds	/1
11	PLACE design, eraser and pencil in front of the person. SAY: ***Copy this design please.***  Allow multiple tries. Wait until person is finished and hands it back. Score only for correctly copied diagram with a 4-sided figure between two 5-sided figures.	1 minute	/1
12	ASK the person if he is right or left-handed. Take a piece of paper and hold it up in front of the person. SAY: ***Take this paper in your right/left hand*** (whichever is non-dominant), ***fold the paper in half once with both hands and put the paper down on the floor***. Score 1 point for each instruction executed correctly. Takes paper correctly in hand Folds it in half Puts it on the floor	30 seconds	 /1 /1 /1
	TOTAL TEST SCORE		/30

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