STANCE TIME VARIABILITY AND ENERGY COST OF WALKING IN OLDER ADULTS

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Purpose: To investigate differences in gait characteristics between two walking conditions, wearing and not wearing a portable gas analysis system, and assess the relationship between stance time variability and energy cost of walking in older adults.
Subjects: Forty older adults with preferred walking speeds between 0.8-1.0 m/s were selected.
Methods: Gait characteristics (gait speed, step length, step width, step time, stance time, single-support time, double-support time, step length variability, step width variability, and stance time variability) were recorded (variability derived as standard deviation of all steps) while participants completed eight passes over a computerized walkway, with and without wearing the portable device. Next, concurrent measures of stance time variability and oxygen consumption were collected during four walking conditions (Overground, Rollator, Treadmill, Treadmill Slow), additional measures were recorded as potential confounders (gait speed, biomechanics, fear of falling, confidence in walking, co-morbidities) or to describe sample (age, race, gender).
Analyses: Paired t-tests were used to assess differences between gait characteristics with and without a portable device. ICC's were calculated to describe the agreement between measures. Bivariate analyses were performed to determine association between stance time variability and energy cost of walking during overground walking. Regression was used to assess for independent contributors to energy cost; confounders simultaneously entered first, followed by stance time variability. Additional bivariate analyses were performed for additional conditions. Individual regressions were performed to assess for independent contributors to cost; confounders simultaneously entered first, followed by stance time variability. Mixed-effects models were used to compare stance time variability and energy cost between walking conditions. Post-hoc analysis used to estimate differences between paired conditions of interest.
Results and Clinical Relevance: Our study showed no evidence suggesting wearing a portable device alters overground gait characteristics. Our study also indicates no direct association between stance time variability and energy cost of walking, across any walking conditions. Stance time variability was lower on the treadmill, however, no subsequent changes were observed in energy cost. Continued efforts are needed to investigate multiple contributors to energy cost and assess the unique interactions, modifying, and mediating influences these variables have on energy cost of walking.

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PREFACE

I wish to thank the students, staff, and faculty of the Department of Physical Therapy for providing many forms of support during my doctoral endeavors at the University of Pittsburgh. Your generous support and assistance was much appreciated.

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1.0 INTRODUCTION

1.1 STANCE TIME VARIABILITY AND ENERGY COST OF WALKING

Walking is one of the most convenient means of traveling from one place to another and is the most prevalent physical activity among adults.[1] Walking often plays an integral role in undertaking activities of daily living and in the ability to complete instrumental activities of daily living among older adults.[2] With such a valued role in daily living, it is alarming that research shows that older adults use a greater amount of energy while walking compared to young.[3-6] Higher values of energy cost of walking are related to reports of poorer function in older adults.[7, 8] Similarly, greater energy requirements during walking may lead to greater feelings of fatigue and tiredness, which in turn could contribute to lower levels of physical activity and increased levels of inactivity among older adults – a pathway sure to lead to increased disability and loss of independence.[9]

Oxygen consumption is a vital measure in assessing the energy cost of walking. Technical advancements in *portable systems* for metabolic gas analysis (versus more traditional stationary units) have allowed measures of oxygen consumption to be made in more natural environments such as in the home, outside, or in the clinic – and across a variety of activities and exercises. Such flexibility in measurement has allowed researchers to record oxygen consumption in more natural environments and conditions versus the sometimes unrealistic environment of a research lab. One of the problems facing researchers interested in investigating the potential contribution of specific gait characteristics to the energy cost of walking (using the

portable gas analysis system), is that there is little research showing that wearing a portable device will not result in changes to the characteristics of gait, especially in older adult populations.

Underlying mechanisms (or contributors) of increased energy cost of walking in older adults are not clearly understood, although studies have suggested that an increase in antagonist muscle activity[10-12] and altered biomechanics of gait[13] may play a significant role. Despite work by Griffin et al (2003), which found that the greatest portion of the metabolic cost of walking can be largely explained by the cost of generating muscular force during the stance phase of gait [14], little, if any, work has focused on the possible influence of stance time variability on energy cost of walking. Stance time variability, the fluctuations in stance time from one step to the next, is greater among older adults compared to young, and is associated with greater mobility disability and is an independent predictor for developing future mobility disability in older adults.[15, 16] Being a characteristic of stance time, in addition to being represented differently between young and old[17] and being associated with mobility disability[16], stance time variability appears to be a prime candidate as a possible contributor to energy cost of walking in older adults. Demonstrating that higher levels of stance time variability are correlated with higher values of energy cost of walking during overground walking would add to the explanation of a higher cost of walking in older adults. If researchers could also demonstrate that a change (increase/decrease) in stance time variability induces a subsequent change (increase/decrease) in energy cost of walking, such work would further support the stance time variability – energy cost of walking relationship. Identifying a contributor to the higher cost of walking in older adults would provide a target for rehabilitation programs aimed at improving the efficiency of walking in older adults.

The specific aims of my dissertation research focus on 1) enhancing the knowledge we have on use of portable gas analysis systems by assessing how wearing such a device may alter the gait characteristics of older adults, and 2) to assess the potential relationship between stance time variability and energy cost of walking in older adults.

1.2 SPECIFIC AIMS AND HYPOTHESES

1.2.1 Specific Aim I

1.2.2 To determine the impact of an ambulatory metabolic measurement system (MedGraphics VO2000®) on the gait characteristics of older adults.

1.2.2.1 Hypothesis I

It is hypothesized that gait characteristics (gait speed, step length, step width, step time, stance time, single-support time, double-support time, step length variability, step width variability, and stance time variability) will be similar between two conditions – walking with and without wearing a portable gas analysis system.

1.2.3 Specific Aim II

To assess the relationship between temporal gait variability (i.e. stance time variability) and energy cost of overground walking in older adults

1.2.3.1 Hypothesis II

We expect stance time variability will be positively associated with energy cost of walking overground. Older adults with higher levels of stance time variability will have greater energy cost compared to older adults with lower levels of variability.

1.2.4 Specific Aim III

To further assess the relationship between stance time variability and energy cost of walking, across conditions expected to vary (increase/decrease) stance time variability.

1.2.4.1 Hypothesis III

We expect stance time variability to remain positively associated with energy cost of walking across all walking conditions (Overground, Rollator, Treadmill, and Treadmill Slow). We also expect that a change (increase/decrease) in stance time variability between paired walking conditions will result in a subsequent change (increase/decrease) in energy cost of walking. If a walking condition reduces stance time variability, we would expect to see a subsequent reduction in the energy cost of walking for the same walking condition.

1.3 BACKGROUND

The first and second sections of this background I will review what is known about oxygen consumption (energy cost of walking) and stance time variability, respectively, as related to developing our study. In the third section I will propose a conceptual model of how the two variables are related, and how this model was used to develop our study conditions investigating the relationship between stance time variability and energy cost of walking in older adults.

1.3.1 Oxygen Consumption and Energy Cost of Walking

Oxygen is the key to accessing the "energy" available in the foods we eat (specifically, carbohydrates, fats, and proteins), as it plays a vital role in the oxidation of these substrates – which ultimately (through various metabolic processes) results in the formation of adenosine triphosphate (ATP).[18] The energy needed for vital body functions, as well as for muscle contractions for movement, is found in the bond of the phosphate anion (~P) in ATP. When ~P is split from ATP, energy is released and available for "use" by the body. Thus, oxygen consumption has a precise relationship with energy production; measuring the amount (rate) of oxygen used during an activity reveals the amount of energy that was used (or that was required) to perform that activity.[18]

Oxygen consumption (or energy expenditure) is simply a measure of the amount of oxygen (energy) that is required for a person of a given height and weight to perform a specific activity at a set pace. There are two basic methods for determining human energy expenditure during rest and physical activity, with direct or indirect calorimetry. Direct calorimetry takes advantage of the fact that all of the body's metabolic processes result in the production of heat

and thus measures an individual's heat and work production in thermally insulated calorimeters as a direct measure of energy.[1, 5] Direct calorimetry is complex and requires considerable time and expense making it impractical for measurement of most activities in laboratory studies.[1, 5]

Indirect calorimetry relies on the measure of oxygen consumption to determine the energy cost of rest or activity. Previous research has demonstrated that the amount of oxygen consumed at rest or during work can be expressed in "heat equivalents" and is equal to the heat produced by the body as determined directly in a calorimeter[19]. Therefore, measuring an individual's oxygen consumption during rest or physical activity provides an indirect estimate of energy expenditure.[18] Oxygen consumption measured by indirect calorimetry can be performed by closed-circuit spirometry or open-circuit spirometry. Open-circuit spirometry remains the most widely used procedure to measure physical activity oxygen consumption as it allows the individual to inhale ambient air versus re-breathing only the oxygen from a large, bulky spirometer as is done during closed-circuit spirometry.[18] Ambient air has a constant composition of 20.93% oxygen, 0.03% carbon dioxide, and 79.04% nitrogen. Differences in oxygen and carbon dioxide percentages in expired air, compared to inspired ambient air, yield an indirect measure of energy metabolism.[1] By analyzing the volume of air breathed during a given time period and the composition of the exhaled air, a useful measure of oxygen consumption can be translated into a measure of energy cost of a given activity.[1, 19] Oxygen consumption during activity has been shown to directly increase with body mass, especially in weight bearing exercise like walking and running.[1] Thus, to eliminate the variation in oxygen consumption due to body mass, oxygen consumption can be expressed as ml/kg min. This reduces differences between individual's, regardless of age, race, gender, and body mass.[1] Finally, to determine the *energy cost* associated with walking at a given speed, the amount of oxygen consumed (ml/kg

min) is divided by the gait speed (m/min) resulting in the measure of energy cost of walking (ml/kg m).[5]

Measurement of oxygen consumption has traditionally occurred in laboratory settings under controlled conditions, such as using a treadmill or cycle ergometer, with sophisticated, and often stationary, equipment.[1, 3, 4, 6, 20] However, the recent introduction of portable gas analysis devices have expanded the possible mode of data collection to activities of daily living (bathing, dressing, feeding,)[2], routine chores (shopping, cleaning, stair climbing)[2], and physical activity (walking)[2, 21, 22]. The portability of the devices allow researchers to collect information regarding oxygen consumption (energy cost) across a variety of "real life" activities, and within everyday environments – versus the novel confines of a research laboratory.

Although these portable devices are smaller, usually placed within a harness worn over the participants' shoulders, they have been shown to provide equally as accurate measures of oxygen consumption as the larger, "stationary" units. Schrack and colleagues (2010) compared the accuracy of the Cosmed K4b[2] portable device against a traditional, stationary system (Medgraphics D-Series) during submaximal walking in a group of young adults, and found that the portable system provided similar measures of oxygen consumption and carbondioxide expiration to the traditional, stationary system (ICC's ranged from 0.93 to 0.97).[21] Similar work by other researchers have reported similar findings, that various brands of portable gas analysis systems provide accurate and similar measures of oxygen consumption as do stationary systems.[23-25] Bales et al (2001) expanded the work on portable gas analysis systems by investigating the influence of wearing the device on measures of aerobic capacity and heart rate; specifically they assessed whether the added weight and potential restrictions of the harness influenced the primary measure of oxygen consumption.[26] Their results indicated no significant

difference in estimated maximal oxygen consumption occurred at low to moderate workloads (submaximal work), while workloads approaching maximal performance did show a difference in oxygen consumption – greater values were observed while wearing the device versus not wearing the device.[26] So, some caution may be warranted in interpreting oxygen consumption data collected with a portable device during high intensity and longer duration activities (approaching maximal participant workloads), as the data may over-estimate the energy cost of the activity being assessed.

While evidence suggests that the portable gas analysis devices provide accurate and similar values as the more traditional, stationary units, and that wearing a portable device does not influence oxygen consumption measures collected during submaximal workloads[26, 27], to our knowledge – few studies have assessed whether wearing a portable gas analysis device alters the gait and biomechanical characteristics. In aging research, it is widely known that the energy cost of walking is greater for older adults than young[3-6]; this fact has spurred great interest among many researchers to study the gait of older adults in an attempt to identify the underlying mechanisms responsible for the difference in energy cost. It is during these instances, when overground walking gait characteristics and energy cost of walking are concurrently being assessed, that knowledge of consequences of wearing a portable gas analysis system on gait characteristics would be vital. Without it, there is uncertainty whether results are based on true findings within participants or are otherwise a result of wearing a portable device.

1.3.1.1 Appropriate Representation of Energy Cost

The intensity of the activity being measured will influence how the energy cost of that activity is calculated. Whether the activity being performed is anaerobic or aerobic will determine how many measures of oxygen consumption will be needed to accurately calculate energy cost. If the activity is anaerobic (involving both the anaerobic and aerobic systems) oxygen consumption must be measured at rest, during activity, and during activity recovery.[19] However, if the activity is performed aerobically, only the oxygen consumption at rest and during the steady state period of activity need to be measured.[19] Resting oxygen consumption must be deducted from the oxygen measured during activity in order to determine the oxygen consumption of the activity alone (net oxygen consumption of activity)[19], although many studies report the gross oxygen consumption of activities which includes the resting oxygen consumption value.

In order for the aerobic measure of oxygen consumption to be accurate and a true representation of the amount of oxygen being used for a given activity, oxygen consumption must be recorded at a period of steady state for the activity. Steady state means that the rate of oxygen consumption reaches a level sufficient to meet the energy demands of the activity (and more-so the body tissues)[5], often seen as a plateau in the values of oxygen consumption, breathing rate and respiratory exchange ratio.[5] Measuring the rate of oxygen consumption during this period of "steady state" reflects the true amount of oxygen (energy) required of the activity.[5] The equation for determining the energy cost (EC) of walking at a comfortable walking pace would is:

$$EC_{net} = \frac{\text{Walking O2 consumption at steady state (ml/kg min)} - \text{Resting O2 consumption (ml/kg min)}}{\text{Gait Speed (m/min)}}$$

1.3.1.2 Factors Associated with Energy Cost of Walking

Previous research has identified individual factors that influence energy cost during activity. A relationship between walking speed and energy cost of walking has been identified. The speed-energy cost relationship has been represented by a J-shaped curve, yielding higher energy cost measures for speeds slower and faster than optimal walking speed of approximately 1.3 m/s.[4, 28, 29] This J-shaped relationship holds true for older adults, however, the curve is shifted upwards, meaning older adults expend greater energy during walking than young, even at similar gait speeds (Figure 1 –adapted from Martin, 1992).[4, 28] The mechanisms involved in the vertical shift are unclear, but gait characteristics and biomechanical differences during walking in older adults may play a role, in addition to a potentially higher cost of generating muscle force for older adults.[4]

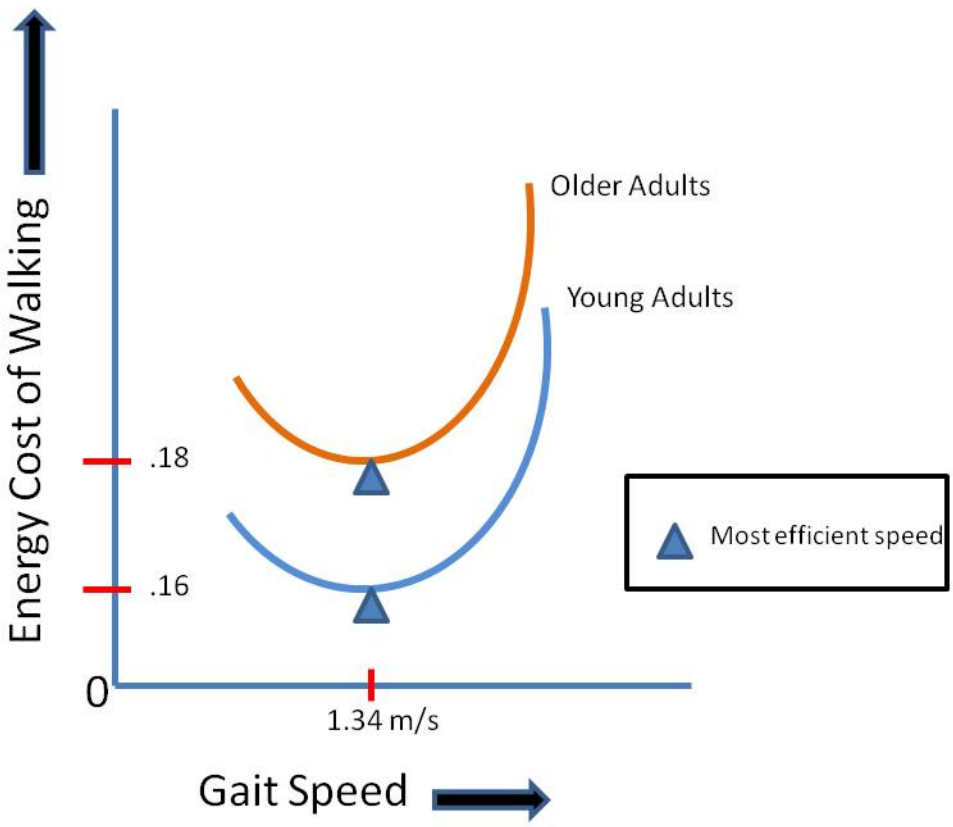


Figure 1: Relationship between Energy Cost of Walking and Gait Speed[4]

There are a number of studies that have investigated the potential influence of gait biomechanics and postures on energy cost of walking.[13, 14, 20, 30-33] Waters et al[34] demonstrated that the energy cost of walking is greater for individuals with hip or ankle fusions and knee motion restrictions than for people without such restrictions. Increases in trunk flexion and subsequent compensations made at the hip and ankle have also been reported to yield greater values of oxygen consumption.[33] Wert and colleagues (2010) assessed the impact of a number of gait characteristics and biomechanical abnormalities in a sample of older adults and found that individually, greater degrees of hip abnormality (reduced hip extension), increased cadence, and step width were significant independent contributors to the energy cost of walking. However,

adding cadence and step width to a regression model controlling for age, gait speed, and hip extension, failed to provide any additional factor explaining the variance in the energy cost of walking. Thus, hip abnormality was shown to be the strongest biomechanical contributor to energy cost in their sample of older adults. The authors also showed that greater degrees of hip abnormality (greater reductions in hip extension) lead to greater values of energy cost of walking.[13] A more flexed posture has also been implicated as possible source for increased energy cost; Saha and colleagues (2007) found that increased trunk flexion required greater oxygen consumption than more erect postures.[33]

Gait characteristics have also been investigated as possible sources for greater cost of walking among older adults. Malatesta et al (2003) tested whether the lower economy (higher energy cost) of walking in older adults is due to greater gait instability (specifically defined as stride time variability).[3] However, their study of healthy older adults showed no significant correlation between gait instability (stride time variability) and energy cost of walking.

Alterations in muscle function have also received a great deal of attention as a possible mechanism contributing to energy cost of walking. Specifically, it has been suggested that older adults have increased co-activation of antagonist muscles during walking than young adults, and that the increased muscle demand may account for increased oxygen consumption. Hortobagyi and colleagues (2009) suggested that an increase in agonist muscle activation and antagonist muscle coactivation (elicited by neural factors) during the gait cycle may account for an increase in the cost of walking seen in older adults compared to young.[11] In a later study, the same group reported the cost of walking in older adults was, indeed, associated with the magnitude of agonist and antagonist muscle coactivation.[10] Agonist muscle activation accounted for 31% of cost of walking, while antagonist muscle coactivation accounted for 43% of the cost. Coactivation rates

in general were higher for older adults compared to young.[10] Hortobagyi's findings were consistent with those of Peterson et al (2010), who reported higher thigh coactivation, versus the shank, among older adults compared to young – especially just before and after heelstrike.[12] Peterson and colleagues also reported an association between total coactivation (shank and thigh) and cost of walking, r=.55 (walking speed 1.12 m/s).

Walking surfaces can also influence the energy cost associated with walking.[1] Similar energy cost exists for level walking on a grass track or paved surface, whereas increased energy expenditure is reported for walking in snow and sand.[1] Data is less certain in reporting the energy cost associated with walking on a treadmill compared to overground at the same gait speed. Early studies by Ralston (1960)[29] reported no differences in energy cost of treadmill walking and overground walking at similar speeds, whereas, more recent work by Parvataneni et al. (2008) reported significant differences in energy cost between the two conditions.[35]

Psychosocial factors associated with walking (fear, confidence, anxiety) may also play a role in energy cost. Emotions such as fear and anxiety can influence heart rate.[1] This increase in heart rate, not due to increase in activity level, can in turn influence oxygen consumption measures in individuals experiencing these emotions. Such could be the case with individuals not familiar with walking on a treadmill. The influence of fear and anxiety on oxygen consumption and gait characteristics, attributed to treadmill walking, can be addressed by allowing familiarization and acclimatization with the condition.[36-38]

Studies working with young adults have shown that as little as 6 – 10 minutes of familiarization and acclimatization to walking or running on a treadmill is needed to reduce influential characteristics.[36-38] Martin, Rothstein and Larish (1992) allowed 30 minutes of acclimatization to treadmill walking for their study involving older adults and reported a

systematically lower (3%) oxygen consumption on the 2nd day retest measures.[4] Un-acclimated older adults were shown to have a 1.3 ml/kg min lower oxygen consumption rate on retest measures of treadmill walking that was associated with learning or habituation effect.[39]

1.3.1.3 Significance of Energy Cost of Walking

Energy cost of walking can provide researchers and clinicians with baseline information regarding the efficiency with which participants are walking compared to age-related norms, and can also serve as an outcome measure used to assess rehabilitation programs aimed at improving the mobility of older adults. Higher costs of walking may place an overall greater demand on body systems (reducing energy reserves available to complete other daily tasks) or may discourage participation in physical activity among older adults – who are known to have higher energy cost of walking than young. This reduced physical activity and increased inactivity may lead to greater risk of mobility disability and functional decline.[9, 40]

Wert et al. (2009,2010) have reported that a higher energy cost of walking is related to poorer self-report of function in older adults with mobility disability and that conversely, reducing the energy cost of walking can lead to improved reports of physical function.[7, 8] Additionally, VanSwearingen (2009) reported a subsequent increase in walking confidence with a decrease in energy cost of walking following a twelve week exercise program.[41] So, although it appears that there are consequences to having higher values of energy cost of walking, research suggests that cost of walking can be improved (reduced) with subsequent benefits to function and confidence. Thus, continued efforts should be made to identify contributors to gait inefficiency and establish rehabilitation programs which address these contributors, returning gait to more efficient and less energy demanding levels.

1.3.1.4 Gaps in Knowledge of Energy Cost of Walking

The literature strongly supports the notion that older adults expend a greater amount of energy during walking compared to young adults, even at similar speeds.[3-6] Researchers continue to explore the possible contribution of a variety factors to the greater energy cost of walking in older adults. Regardless of age, gait research has shown that the general metabolic cost of walking is largely explained by the cost of generating muscular force during the stance phase of gait.[14] Considering that older adults have a higher cost of walking and that greatest portion of the cost in walking comes from generating muscle forces during stance – investigating the stance phase of gait in older adults, and how it may differ from that of young adults, may provide information regarding the greater energy cost observed in older gait.

One stance time gait characteristic known to vary more among older adults than young, is stance time variability – the fluctuation in stance time from one step to the next throughout the gait cycle. Higher values of stance time variability have been associated with poorer mobility among older adults.[15] Despite this knowledge of stance time variability, no studies, to our knowledge, have investigated whether the higher energy cost of walking observed in older adults is related to stance time variability.

1.3.2 Stance Time Variability

1.3.2.1 Definition, Prevalence and Etiology.

The gait cycle is divided into two main periods or phases, stance and swing. Stance time is the entire period during which the foot is on the ground and is subdivided into three intervals, initial

double stance, single limb support, and terminal double stance.[42] Stance time comprises 60% of the gait cycle[42] and has been implicated as the period of gait most associated with the energy cost of walking.[14]

Stance time *variability* (STV) is the fluctuation in stance time from one step to the next during gait and is frequently reported as the within-subject standard deviation of all steps during a given walking trial or as the coefficient of variation calculated as within-subject standard deviation/within-subject mean. It has been suggested that among community-dwelling older adults (free from neurological disease), 35-45% may have variable gait (significant fluctuations in gait characteristics from one step to the next) during usual overground walking.[17] The percentage is even greater among those older adults who are institutionalized or have neurological disorders[43, 44].

1.3.2.2 Toward a Better understanding of Origin of Stance Time Variability.

Little is known about the specific influence on the neural control mechanism responsible for irregular gait, but it has been suggested that complex modifications in the brain, spinal cord and peripheral nerves may contribute to the decline in motor consistency seen in aging.[45, 46] Some researchers have suggested that age-related "reorganization" of cortical and subcortical structure communication in response to neurodegeneration[45, 46], may alter the input to spinal central pattern generators responsible for automated, smooth stepping.[47] Central pattern generator activation can also be driven and influenced by peripheral/biomechanical input; in humans, the sensory stimulus for regulated stepping is the signal derived from sensory nerves of the anterior thigh – signaling the transition from stance to swing.[48] This signal is "activated" by hip

extension during gait, thus, biomechanical changes related to reduced hip extension that may occur with aging, may reduce or blunt the peripheral cue for stepping. So it may be the combined altered signals arising from central and peripheral regions, which converge on spinal central pattern generators that give rise to inconsistent signals for motor output (ie. muscle function during gait). These inconsistent signals to the spinal cord would alter the otherwise consistent stepping mechanism, resulting in a more inconsistent and less smooth manner of walking.

Early work by Gabell and Nayak (1984) hypothesized that certain gait characteristics, such as step length and stride time, were representative of the automatic stepping mechanism of gait (central pattern generator for gait) and that step width and double-support time are representative of balance control, and a failure or disruption to either the automatic stepping mechanism or balance control, would lead to an increase in the variability of the respective characteristics of gait.[49] The collective works by Moe-Nilson and Helbostad (2005), Helbostad (2007) and Aaslund and Moe-Nilssen (2008), together, have further supported the notion that different measures of variability indicate different or opposing aspects of motor behavior[50-52], and thus the different variability measures may in fact represent different constructs. Recently, Moe-Nilssen and colleagues (2010) investigated whether gait variability measures represent different constructs in a sample of older adults. The results they reported were interesting as step length variability and step time variability were not correlated, suggesting that spatial and temporal variability may represent different constructs. When looking at the correlation between footfall variability measures and trunk variability measures, the investigators found that step length variability showed a strong association to anterior-posterior interstep trunk acceleration variability (r=-0.76) while step time variability was strongly associated with vertical interstep

trunk acceleration variability. Medial-lateral interstep trunk acceleration variability did not correlate significantly with any of the other measures. Such findings support the previous researchers[50-52] notion that different variability measures may represent different constructs and thus may provide different insight into the nature of the variability as being either adaptive or sign of impairment in the control mechanism.[53]

Brach et al. (2008) completed a similar study but included stance time variability as a temporal variable. They examined the contribution of CNS and sensory impairments to the variability of spatial and temporal gait characteristics in a sample of older adults without overt disease.[54] Brach and colleagues hypothesized that central nervous system impairment (ie. alterations in executive function and central processing) would affect motor control and thus the gait characteristics stance time and step length variability, whereas sensory impairments (reduced vision and lower extremity vibration sense) would affect balance mechanisms and thus, step width variability. Indeed, stance time variability (more so than step length variability) was associated with measures of cognitive function and central processing, factors related to central nervous system impairment. Step width variability was only associated with sensory impairment measures. The fact that stance time variability was not associated with sensory impairments and step width variability was not associated with central nervous system impairments is an equally as strong finding, indicating that different contributions exist for various types of gait variability based on the specific gait characteristic.

1.3.2.3 Factors Associated with Stance Time Variability

Stance time variability has been associated with gait biomechanics and gait speed, among older adults. Brach et al. (2008) took a closer look at the cross-sectional and longitudinal associations between gait biomechanics and stance time variability and found that at baseline, individuals with obvious hip extension compared to those with just barely visible hip extension (determined by the modified Gait Abnormality Rating Scale – GARSM) were less variable in stance time.[55] Furthermore, individuals who had minimal or moderate improvement in biomechanics after a 12 week exercise program had decreases in stance time variability (0.016 and 0.010s, both $p<0.002$, respectively) compared to those who demonstrated no change or worse biomechanics, (0.003s, $p=.39$).[55]

As with other forms of gait variability, stance time variability has also been shown to be affected by the speed that one is walking (gait speed). Brach et al. (2006) investigated the effect of gait speed on step length, step width, and stance time variability in a sample of young and older adults and found that older adults compared to young had increase in stance time and step length variability during slow speeds compared to usual speeds. Alternately, stance time variability decreased in both young and older adults with faster walking speeds compared to usual speed.[17] Given the influence of gait speed on stance time variability in older adults, the researchers suggested that older adults may have increased difficulty with motor control when challenged (asked to walk slowly compared to usual speed), and for some older adults – walking slowly may be more challenging than walking fast.

Considering that many older adults tend to walk more slowly as they age and that slower gait speed and small increases in stance time variability have been shown to be strong predictors

of future mobility disability, the relationship between the two gait characteristics should be of significant interest to all healthcare workers involved in improving the mobility and function of older adults.

1.3.2.4 The Significance of Stance Time Variability.

Having greater values of stance time variability is associated with greater limitations in mobility. Brach et al. (2007) have reported that a difference at baseline in stance time variability of 0.01s was associated with a 13% higher incidence of mobility disability (HR 1.13, 95% CI, 1.01-1.27) over a 54 month follow up period.[16] In a more recent longitudinal study of 241 older adults, Brach et al. (2010) reported that among older adults who reported no change in mobility (walking ability) over the course of one year, measures of stance time variability were stable. In contrast, those individuals who reported a decline in mobility, stance time variability increased.[56] These two studies have expanded the knowledge on stance time variability by demonstrating that both baseline levels and change over time (increases) in stance time variability are important indicators of impaired mobility in older adults.

As clinicians and researchers, we are also interested in knowing "how much" change in a given outcome measure is clinically meaningful. Such values would also prove useful in evaluating the effectiveness of certain interventions aimed at improving gait variability as well as helping to determine appropriate sample size and power computations when planning future studies investigating gait variability. Based on distribution-based and anchor-based approaches, Brach et al. (2010) recently provided such estimates of meaningful change in stance time variability; based on effect size estimates for change, a small change was considered to be 0.005 (s), while a difference of 0.014 (s) was reported as a moderate change.[56] These results were

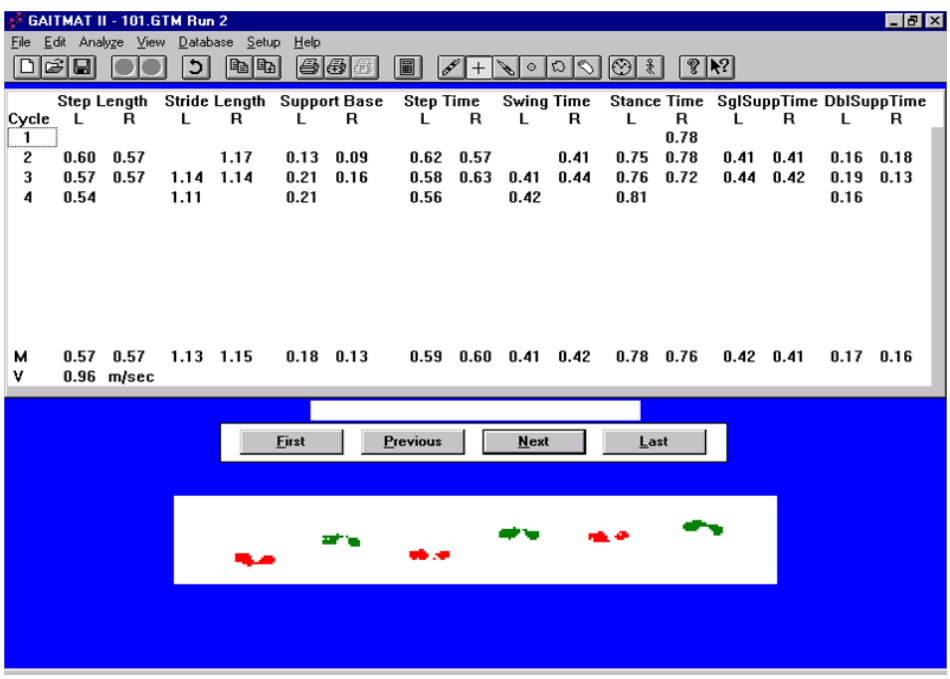
consistent with the groups previous findings where they reported a 0.01 s increase in stance time variability was associated with a 13% increase in risk of mobility disability.[16]

1.3.2.5 Measurement Methods of Gait Variability

Stance time variability, along with other general gait characteristics, can be measured by a variety of methods; instrumented computerized walkways, footswitch systems, triaxial accelerometry, and observational assessments have all been used to measure and assess mean gait characteristics and variability.

Instrumented, computerized walkways are either rigid or flexible mats with embedded pressure sensitive switches that open and close in response to pressure (Figure 2). These specialized walkways can vary in length but often cover a minimum distance of 4 meters, with inactive sections on either side to allow for acceleration and deceleration. Such systems often implement software that automatically detects each footfall and analyzes the complete gait sequence and records both spatial (ie. step length and width) and temporal (ie. swing and stance time) parameters of gait (Figure 3).

Figure 2: Instrumented computerized walkway



	Step Length		Stride Length		Support Base		Step Time		Swing Time		Stance Time		SglSuppTime		DblSuppTime	
Cycle	L	R	L	R	L	R	L	R	L	R	L	R	L	R	L	R
1												0.78				
2	0.60	0.57		1.17	0.13	0.09	0.62	0.57		0.41	0.75	0.78	0.41	0.41	0.16	0.18
3	0.57	0.57	1.14	1.14	0.21	0.16	0.58	0.63	0.41	0.44	0.76	0.72	0.44	0.42	0.19	0.13
4	0.54		1.11		0.21		0.56		0.42		0.81				0.16	
M	0.57	0.57	1.13	1.15	0.18	0.13	0.59	0.60	0.41	0.42	0.78	0.76	0.42	0.41	0.17	0.16
V	0.96	m/sec														

Figure 3: Real-time display during data collection (GaitMat II)

Brach et al. (2008) have reported on the reliability and validity of measures of gait using an instrumented walkway; test-retest reliability of gait speed and mean gait characteristics in their sample of older adults using a computerized walkway was excellent (ICC ≥ .80) while measures of gait variability were fair (stance time variability, ICC = 0.63).[57] Reliability of gait variability was greatest for two 4-meter passes over the walkway compared to a single 4-meter pass; investigators suggested that further work be done to determine whether consistency can be further improved by increasing the number of steps analyzed.

Footswitch systems allow for collection of temporal gait characteristics without the restraint of repeated walking passes performed only on the instrumented walkway. Such portable systems allow for continuous walking over longer distances within usual clinic or laboratory environments (Figure 4). Footswitch systems operate by using pressure sensitive switches which can be placed inside or on the bottom of a shoe; common placement is at the base of the first metatarsal head (medially) and at the point of heel contact (posterior-laterally) on each foot. (Figure 9) Switches can vary regarding size as well as the force required to depress or activate the switch. Signals created by the depression and reflection of the footswitches are relayed to a computer via wireless technology and various software programs are used to analyze the raw data. The values for temporal gait characteristics are determined by analyzing heel contacts and toe-offs of each foot, for each step. Gait variability (for the gait characteristic of interest) is determined either by calculating the standard deviation of all steps recorded or by calculating the coefficient of variation (CV) for a particular characteristic. Footswitch systems have been used in previous research [58-61]to record the temporal gait characteristics within various populations, and have been shown to be a valid and reliable measure for recording temporal parameters of gait.[58, 60]

Figure 4: Footswitch System

1.3.2.6 Amenability of Stance Time Variability

A number of studies have demonstrated that stance time variability can be reduced while walking on a treadmill compared to walking at a similar speed overground, in both healthy older adults and older adults with neurological impairments.[62-65] Studies by Herman et al. (2007) and VanSwearingen et al. (2009) have also shown that gait training on the treadmill and other skilled interventions aimed at restoring normal stepping can reduce gait variability in older adults.[41, 66] Thus, stance time variability has been shown to be immediately amenable during treadmill walking as well as modifiable following a period of specific movement intervention and

treadmill training, making it a viable gait characteristic to address during physical therapy gait analysis in older adults.

1.3.3 Conceptual Model for Stance Time Variability / Energy Cost Relationship: The Foundation for our Study

As previously discussed, scientists haven't fully identified the neural etiology behind irregular gait (such as high stance time variability), although some researchers have hypothesized "altered" information arising from cortical and subcortical regions of the brain, as well as from the periphery (biomechanics), may be a key factor.[11, 47, 48] This altered information, which converges on spinal central pattern generators (CPG's), may result in inconsistent signals in motor output (ie. muscle function during gait). The disruption of the automatic (consistent) stepping mechanism from spinal CPG's, likely results in a more inconsistent and less smooth manner of walking (variable gait) by way of altered duration of muscle firing, changes in firing pattern of muscles, and overall greater muscle recruitment during gait.[47] We suggest the increase in energy demand requested from the muscles requires an increase in oxygen consumption in order to produce the demand-specific production of energy (ATP). Since the greatest consumption of energy during walking is attributed to the work of muscles involved during the stance phase of gait[14], it is feasible that the increased demand placed on the muscles of gait contribute to the greater energy cost of walking observed in older adults (Figure 5). Although our primary interest lies in exploring the possible contribution from stance time variability, we also recognize that other factors may also contribute to energy cost of walking.

Increased coactivation of muscle antagonists and agonists has been implicated as a possible contributor to the increased energy cost of walking observed in older adults.[10-12, 47] Increased muscle activity, thought to arise from a general decrease in the body's overall inhibition capabilities, is thought to increase the demand in oxygen consumption (thus energy) required during walking.

Personal factors such as fear of falling, anxiety, and confidence in walking may also play a role in the cost of walking among older adults. Researchers have shown that older adults who are fearful of falling demonstrate greater muscular coactivation compared to their counterparts who are not fearful.[67] Furthermore, coactivation has also been shown to be higher among older adults with poor postural control compared to older adults with good postural control.[68] Despite the increased muscular activity observed in conjunction with fear of falling in some older adults, little if any research has looked at concurrent measures of energy cost of walking in this same subsample of older adults; thus the influence on cost of walking remains unknown. However, researchers have suggested that the increased muscular activity associated with coactivation further increases the oxygen consumption demand – and eventually yields a higher cost of walking.[47]

Finally, environmental factors, such as type of footwear and quality of walking surface (flat, uneven, dry, wet, etc…), may also contribute to the energy cost of walking.[1, 29] Walking on paved or grass surfaces yield similar energy costs, whereas walking in snow and sand increase the energy demand of walking.[1] The weight of the shoe worn during walking can also add to the cost of walking; adding 100 g to each shoe can increase oxygen consumption by up to 1% during moderate running.[69] Other researchers have reported that the cushioning of the shoe can also affect cost of walking/running. Softer cushioning versus more rigid (firm) cushioning can

reduce the cost of running at a moderate speed by 2.4%, even if the softer soled shoes weighs slightly more.[70] Despite what we know about shoe weight and cushioning, less is know how these factors affect the cost of walking in older adults, but we would expect similar findings.

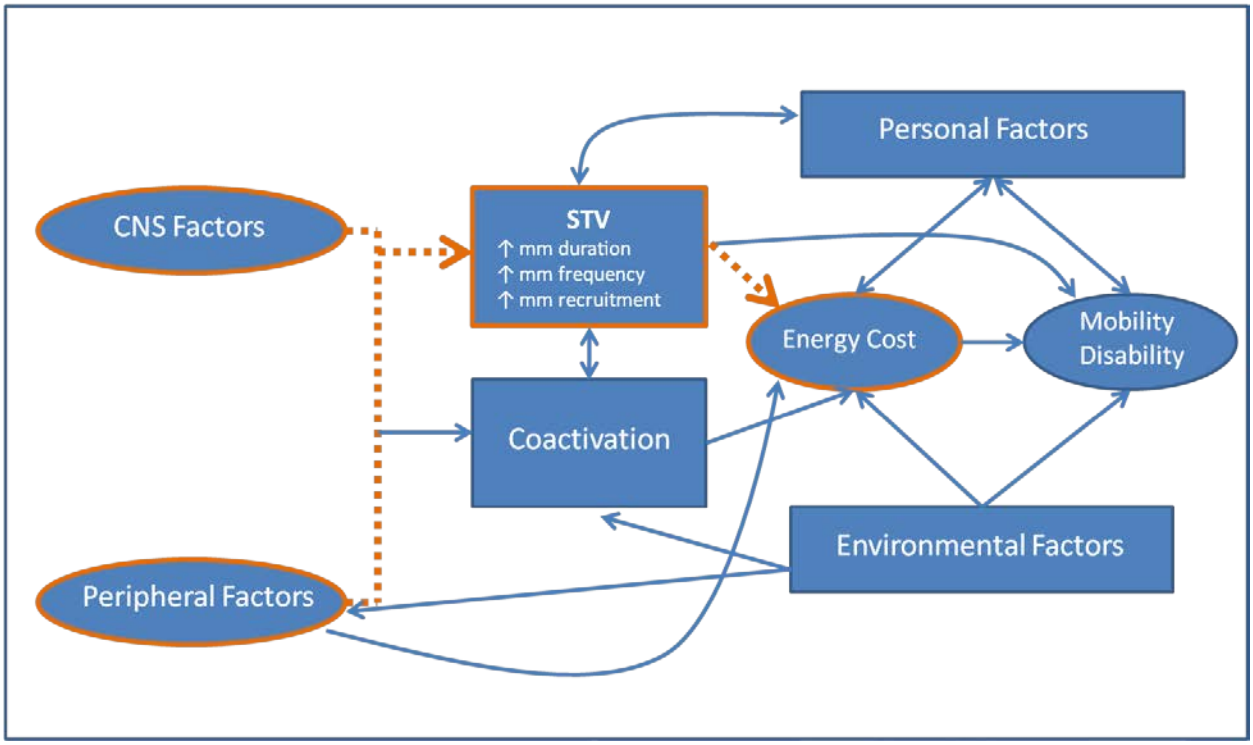


Figure 5: Conceptual Model

The walking conditions selected for our study are based on the idea, if central nervous system factors and/or biomechanical factors are "manipulated", stance time variability would be affected (increased or decreased). And, if a relationship existed between stance time variability and energy cost of walking – a subsequent increase or decrease would be observed in energy cost of walking.

The treadmill was selected as a way to reduce stance time variability. We propose treadmill walking reduces stance time variability via "enhanced" peripheral sensory input to

spinal central pattern generators neurons. The sensory stimulus for regulating stepping in humans, arises from sensory nerves of the anterior thigh[48], and is activated with hip extension as the foot moves behind the body.[48] Hip extension during gait has been shown to be less in older adults compared to young [71, 72], thus, the peripheral signal for stepping may be "dampened" with aging. We propose during treadmill walking, the stance limb is pulled behind the body to a greater extent than what occurs overground – improving the peripheral signal for stepping. The treadmill may also act as a consistent external "timing" cue to the spinal stepping mechanism by *consistently* providing a peripheral cue for stepping at the hip at regular intervals (as speed is kept constant). Preliminary work in our lab supported the expectation of reduced stance time variability while walking on the treadmill.[65]

Since slow walking has been shown to increase variability measures of gait[17], we selected slow walking on the treadmill as method of increasing stance time variability. We opted to have participants walk slowly on the treadmill, versus overground, as we wanted to ensure walking remained slow consistently throughout the duration of the walking trial. We performed preliminary work before beginning our study, which showed that a .4 mile per hour decrease in gait speed (from preferred walking speed) was required to elicit a meaningful decrease (.01 s) [16] in stance time variability in a small sample of older adults.

Using a rollator (four-wheeled walker) has been shown to improve the walking efficiency in adults with stable COPD[73], although the mechanisms by which this occurs are less elucidated. Alkjaer et al (2006) reported the angular impulse of the hip extensors was significantly increased during rollator walking, suggesting that joint biomechanics and subsequent actions of muscles are different when walking with a rollator.[74] As such, we selected the rollator as another potential method of reducing stance time variability, overground, versus the treadmill. We

propose that using a rollator may improve hip extension, similar to the treadmill, and thus providing an enhanced peripheral signal for stepping.

In summary, we proposed a model of the relationship between stance time variability and energy cost of walking that implicates an increase in demand from muscle during stance phase of gait (due to altered central and peripheral input to spinal central pattern generator neurons) as contributing to the energy cost of walking in older adults. We selected walking conditions thought to act on either the central or peripheral mechanisms for stepping, as a way to assess our suspicions of a potential relationship between stance time variability and energy cost of walking – by way of altering stance time variability and assessing the subsequent impact on energy cost of walking. The following chapters will present the specific results of our studies undertaken to assess the proposed model and influence stance time variability on energy cost of walking.

2.0 CHAPTER II – THE INFLUENCE OF WEARING A PORTABLE GAS ANAYLSYS SYSTEM ON THE GAIT CHARACTERISTICS OF OLDER ADULTS

2.1 INTRODUCTION

Critical thresholds in aerobic capacity (20 ml/kg min^{-1}) have been established for independent living among older adults; falling below this threshold has been associated with an 8-fold decline in physical function.[75] Since routine walking and completion of daily tasks can consume a substantial portion of older adults' total capacity, any substantial loss of capacity or additional energy demand could threaten their independence.[9, 40] Indeed, daily tasks can require up to 30-50% of some older adults' total aerobic capacity[2], while some lower functioning older adults (< 20 ml/kg min^{-1} total aerobic capacity) may use up to 87% of their capacity just during walking. When basic tasks such as walking exact such high energy cost, older adults are left with very little "reserve" for performing the rest of their vital tasks such as bathing, cooking, and shopping; that limitation, in turn, may lead to reduced physical activity and greater risk for disability. [2, 9, 11, 40] Wert et al. (2009,2010) have reported that a higher than normal energy cost of walking is related to poorer self-report of function in older adults with mobility disability, but their research also suggests that these patients' predicament is not insoluble: reducing the energy cost of walking can lead to improved reports of physical function.[7, 8]

Oxygen consumption is a crucial measure used to assess aerobic capacity and derive the energy cost of walking. Historically, cumbersome, stationary measurement devices have confined assessment of oxygen consumption to laboratory settings, where participants either

walk on a treadmill or pedal on a cycle ergometer. However, such laboratory-defined modes of activity do not match the true activity of older adults, which may influence the generalizability of study results and limit the relevance of the findings when compared to more realistic activities.

However, technical advancements in *portable systems* for metabolic gas analysis permit researchers to measure oxygen consumption in more natural environments (home, outside, clinic), while participants engage in clinically relevant activities and exercises. These less artificial measures provide greater insight into the cost of performing a variety of daily, routine tasks. Numerous studies have reported on the reliability and validity of portable measuring devices compared to the standard fixed models.[21, 23-25, 76-78] Although the reliability of the many portable devices available on the market can vary from brand to brand[21, 23-25, 76-78], portable gas analysis systems have been shown to be a reliable and valid for measuring oxygen consumption.[21, 23-25, 76-78]

But, we have some concern that wearing such portable devices may alter gait characteristics, especially among older adults. Although portable devices have previously been used to examine physiological and metabolic responses during walking and activity, only a few studies actually report on the potential changes to cardiovascular and physical performance measures caused by wearing a portable gas analysis system.[26, 27] Bales et al. (2001) examined the effect of wearing a portable gas analysis system on aerobic capacity and heart rate during a high-intensity step test in young adults and found that wearing the device had no statistical significant effect upon estimated maximal oxygen consumption or in heart rate. Similarly, Gault and colleagues (2009) found that wearing a portable gas analysis system (compared to not wearing the device) had no impact on l-mile walking performance (walking time, speed, heart rate and predicted VO2max) in older adults.[27] Especially in older populations, the 3-9 pound

weight of the portable system and the distraction of wearing a face mask may change waking performance and characteristics. Knowledge of such interference would help researchers, devoted to investigating concurrent measures of gait and energy cost of walking in older adults, ensure that gait data is not being altered as a result of wearing a portable device. If gait is shown to be affected, data may also prove useful in establishing correction factors to account for the influence of wearing such equipment.

The purpose of this study is to investigate potential differences in gait characteristics of older adults with slow gait during two usual-paced walking conditions, wearing and not wearing a portable gas analysis system (VO2000™, MedGraphics Corp.). Specifically, we are interested in assessing *mean* spatial (step length, step width) and temporal (step time, stance time, single-support time and double-support time) characteristics as well as three measures of gait *variability* (step length variability, step width variability, and stance time variability). We hypothesize that wearing a portable gas-analysis device will result in no significant differences in gait characteristics when compared to measures recorded while not wearing the device.

2.2 METHODS

2.2.1 Study Design

A cross-sectional study design was used to assess gait characteristics measured during two different walking conditions (with and without wearing a portable gas analysis system).

2.2.2 Participants

Individuals were eligible to participate in this study if they: were 65 years of age or older, reported the ability to tolerate a 5 hour session (with rest periods) of answering questions and performing movement and walking tests, were able to ambulate a minimum of household distances (approximately 50 feet) without the use of an assistive device and without the assistance of another person, and had a usual overground walking speed in the range of 0.8-1.0 m/s.

Exclusion criteria included the inability to provide informed consent, concomitant neuromuscular disorders that impair movement, diagnosis of cancer with active treatment within the past 6 months, severe pulmonary disease, non-elective hospitalization for a life-threatening illness or major procedure within the past 6 months, chest pain with activity or a cardiac event within the past 6 months. Forty participants were recruited from previous studies of mobility and balance in older adults that took place within the Pittsburgh Claude D. Pepper Older Adults Independence Center (Pittsburgh OAIC). Participants verbally expressed an interest in participating in future studies involving older adults and agreed to have their name and contact information shared with other studies currently undergoing recruitment of participants. All participants who agreed to participate in this study signed a consent form approved by the Institutional Review Board of the University of Pittsburgh. Demographic characteristics of the participants are reported in Table 1.

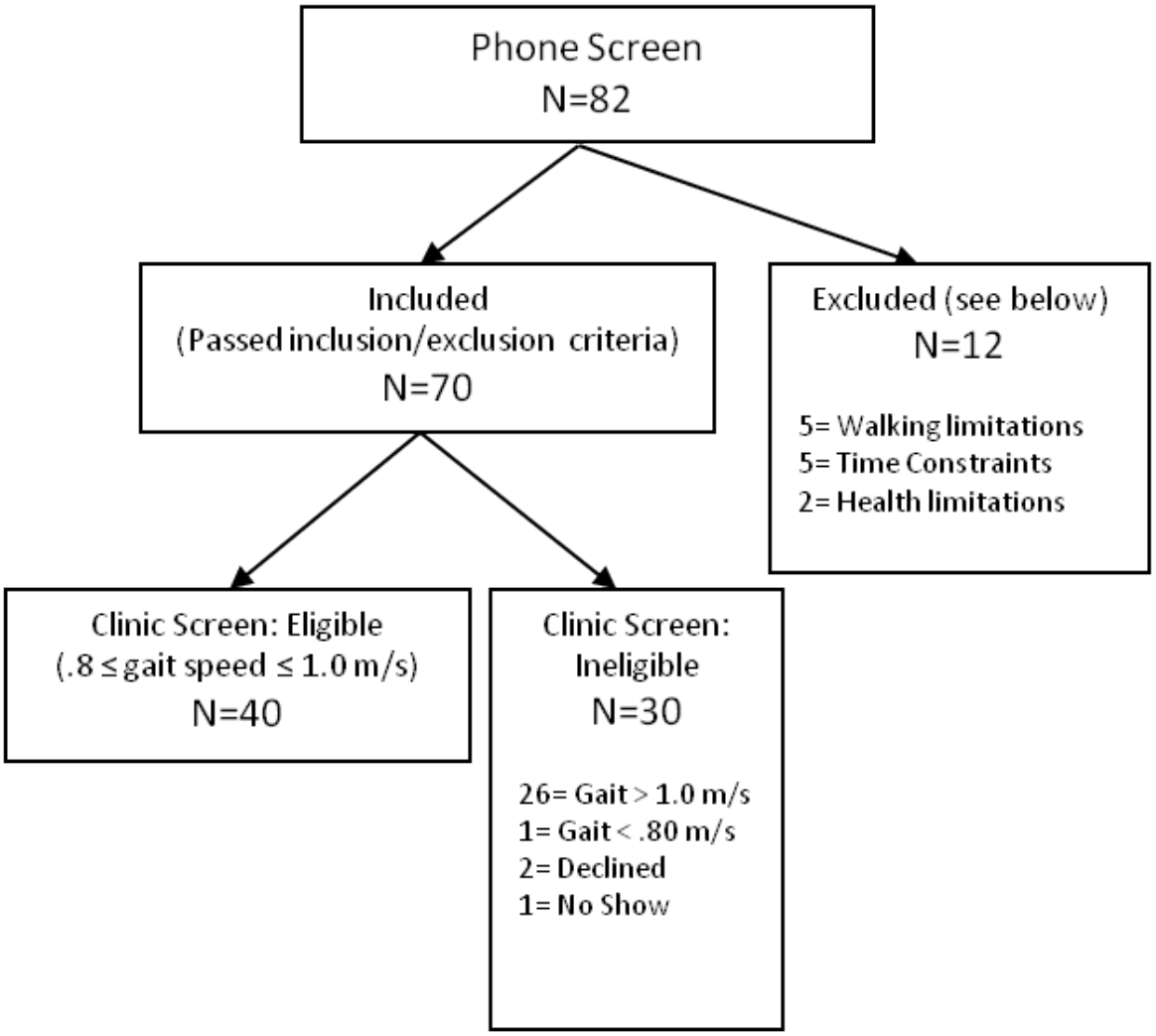


Figure 6: Study Flow Chart

Table 1: Demographic Characteristics of Study Sample

Variables	Mean (standard deviation) or Percentage (n)
Age, years	76.9 (6.8)
% Female	72.5 (29/40)
% Caucasian	90 (36/40)
Co-morbidities (0-8)	2.9 (1.4)

2.2.3 Procedures

Participants completed a brief walking assessment (2 passes over a 4 meter computerized walkway at a self-selected usual walking pace) to determine eligibility based on the gait speed inclusion criteria (gait speed between 0.8-1.0 m/s). Once eligibility was confirmed, gait characteristic data were collected during two walking conditions, *with* and *without* wearing a portable gas-analysis system. The "*with device"* condition required participants to wear a lightweight portable gas analysis system and face mask (VO2000™, MedGraphics Corp.) attached to the front of a nylon shoulder harness worn by the participant (Figure 7). The second condition was performed in the same manner but without the device, and thus referred to as the "*without device*" condition. The order of the walking conditions was randomly assigned for each participant.

Figure 7: Portable gas analysis system

2.2.4 Measures

2.2.4.1 Gait Characteristics

Gait characteristics (gait speed, step length, step length variability, step width, step width variability, step time, stance time, stance time variability, single-support time, and double-support time) were recorded while participants walked at their self-selected, usual walking speed over an 8 meter long computerized walkway (GaitMat II). The initial and last two meters of the walkway are inactive and allow for acceleration and deceleration, while the middle four meters are used for data collection. The computerized walkway, an automated gait analysis system, is based on the closing of pressure-sensitive switches in response to footfalls on the walkway. The program software automatically calculates both spatial and temporal measures of gait. The

GaitMat has shown good reliability in measuring gait characteristics, test-retest values range from ICC= .89 -.99 for mean gait characteristics and from ICC=.40-.63 for gait variability.[57] Measures of gait variability recorded on the GaitMat demonstrated concurrent validity against measures of general health, functional status, and physical activity level.[57]

After two practice passes on the computerized walkway, participants completed 8 consecutive passes over the walkway at their usual, self-selected pace for each walking condition. A rest period was provided between the two walking conditions as needed.

Mean values for each gait variable were calculated by incorporating all steps within the 8 passes for each condition; variability measures were reported as the standard deviation of all steps within each condition. Gait speed was recorded in meters/second, while spatial characteristics (step length, step width, step length variability, and step width variability) and temporal characteristic (stance time, single-support, double-support time, and stance time variability) were measured in meters and seconds, respectively.

Because previous research has shown spatiotemporal gait data to be asymmetric [79], we provided values for left, right, and pooled (left-right) steps for variables appropriate for individual step interpretation.

2.2.4.2 Questionnaires

Each participant completed a demographic questionnaire and an assessment of co-morbidities to provide a description of our sample of older adults. The presence of co-morbidities was ascertained with the Co-Morbidity Index[80], which includes 18 different diseases, categorized to 8 domains. The total number of positive domains (0-8) was recorded.

2.3 DATA ANALYSIS

Descriptive statistics, mean and standard deviation, were calculated to summarize the data. Paired t-tests were used to assess significance of mean differences between the two walking conditions for each variable. The Intra-Class Correlation (ICC) (1,k) coefficient and its 95% confidence interval (CI) was used to assess the consistency and absolute agreement between the two walking trials. ICC's were interpreted as follows: less than 0.40, poor; 0.4 to 0.75, fair to good; and more than 0.75, excellent.[81] PASW Statistics 18 (SPSS) software was used for all statistical analyses.

2.4 RESULTS

2.4.1 *Mean* Spatial and Temporal Gait Characteristics

Mean values and standard deviations, along with mean differences and associated p-values, for both walking conditions are reported in Table 1. Measures of gait speed, step length, step time, stance time, single-support time, and double-support time showed no significant difference between the two walking conditions ($p > .05$). Only the measure of step width differed significantly between the two conditions (difference = -.003, p=.006). The negative value of the difference suggests that step width was greatest when wearing the portable gas-analysis system (Table 2).

Differences between left and right footsteps for all measures collected without the portable system were also insignificant ($p>.05$). However, when wearing the portable device,

differences of .005 s (p<.05) and .007 s (p<.001) were observed between left/right stance time and left/right single-support time, respectively.

ICC values for each of the mean gait characteristics are reported in Table 2. All seven gait characteristics had excellent agreement between walking conditions; the highest values were found for single-support time, step width, step length and stance time (0.95-0.98), whereas gait speed was slightly lower (0.94).

Table 2: Mean Gait Characteristics, Differences, and Agreement between Walking Conditions

Variables	*Without* Equipment Mean ± SD	*With* Equipment Mean ± SD	Difference Mean ± SD	t	p	ICC (95% CI)
Gait Speed (m/s)	.935 ± .114	.929 ± .122	.006 ± .040	.942	.353	.94 (.89-.97)
Step length (m): L	.544 ± .061	.542 ± .064	.002 ± .018	.608	.547	.96 (.92-.98)
Step length: R	.548 ± .066	.542 ± .068	.006 ± .018	1.903	.066	.96 (.92-.98)
Difference (L - R)	-.004 ± .027	-.000 ± .029	-	-	-	-
Step length: LR	.546 ± .062	.542 ± .064	.004 ± .017	1.360	.183	.96 (.93-.98)
Step width (m): LR	.038 ± .030	.041 ± .032	-.003 ± .007	-2.961	.006**	.97 (.95-.99)
Step time (s): L	.588 ± .049	.591 ± .049	-.003 ± .017	-1.214	.233	.94 (.89-.97)
Step time: R	.587 ± .053	.585 ± .054	.002 ± .014	.658	.515	.97 (.94-.98)
Difference (L - R)	.001 ± .020	.006 ± .018	-	-	-	-
Step time: LR	.587 ± .050	.588 ± .051	-.001 ± .014	-.327	.745	.96 (.93-.98)
Stance time (s): L	.749 ± .071	.752 ± .073	-.003 ± .022	-.671	.506	.96 (.91-.98)
Stance time: R	.753 ± .071	.758 ± .072	-.005 ± .022	-1.266	.214	.95 (.90-.97)
Difference (L - R)	-.004 ± .015	-.005 ± .014*	-	-	-	-
Stance time: LR	.751 ± .071	.755 ± .072	-.004 ± .022	-.965	.342	.95 (.91-.98)
Single-support time (s): L	.423 ± .049	.421 ± .048	.002 ± .009	1.400	.171	.98 (.97-.99)
Single-support time: R	.427 ± .046	.428 ± .045	.001 ± .013	-.054	.957	.96 (.92-.98)
Difference (L - R)	-.004 ± .016	-.007 ± .015**	-	-	-	-
Single-support time: LR	.425 ± .047	.424 ± .046	.001 ± .009	.673	.506	.98 (.96-.99)
Double-support time (s): LR	.162 ± .033	.165 ± .032	-.003 ± .010	-1.732	.092	.95 (.91-.98)

**p<.01; *p<.05; L = left foot; R = right foot; LR = pooled left and right feet

2.4.2 Gait *Variability* Measures

Similar to the *mean* characteristics of gait reported above, the three gait *variability* measures (step length variability, step width variability, and stance time variability) were similar between the two walking conditions (Table 3). Likewise, differences between left and right footsteps within each walking condition were minimal and failed to reach significance (p>.05).

ICC values were lower for all three measures of gait variability compared to the values reported for mean gait characteristics (Table 3). The lowest ICC values were reported for individual left and right step length variability, 0.32 and 0.39 respectively. However, the pooled left-right measure for step length variability had a greater ICC value of 0.73. Step-width variability (ICC = 0.79) and LR stance time variability (ICC = 0.56) also showed fair-good agreement.

Table 3: Variability Gait Characteristics, Differences, and Agreement between Walking Conditions

Variables	*Without* Equipment Mean ± SD	*With* Equipment Mean ± SD	Difference Mean ± SD	t	p	ICC (95% CI)
Step length variability (m): L	.027 ± .007	.025 ± .007	.002 ± .008	1.285	.207	.32 (-.002-.59)
Step length variability: R[α]	.026 ± .007	.027 ± .008	-.001 ± .008	-	.719	.39 (-.07-.64)
Difference (L - R)	.001 ± .007	-.002 ± .008	-	-	-	-
Step length variability: LR[α]	.030 ± .007	.029 ± .008	.0002 ± .006	-	.935	.73 (.53-.85)
Step width variability (m): LR	.031 ± .008	.030 ± .008	.001 ± .005	1.024	.313	.79 (.63-.89)
Stance time variability (s): L	.030 ± .006	.031 ± .009	.001 ± .007	.637	.528	.56 (.28-.75)
Stance time variability: R	.030 ± .006	.032 ± .011	-.002 ± .009	-1.227	.228	.51 (.22-.72)
Difference (L - R)	.001 ± .005	-.001 ± .006	-	-	-	-
Stance time variability: LR	.031 ± .005	.032 ± .009	-.001 ± .007	-.441	.662	.56 (.28-.75)

CI = confidence interval; L = left foot; R = right foot; LR = pooled left and right feet

2.5 DISCUSSION

This study assessed the potential impact of wearing a portable gas-analysis device on the characteristics of walking in a sample of community-dwelling older adults with slow gait; we found no evidence to suggest that wearing a portable system for assessing oxygen consumption alters the gait characteristics of walking in older adults.

Despite wearing a portable gas-analysis system for one of the walking conditions, older adults in our study have similar values for 95% (21/22) of the observed gait characteristics across both walking conditions, the lone difference was observed in step width. Although a difference in step width was found to be statistically significant (-.003 m, p=.006) between the two walking conditions, this extremely small discrepancy is likely to be considered clinically insignificant - especially in view of the standard error of the measure [SEM = .01; calculated as (sd√1-ICC)]. Other studies, which have assessed similar measures of step width in older adults, have also reported similar SEM, ranging from a low of .012 to a high of .05.[16, 56, 57] Furthermore, if we calculate a small effect size (.2)[82] for step width in our sample, the difference in step width between walking conditions would have to exceed .01 m [Effect size = (sd).2], which again, our reported difference is well below. The calculated effect size for our study is consistent with those from other studies (.08, .02, .02) with measures of step width in older adults. [16, 56, 57]

Not only are the mean spatial and temporal gait characteristics from our sample of older adults alike across both walking conditions but they are also consistent with mean gait characteristics previously reported in the gait literature regarding a similar sample of older adults.[56] Comparing our mean values of gait to those of Brach et al. (2010), we find that gait speed (.93 m/s to .93 m/s), stance time (.76 s to .76 s), step length (.54 m to .55 m), step length variability (.029 m to .029 m), and step width variability (.030 m to .037 m) are almost identical.

Of the six gait variables that were similarly collected for both studies, only stance time variability differed between the two studies of older adults; older adults in our study were slightly less variable in stance time (0.031 s) than the older participants in the Brach study (0.038 s).

Our values of reliability (ICC) are also consistent with other studies using similar gait analysis equipment. Comparing our gait reliability values to similar measures recorded by Brach et al. (2008), gait speed (ICC = 0.94 vs 0.98), step length (ICC = 0.96 vs. 0.99), and stance time (ICC = 0.95 vs. 0.98) show the greatest similarities.[57] The reliability of step width in our study (ICC=.97) was higher than that reported by Brach et al. (ICC=.89), however, both values show excellent reliability in the measure. Paterson et al (2008) also examined the reliability of spatiotemporal gait parameters in a sample of older adults (although all women) and reported excellent ICC's, step length (.95), step time (.87), and stance time (.91).[83] Once again, step width was slightly lower compared to our study, ICC=.68, although still within the fair-good range.[81]

The reliability values for gait variability reported in our study are slightly higher than those observed by Brach et al (step length variability ICC = .73 versus .63; step width variability ICC = .79 versus .40, and stance time variability ICC = .56 versus .50), which may be attributed to the increased number of passes we used during our walking conditions. It has been suggested that a greater number of steps used in the calculation of gait variability measures may yield greater consistency among measures of variability[57, 84], which is likely the case with our findings as our participants completed approximately 40-48 steps compared to approximately 10-12 steps performed in Brach's study. Interestingly, our test-retest reliability of gait variability measures on the GaitMat without equipment was; step length variability (ICC=.84), step width variability

(ICC=.77) and stance time variability (ICC=.55), again – similar to what we reported above for "between" walking conditions.

Overall, studies investigating the reliability of gait characteristics have shown that the reliability of gait variability is less than that of mean measures of gait [57, 83]; one possible reason for the lower reliability may be inherent in the measure itself. If a participant is inconsistently variable, we would expect the reliability of the measure to be low – as there is inconsistency between measures. However, if a participant is consistently variable or consistently non-variable, then we would expect the reliability of the measure to be higher (more consistent). A mix of both types of participants may yield a reliability somewhere in the middle. Future research is needed to better understand the variable nature of gait variability and how such variability impacts measures of gait variability.

Although not a primary aim of this study, we were also able to show that left and right footstep measures were similar across most gait characteristics, suggesting individual (unilateral) footsteps could be used for analyses of gait in like populations of older adults. Although differences between left and right footsteps were reported for stance time and single-support time, both differences were below their respective SEM from our study, .02 and .01 respectively, as well as from other studies [16, 56, 57]). Also, both differences were less than the calculated moderate difference based on an effect size of .2. We do recognize, however, that left-right differences may be apparent in participants with unilateral lower extremity limitations, where such impairments could affect usual, smooth, and coordinated walking patterns (severe osteoarthritis to hip, knee, or ankle; recent hip/knee arthroplasty, neurological gait impairments). Though it appears that individual foot data can be used for analysis of gait, it should be noted that the reliability of pooled left-right gait characteristics, in our case [step length variability

(ICC=.73) and stance time variability (ICC=.56)], may be better than reliability values for individual foot measures [step length variability L (ICC=.32), R (ICC=.39); stance time variability L (ICC=.56), R (ICC=.51)].

This study has important strengths. We collected data on a number of spatial and temporal gait characteristics, including measures of gait variability, for left, right, and pooled left-right footsteps. This allowed us to perform a comprehensive assessment of the impact wearing a portable device may have on a wide range of gait characteristics. Furthermore, we collected data over eight passes of 4-meter walking for each condition, increasing the total number of footfalls to be used for analyses, which in turn provided more reliable measures of gait variability in our sample of older adults.

We also recognize a few limitations to our study. We assessed the impact of wearing one type of portable gas-analysis system, the VO2000 (MedGraphics Inc.). We understand that there are differences among portable devices and therefore the findings should only be generalized to like systems. Additionally, we had a relatively small sample of older adults (n=40) which may not represent all community-dwelling older adults. Our sample also consisted of older adults with slow gait; therefore, generalizations should be made to like groups.

As researchers and clinicians strive to collect oxygen consumption and energy cost information on older adults during natural activities and conditions that exist outside the laboratory, a portable gas-analysis system is becoming necessary. For researchers, like those devoted to investigating the contribution of changes in gait characteristics to the energy cost of walking in older adults, it is essential (especially when concurrent collection of measures is undertaken) that we understand the potential consequences that such devices may have on the gait patterns within this population . Wearing a portable gas-analysis system, as described in our

study, does not alter specific spatial and temporal patterns of walking in our sample of older adults with slow gait. However, we recommend further research in other preferably larger samples and across different portable devices to add to the strength of the literature supporting the use of this exciting alternative to restrictive laboratory data collection.

3.0 CHAPTER III – THE RELATIONSHIP BETWEEN STANCE TIME VARIABILITY AND ENERGY COST OF WALKING OVERGROUND

3.1 INTRODUCTION

Older adults expend more energy during walking than young adults, even at similar preferred walking speeds. [5, 6] This poses some concern - as higher cost of walking has been shown to be related to poorer self-report of function in older adults[7], and may in some instances account for up to 87% of an older adults maximum aerobic capacity.[41] Older adults who expend greater amounts of energy just to walk are left with very little "reserve" to complete additional daily tasks and activities. [2, 9, 11, 40, 41] Despite the potential negative consequences of increase energy cost of walking for older adults, improvements (reductions) in the efficiency of walking have been related to subsequent improvements in reports of physical function. In a study of older adults with mobility disability, Wert and colleagues (2010) [8]found that energy cost of walking explained 29% of the change in self-report of physical function.[8]

The body's greatest consumer of energy (oxygen) is muscle[1, 18]; muscles perform a variety of energy-demanding tasks during walking by operating as motors, tensile struts, and brakes.[85-88] Griffin and colleagues (2003) investigated the role of leg muscle function in the determination of energy cost of walking, and found that active muscle volume required to generate force on the ground and the rate of generating this force accounted for >85% of the increase in metabolic rate during the stance phase of gait.[14] As such, we might suspect that the higher energy cost of walking observed in older adults may be the result of an age-related change (increased) in muscle demand during the stance phase of gait. One gait characteristic known to:

1) occur during stance phase, 2) have higher (more abnormal) values among older adults compared to young, and 3) is suspected to result from age-related alterations in motor output (muscle function), is *stance time variability* - the fluctuation in stance time from one step to the next.

So, specifically, how could stance time variability contribute to increased oxygen (energy) demand from muscles? Although little is known about the neural control of irregular walking, some researchers have attributed the age-related progressive loss of motor abilities (exhibited in gait characteristics – like stance time variability) to "changes" in the input or signaling to spinal mechanisms responsible for automatic stepping. [47, 89, 90] Similarly, changes or disruptions to the peripheral signal for stepping, arising from sensory nerves of the anterior thigh[48], have also been implicated in altering the automatic stepping pattern via age-related changes in gait biomechanics. The combined "altered" contribution from central nervous system and peripheral input onto spinal mechanisms is thought to result in altered duration and frequency of muscle activity during gait, potentially creating an increase in demand (energy) from muscles during walking.[47] Thus, we propose the underlying mechanism which relates stance time variability to energy cost of walking is the increased oxygen demand placed on the muscles due to altered output from spinal centers responsible for automatic stepping.

Like energy cost of walking, stance time variability has been related to poorer function (mobility) in older adults. In a sample of community-dwelling older adults, Brach et al. (2007) reported a difference at baseline, in stance time variability, of 0.01s was associated with a 13% higher incidence of mobility disability (HR 1.13, 95% CI, 1.01-1.27).[16] Furthermore, the same group completed a longitudinal study assessing the onset of mobility disability (measured as a difficulty in walking) as related to stance time variability[56], and found that among older adults

who reported no change in mobility (walking ability) over the course of one year, measures of stance time variability were stable. Individuals reporting a decline in their walking ability, however, had an increase in their measure of stance time variability.[56]

Despite the evidence above, which appears to suggest that stance time variability could be a viable factor contributing to energy cost of walking, no research, to our knowledge has directly assessed whether a relationship exists between stance time variability and energy cost of walking in older adults. Therefore, the purpose of this study is to assess the relationship between stance time variability and energy cost of overground walking in older adults. Our hypothesis is stance time variability will be positively associated with energy cost of walking, such that older adults with higher levels of stance time variability, compared to those with lower levels, will have a greater energy cost of walking.

3.2 METHODS

3.2.1 Study Design

A cross-sectional design was used to assess the relationship between stance time variability and energy cost of walking in a sample of older adults with slow gait.

3.2.2 Participants

Individuals were eligible to participate in this study if they: were 65 years of age or older, reported being able to walk 4-6 minutes non-stop at their preferred walking pace, without the use

of an assistive device and without the assistance of another person, and had a usual overground walking speed in the range of 0.8-1.0 m/s.

Participants were excluded from participating in our study if they: were unable to provide informed consent, had concomitant neuromuscular disorders that impair movement, had a diagnosis of cancer with active treatment or severe pulmonary/cardiac disease, or were recently hospitalized for a life-threatening illness or major procedure. Forty participants were recruited from previous studies of mobility and balance in older adults that took place within the Pittsburgh Claude D. Pepper Older Adults Independence Center (Pittsburgh OAIC). Of the 40 older adults eligible for the study, 30 had complete data and were used for subsequent analyses. The 10 individuals not included in the analyses were missing either footswitch or oxygen consumption data due to equipment malfunction. Baseline characteristics were similar between the older adults with complete data and those with incomplete data (Table 4). All participants who agreed to participate in this study signed a consent form approved by the Institutional Review Board of the University of Pittsburgh.

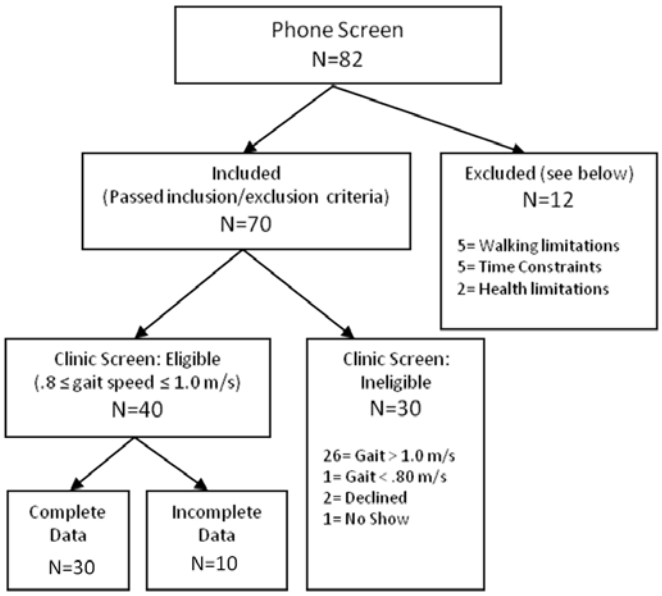


Figure 8: Study Flow Chart

3.2.3 Procedures

Participants completed a brief walking assessment (2 passes over a 4 meter computerized walkway at a self-selected usual walking pace) to determine eligibility based on the gait speed inclusion criteria (gait speed between 0.8-1.0 m/s). Once eligibility was confirmed, participants had height, weight, blood pressure and heart rate assessed, and completed a series of questionnaires regarding their confidence in walking and self-reports of fear of falling.

Next, oxygen consumption and gait characteristics were recorded during a single trial of self-selected, usual-paced walking (lasting between 4-6 minutes) around a 150 foot indoor track. Additionally, participants were video-taped during a segment of the walking trial which was later used for an observational assessment of gait abnormalities and mechanics.

3.2.4 Measures

3.2.4.1 Gait Characteristics

Temporal gait characteristics were recorded during 4-6 minutes of overground walking at a self-selected, usual walking pace using a portable footswitch system. Footswitches (1 inch x 1 inch membrane switches with a stainless steel metal dome that depresses with 20 ounces of force) were placed on the soles of the participants' shoes – at the first metatarsal head and laterally at site of heel strike (Figure 9).

Figure 9: Footswitch placement

Data were sampled at 100 Hz with 16 byte precision, using a Class 1 Bluetooth protocol to transmit data to a desk-top computer system. Temporal measures of gait (stance time) were determined using a custom written LabView program. Using MATLAB, heel-contact and toe-off instances were identified (Figure 10) using thresholding methods (minimum set at 2.5 Hz). Heel-contacts and toe-offs were then used to calculate stance time while the standard deviation of all steps was used as the measure of stance time variability.[56] Temporal gait characteristics were data synced, using computer time, with oxygen consumption to allow for concurrent recording of both measures. Thus, stance time variability was determined during the same time period as oxygen consumption (and ultimately – cost of walking) at steady state. The reliability of our footswitch system to measure stance time variability during overground walking was consistent with values of stance time variability reported from the GaitMat (ICC = .51 vs. ICC = .50, respectively), the gold standard in gait measurement.

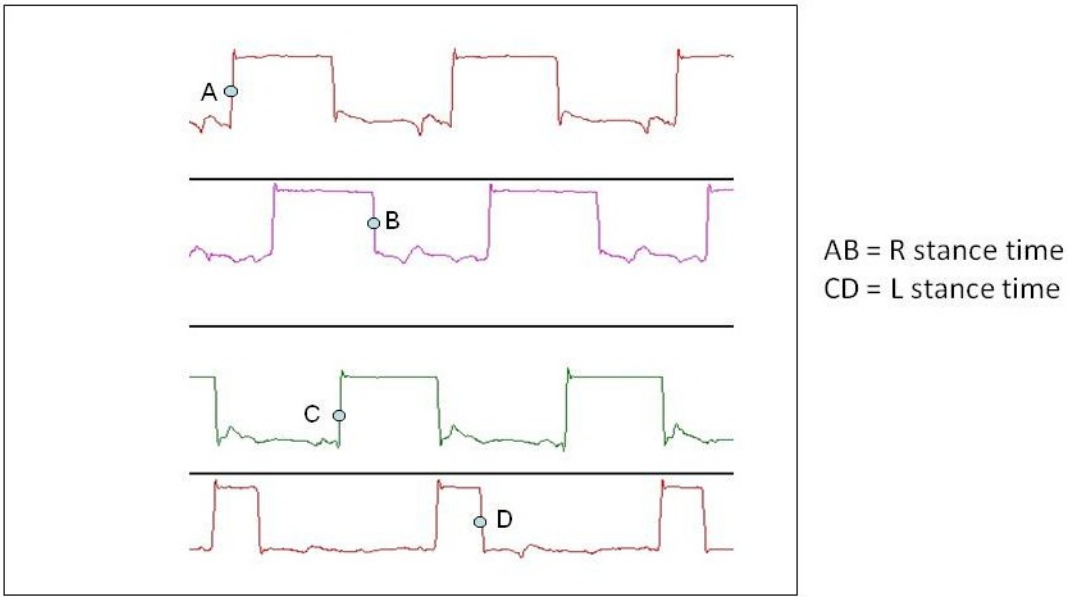


Figure 10: Footswitch signal for heelstrike and toe off

3.2.4.2 Overground Gait Speed

Two 4-meter distances were marked at opposite ends of a 150 foot indoor track. Time to complete each 4 meter distance was recorded for ~ 75% of the laps completed during the overground walking condition. On occasions where 4-meter times were not collected, researchers were monitoring collection of footswitch and oxygen consumption data. Gait speed (meters/second) was calculated by dividing the distance (4 meters) by the recorded time (in seconds) for each 4 meter pass; the mean gait speed for the condition was calculated from all gait speeds recorded during the condition. Individual gait speeds were consistent within the walking condition; in a sub-sample of our study (n=20), we calculated energy cost of walking (for each participant) using the (within subject) mean of all gait speeds collected during the overground condition, and compared it to the energy cost of walking calculated using a single measure of gait speed collected towards the end of the walking condition. There was no difference between the two measures of gait speed (mean difference = .005 m/s, t_{paired}= -.54, p=.59) or cost (mean difference = -.001 ml/kg m^{-1}, t_{paired}= -.49, p=.63), suggesting the gait speeds were similar. The reliability of overground gait speed for our study was, ICC=.84.

3.2.4.3 Oxygen Consumption

Oxygen consumption was determined by indirect calorimetry and analysis of expired gases.[22, 91] A portable metabolic measurement system (VO2000, MedGraphics Corp.™, Figure 7) was used to measure oxygen consumption (ml/kg min) while the participant walked around a 150 foot indoor track at a self-selected, preferred walking pace. Participants wore a light-weight, portable

gas analysis system (VO2000, MedGraphics Corp. ™) attached to a nylon harness worn over the shoulders, and a neoprene mask which fit comfortably over the nose and mouth (total weight, 4 lbs).

Mean rate of oxygen consumption and carbon dioxide production was determined over one-three minutes after physiologic steady state was reached (approximately 2-3 minutes after the initiation of walking).[1, 41, 92] Parameters for the determination of steady state were defined as a combination of the following: the plateau of oxygen consumption (values not to fluctuate more than 3 ml/kg min.) and a stable RER value (not to exceed a value of 1.1). Mean oxygen consumption was determined by averaging oxygen consumption values recorded during the defined steady state period. The *energy cost of walking* (ml/kg m^{-1}), an estimate of energy expenditure per unit of gait speed[5], was calculated by dividing body mass-corrected mean oxygen consumption (ml/kg min^{-1}) by mean overground gait speed (m/min.). As previously described, mean overground gait speed was calculated from all gait speeds recorded throughout the participant's walking trial. Energy cost is time independent and can be compared across individuals and over time, regardless of changes in gait speed.[22, 93] Portable gas analysis systems have been shown to be reliable and valid for measuring oxygen consumption. [21, 23-25, 76-78] The reliability of energy cost during overground walking in our study was determined to be ICC (1,k) = .72. All participants were provided time for walking familiarization around the indoor track with the portable gas analysis system prior to completing the recorded walking trial.

3.2.4.4 Gait Abnormality/Mechanics

Gait biomechanics (abnormalities) were assessed as potential confounders for our study; certain measures (hip and trunk) have been shown to be related to energy cost of walking and stance

time variability.[13, 33, 55] Video recordings of front, back and lateral views of participants were recorded during overground walking, and used for the assessment of gait abnormality and mechanics. Assessors used the modified Gait Abnormality Rating Scale (GARS-m)[94], a 7 item criterion-based observational assessment of gait abnormality. We decided only to use the GARS-m hip item, as it is the only lower extremity GARS-m item shown to be significantly associated with energy cost of walking.[13] A hip score of 0-3 was recorded for each participant; a higher score represents greater biomechanical abnormality (reduced hip extension). The GARS-m has been shown to be a reliable and valid assessment tool for analyzing gait and the psychometric properties of individual item scores have been demonstrated.[94, 95]

3.2.4.5 Questionnaires

Each participant completed a demographic questionnaire and an assessment of co-morbidities. The presence of *co-morbidities* was ascertained using the Co-Morbidity Index[80], which included 18 different disorders, categorized to 8 domains. The total number of positive domains was recorded.

The Gait Efficacy Scale and Survey of Activity and Fear of Falling in the Elderly fear subscale were collected to describe the psychosocial well-being of our sample of older adults. Also, due to the potential to influence gait characteristics, both confidence in walking and fear of falling served as potential cofounders for study analyses.

Confidence in walking was measured using the Gait Efficacy Scale (GES), an index of confidence in walking over various surfaces and conditions. Item scores (ranging from 1= no confidence to 10= complete confidence) for each of the 10 walking conditions are summed for a

total GES score ranging from 10-100 (higher scores indicating greater confidence). The GES has been previously validated as a measure of confidence in walking.[96]

Fear of falling was assessed using the Survey of Activity and Fear of Falling in the Elderly (SAFFE) Fear subscale. The SAFFE Fear subscale is a mean score of fear across 11 different activities rated on a 0-3 scale (0= not worried about falling during a specific activity to 3= very worried about falling during a specific activity). Scale validation and construct validity have been previously established. [97, 98]

3.2.5 Data Analysis

Descriptive statistics (mean and standard deviation) were used to describe primary outcome variables (stance time variability and energy cost of walking) and other physical performance and self-report measures considered possible confounders (gait speed, GARS-m hip, SAFFE fear, GES, Co-morbidity, and age). Bivariate correlations were performed to 1) identify confounders (variables significantly correlated with stance time variability and energy cost of walking) to be used in later regression analyses, and 2) to assess the strength and direction of the relationship between stance time variability and energy cost of walking. Linear regression analysis was completed to further explore unique contributions of primary variables and individual confounders to energy cost of walking. Only variables significantly correlated with both stance time variability and energy cost of walking, or having a well established relationship with both variables per previous studies, were included as confounders in the analysis. Potential confounders were entered simultaneously into the model, followed by stance time variability.

Separate bivariate and regression analyses were performed (not shown in results) to assess the need to include a quadratic term for gait speed, as it has a curvilinear relationship with

energy cost. The quadratic term was not significant with respect to other measures in our bivariate analysis, nor was it a significant contributor to energy cost of walking when added to the regression model (after the linear term for gait speed). Thus, the quadratic term was not used in the final regression analyses. PASW Statistics 18 (SPSS) software was used for the statistical analyses.

3.3 RESULTS

3.3.1 Subject Characteristics

Our study sample, comprised primarily of white females, had a mean age of 76.3 years. Participants were relatively healthy, having little co-morbidity, and appeared confident in their walking abilities (Table 4). Likewise, compared to previously established values, our sample of older adults reported little fear of falling with activity[98], had low gait variability[16], and slightly increased energy cost of walking (age-appropriate normal value ~ .16 ml/kg m^{-1})[5]. Self-selected, usual gait speed was at the high end of our inclusion range of .8-1.0 m/s, and is clinically situated between two "diagnostic levels" levels[99] – mildly abnormal (.6-1.0 m/s) and normal (1.0-1.4 m/s).

Table 4: Characteristics of Study Participants

Characteristics [a]	Participants (n=30) mean (range)	Excluded (n=10) Mean (range)
Age (years)	76.3 (73.8-78.8)	78.7 (72-90)
Gender (% female)	63	100
Race (% White)	90	90
Co-Morbidities (0-8)	2.7 (0-5)	2.8 (0-6)
GES (10-100)	83.1 (77.4-88.9)	77.5 (35-94)
SAFFE Fear (0-3)	.42 (.28-.57)	.43 (0-1.6)
GARS-m (0-3)	1.0 (0-3)	1.0 (0-2)
Gait Speed (m/s)	.98 (.93-1.04)	.95 (.79-1.09)
Stance Time Variability (s)	.026 (.023-.029)	-
Energy Cost of Walking (ml/kg m^{-1})	.17 (.16-.18)	-

[a] Reported as mean (95% confidence interval) unless otherwise indicated, GES = Gait Efficacy Scale, SAFFE Fear = Survey of Activity and Fear of Falling in the Elderly Fear subscale, GARS-m = Modified Gait Abnormality Scale

3.3.2 Correlations

Bivariate correlation coefficients for the two primary variables (stance time variability and energy cost of walking) and potential confounding variables (gait speed, GARS-m hip, SAFFE fear, GES, co-morbidity, and age) are summarized in Table 5. Stance time variability and energy cost of walking were not related in our sample of older adults (r= -.078, p=.683). A scatter plot of the association between the two variables is shown in Figure 11. Three potential confounding variables were significantly associated with stance time variability (gait speed, r = -.512; SAFFE fear, r = .423; and co-morbidity, r = .362), whereas, none of the potential confounders were significantly correlated with energy cost of walking.

Table 5: Correlations between Potential Confounders, Gait Variability, and Energy Cost of Walking

Variable	Correlation Coefficient (p) for Overground Walking:							
	Energy Cost	Stance Time Variability	Gait Speed	GARS-m Hip	SAFFE Fear	GES	Co-Morbidity	Age
Energy Cost	1.00							
Stance Time Variability[a]	-.078 (.683)	1.00						
Gait Speed	-.280 (.134)	-.512 (.004)*	1.00					
GARS-m Hip	.241 (.199)	-.037 (.845)	-.401 (.028)*	1.00				
SAFFE Fear	-.055 (.773)	.423 (.020)*	-.290 (.120)	-.011 (.953)	1.00			
GES[a]	-.228 (.226)	-.164 (.385)	.547 (.002)*	-.308 (.098)	-.317 (.088)	1.00		
Co-Morbidity	.189 (.318)	.362 (.049)*	-.304 (.102)	.132 (.487)	.288 (.123)	-.299 (.224)	1.00	
Age	.182 (.336)	.241 (.200)	-.355 (.054)	.237 (.207)	.198 (.294)	-.350 (.058)	.176 (.352)	1.00

[a]Spearman Rank Correlation (all others are Pearson Correlations); * significant at <.05

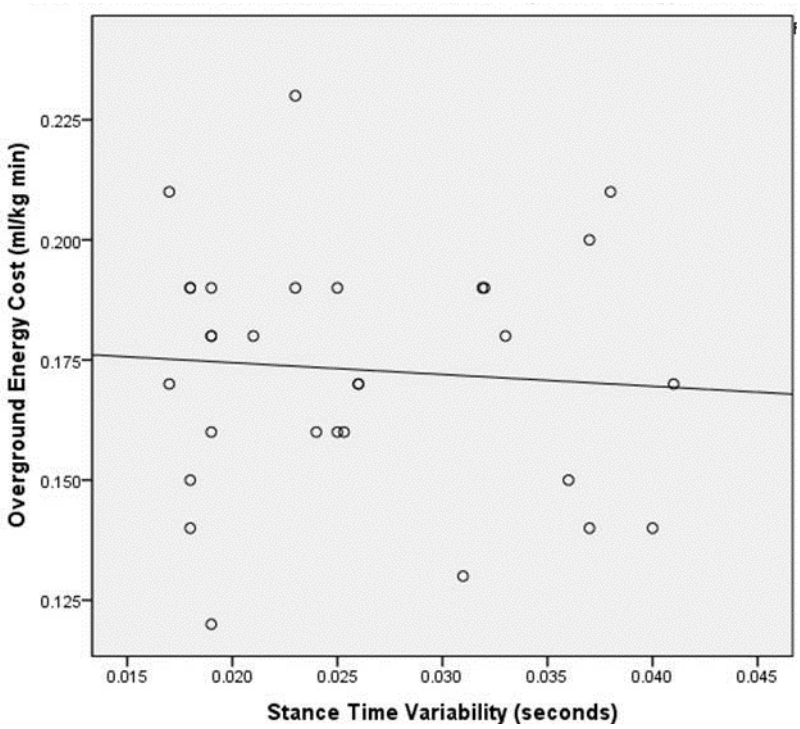


Figure 11: Association between stance time variability and energy cost of walking

3.3.3 Regression Analysis

To assess for individual variable contribution to energy cost of walking, we performed linear regression analysis (controlling for gait speed). Although gait speed was not significantly correlated with energy cost of walking in our study, it was entered into the model based on previous works which showed a strong relationship between the two measures.[4, 6, 29] Based on the bivariate analysis, no other variables were selected to be entered into the regression model. Gait speed was entered into the model first, followed by stance time variability (Table 6). In our sample, gait speed alone was not a significant contributor to energy cost of walking (p=.134). Adding stance time variability in Step 2 failed to provide any additional explanation of energy cost of walking; however, with stance time in the model, gait speed became an independent contributor in explaining some of the variance in energy cost of walking (15%, p=.041).

Table 6: Linear Regression Analysis on Energy Cost of Walking - Controlling for Gait Speed

Models	B	SEB	β (p)	sr^2	R^2 (p)	ΔR^2 (p)
Step 1 Gait Speed	-.001	.001	-.280	.08	.078 (.134)	-
Step 2 Gait Speed Stance time variability	-.001 -1.042	.001 .685	-.453 (.041)* -.321 (.140)	.15 .07	.15 (.109)	.073 (.14)

*p<.05, (n=30); B= beta, SEB= standard error of beta; β= standardized beta, sr^2=squared partial correlation

3.4 DISCUSSION

The main finding of this study was that no significant correlation was found between stance time variability and energy cost of walking. And although gait speed was shown to be a significant contributor to energy cost of walking, the final model was not significant.

Surprisingly, despite the correlations found with stance time variability, none of the potential confounders were related to the energy cost of walking in our sample. The lack of association between gait abnormality (GARS-m Hip) and energy cost of walking is in contrast to work completed by Wert et al. (2010), which reported a positive association ($r=.523$, $p<.01$) between the two variables. Similarly, our study did not show a correlation between gait speed and energy cost of walking ($r= -.280$, $p=.134$), while Wert and colleagues (2010) reported a significant negative association between the two measures ($r= -.286$, $p=.04$). [13]

There are two possible explanations that may provide insight as to why the study results differed from our hypothesis: 1) we may have inadvertently narrowed the range of the two primary variables (STV and energy cost) while trying to preemptively control for gait speed, and 2), energy cost of walking may be multifactorial in nature, of which stance time variability is simply not a main contributor in our sample of older adults.

3.4.1 Influence of Narrow Gait Speed Range

One of the inclusion criteria for our study was a preferred walking speed between 0.8 and 1.0 m/s, a slow gait speed compared to the usual speed of 1.2 m/s for adults.[99] The intent of narrowing the range of gait speeds was to preemptively reduce the potential for gait speed to

influence energy cost of walking, a well established relationship.[3, 4, 29] To address the concern that we may inadvertently reduce the range of stance time variability and energy cost of walking by narrowing the range of gait speeds, we looked at the range of values for the two variables in other datasets where we restricted the range of speeds to 0.8-1.0 m/s. After restricting the gait speed range in other studies, both variables continued to show adequate range. Nonetheless, selecting our sample of older adults based on a narrow gait speed range may have, in fact, reduced our ability to find a significant relationship between stance time variability and energy cost of walking. The range of energy cost for our sample was .12-.23 ml/kg m^{-1}, with an age-appropriate mean (standard deviation) of .17 ml/kg m^{-1} (.03). If we recognize .05 ml/kg m^{-1} as a moderate difference in energy cost (based on a moderate effect size of .5 and a standard deviation of .10 ml/kg m^{-1}, reported from a similar study population[41]), the minimum and maximum values of our range in energy cost only just approach values that would be considered different from the mean [.17 ml/kg m^{-1} ± .05 ml/kg m^{-1} = range of (.12 -.22 ml/kg m^{-1})] . Additionally, due to our small standard deviation, approximately 65% (20/30) of participants had energy cost values within the range of .14-.20 ml/kg m^{-1}, suggesting that most individuals are within the 95% confidence interval for the mean (values less than .12 or greater than .22 ml/kg m^{-1}). Also, only the older adults at the very ends of the .14-.20 ml/kg m^{-1} range differ from one another (.20 ml/kg m^{-1} - .14 ml/kg m^{-1} = .06 ml/kg m^{-1} difference), otherwise most older adults in this range did not differ from one another in terms of energy cost of walking. Upon further inspection, only four individuals (13%, 4/30) had values above .20 ml/kg m^{-1}, of which only one would be considered having a cost value significantly different from the age-appropriate norm (.17 ml/kg m^{-1}). In summary, with regards to energy cost, our sample consisted of a homogeneous group of older adults who – in most cases- were efficient walkers.

If we were to have expanded our range of gait speeds, we may have had more individuals who differed significantly from age-related norms in energy cost, as well as from one another (Firgure 12). VanSwearingen et al (2009) selected their sample of older adults based on a slow preferred walking speed of 0.6 to 1.0 m/s (0.2 m/s greater range than our range), and had a mean (standard deviation) energy cost value of .30 ml/kg m^{-1}(0.10). Not only did the mean value significantly differ from normal (.17 ml/kg m^{-1}) but the standard deviation was large, meaning that 65% of the sample fell within the range of .20 to .40 ml/kg m^{-1}. This suggests that these older adults, had values of energy cost that differed significantly from the mean and from one another. Consequently, 73% (35/48) of the sample were considered as having high energy cost of walking values – compared to our sample of only 13% (4/30).

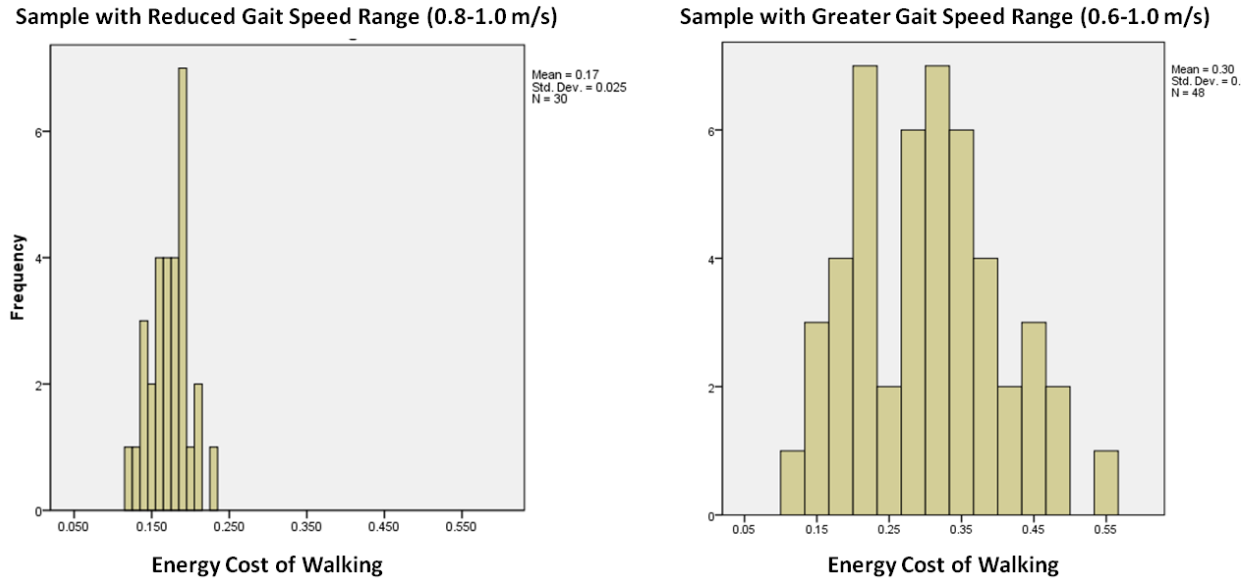


Figure 12: Comparison of range in energy cost of walking

Similar results were observed for our range of stance time variability; only 17% (5/30) of study participants had values of stance time variability considered to be abnormal (≥ .037 s, associated with future disability)[16]. Although the minimum (.017 s) and maximum (.041 s) values of stance time variability appear to create an adequate range, with a mean of .026 s and standard deviation of .01, as mentioned above - the majority of participants fell below a meaningful value of stance time variability (.037 s).

It would appear that restricting the range of gait speeds to between 0.8 and 1.0 m/s limited the number of our participants with "higher" values of STV and energy cost, which in turn resulted in a homogenous group of older adults with regards to these two primary variables. Having a homogenous sample with a restricted variable range reduces the strength of association between the two variables and in turn decreases the likelihood of observing any existing relationship between the two variables[100, 101] (Figure 13).

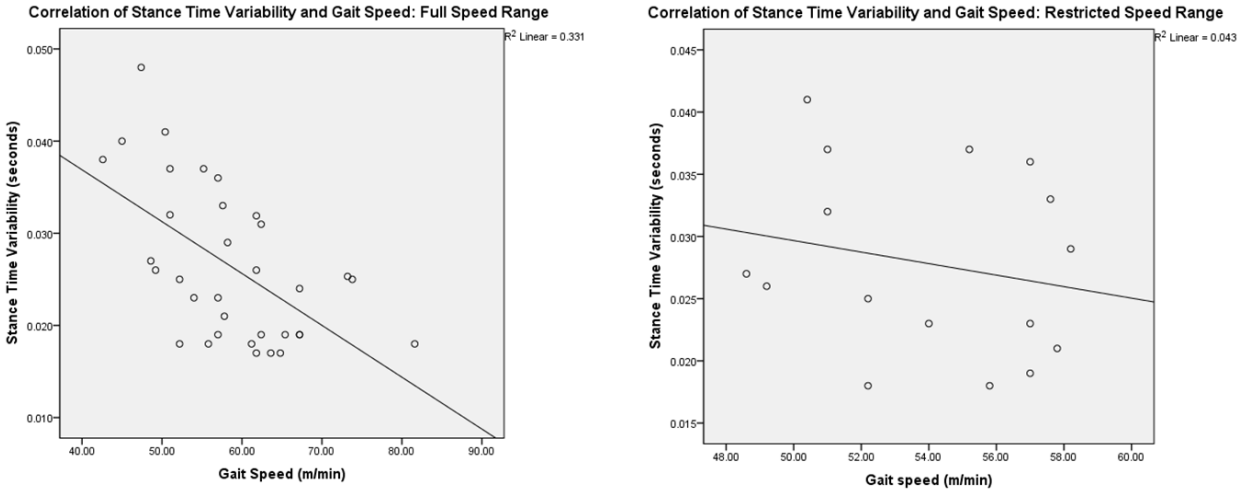


Figure 13: Influence of restricted range on correlations

While none of our older adults were considered to have high values of stance time variability *and* energy cost of walking, a few participants did have high values for at least one of the two measures. One participant presented with a high energy cost of walking (.23 ml/kg m^{-1}), but unlike what we hypothesized, their stance time variability was low (.023 s). This participant was the oldest member of our study (91 years) and also had one of the highest total GARS-m scores (12/21), both of which would suggest he should have a high value of variability. But it would appear that this participant was efficient in managing their many biomechanical abnormalities (while walking). Similarly, two participants had higher stance time variability but low energy cost of walking. The only distinguishing characteristics were, both had higher levels of fear of falling (SAFFE fear = .8 and 1.0), lower confidence scores (GES = 76/100 and 63/100), and lower oxygen consumption values (6.49 ml/kg min^{-1} and 7.83 ml/kg min^{-1}). Since fear of falling has been associated with greater muscle coactivation in some older adults, one might suspect these individuals would have a higher cost of walking; however, this was not the case, as both participants had energy cost values of .14 ml/kg m^{-1}. Collectively assessing these three participants would suggest that stance time variability and energy cost of walking have a negative relationship, opposite of what we hypothesized. However, a larger sample of older adults with greater variation in stance time variability and energy cost of walking is needed to better examine the relationship between these two variables.

3.4.2 Contribution of Other Factors Towards Energy Cost of Walking

The findings of our study may also be attributed to the fact that energy cost of walking may be influenced by a variety of factors, a view posed by other researchers as well [3, 10, 11, 40], of which

gait variability (in our case – stance time variability) is not a primary contributor. Increased muscle coactivation, altered gait biomechanics, changes in spatial and temporal gait characteristics, and cellular health have all been implicated as possible factors related to the age-associated increase in energy cost of walking.

Hortobagyi and colleagues (2009) suggested that an increase in agonist muscle activation and antagonist muscle coactivation (elicited by neural factors) during the gait cycle may account for an increase in the cost of walking seen in older adults compared to young.[11] The same group later reported in another study (2011) that agonist muscle activation accounted for 31% of cost of walking, while antagonist muscle coactivation accounted for 43% of the cost.[10] Coactivation rates in general were higher for older adults compared to young.[10] Hortobagyi's findings were consistent with those of Peterson et al (2010), who reported higher thigh coactivation, versus the shank, among older adults compared to young – especially just before and after heelstrike.[12] Peterson and colleagues also reported an association between total coactivation (shank and thigh) and cost of walking, r=.55 (walking speed 1.12 m/s).

The greater cost of walking observed with coactivation is thought to result from the "age-related" increase in antagonist muscle coactivation – which in turn requires the respective agonist muscles to produce additional force by recruiting a greater portion of muscle mass. The increased recruitment of muscle mass would require greater metabolic energy to offset the opposing force of the antagonist muscles.[10] We propose that the increase in energy demand, arising from the altered timing and contraction of muscle function "associated" with stance time variability, is less than the magnitude of increased energy demand resulting from increased coactivation, and as such, may be too subtle to account for an identifiable change in energy cost of walking. Perhaps stance time variability, or gait variability in general, is a precursor to

increased co-activation; as variability worsens over time, co-activation begins to increase as an adaptation to a more inconsistent gait pattern. Until gait variability has crossed a given threshold, and co-activation has become greater than normal, energy cost of walking may not be influenced. However, we are unaware of any studies that have investigated the potential relationship between stance time variability and co-activation.

Other researchers have associated energy cost of walking with certain age-related abnormalities in gait biomechanics and posture.[13, 33] Wert and colleagues (2010) found that greater hip abnormality (reduced extension) accounted for 22% (p=.002) of the energy cost of walking in their sample of older adults. In our study, GARS-m hip was not significantly correlated with energy cost of walking (r = .241, p=.20); the contrast from previous work is likely due to the homogeneous nature of our sample, where the majority of participants (63% or 19/30) had a "mild" GARS-m hip score of 1, followed by 27% (8/30) having a score of "moderate" 2, and the remaining 10% (3/30) having no hip abnormality.

Changes in trunk position have also come into question as possibly contributing to the energy cost. The influence of trunk position has been assessed by Saha and colleagues (2007); they noted an increase in energy cost as subjects moved from more erect postures to more flexed trunk positioning.[33] Further research is needed to determine whether trunk position continues to play a role during walking, as the previous study only reported contributions to cost of standing.

The potential role of temporal and spatial gait characteristics in contributing to energy cost of walking has also been investigated. Malatesta and colleagues (2003) tested whether the higher cost of walking in older adults was due to greater gait instability (defined as greater stride time variability). Their results were similar to our study; although stride time variability was

significantly greater in their sample of older adults compared to young, they did not find a significant correlation between stride time variability and energy cost of walking. [3]

Wert and colleagues (2010) found that the specific gait characteristics, cadence and step width, could also independently contribute to energy cost of walking. Step width is believed to represent balance control[49]; therefore, it is likely that individuals presenting with extremes in step width, may have higher levels of coactivation – potentially resulting in a higher cost of walking. The higher cost of walking associated with cadence is likely due to the increased muscular demand resulting from taking more steps over a given distance. This would increase the frequency of firing for lower limb and trunk agonists and antagonists, inciting a greater overall metabolic demand.

Finally, others have suggested events that occur at the cellular level as possibly playing a role in the cost of walking[10], particularly as it pertains to implications of aging and free radical effects on mitochondrial health and energy production.[102] Both aging and continued exposure to free radicals, are thought to result in a reduced number of mitochondria as well as an altered efficiency with which mitochondria function, resulting in a poorer ability to generate "energy". [102] Given the clinical nature of our study, along with the financial restraints of a pilot study, we were unable to assess concurrent cellular health of our participants.

As more studies begin to find associations with energy cost of walking, the feasibility of combining such multifaceted variables, in order to assess more comprehensive contributions, becomes more of a reality. A broader understanding of energy cost of walking would take us closer to developing improved therapeutic interventions to enhance the efficiency of walking in older adults.

3.4.3 Is Stance Time Variability "Variable"?

Little is known about the "variability" of variability; are individuals consistently variable or will their variability change in magnitude from session to session? To place the relevance of this concept in the framework of our study; it would benefit us to know whether our participants' measure of stance time variability were stable (consistent) or unstable (inconsistent). Participants with unstable measures of stance time variability may have had low values during data collection, but may have had higher values if the measure was taken the next day, or even during another trial.

In terms of instrument reliability, we reported fair consistency with our footswitch system (ICC= .51), a value consistent with ICC's of stance time variability recorded from the gold standard (GaitMat, ICC=.50). However, since the ICC for stance time variability is considerably lower than that of other gait characteristics (step length ICC=.96, step width ICC=.97, step time ICC=.96, step width variability ICC=.79), it is possible that there is inconsistency in stance time variability within our participants.

Inconsistency of performance has been widely studied in psychology regarding cognitive behavior. Slifkin & Newell (1998) defined individuals whose performance varies little from session to session, regardless of level, as consistent, while those individuals whose performance varies dramatically from session to session as inconsistent.[103] Inconsistency of performance on simple cognitive tasks has been considered a measure of variability in central nervous system functioning[104], and has been found to be related to age, injury, health, and intelligence.[105] Greater inconsistency seems to be a marker of impending decline or low functionality.[104] Greater inconsistency from session to session was reported to be more pronounced in a sample of

older adults who had lower scores on a measure cognitive performance than for those who had higher (better) scores on the measure.[106]

Unpublished work from our lab looked at the issue of inconsistency regarding stance time variability. Older adults, who were either consistently variable or consistently not variable in measures stance time variability, performed the worst and best respectively, on measures of executive function. The group who had inconsistent measures of stance time variability between sessions ended up in the middle. This group could be considered to be in "transition", moving back and forth between normal (consistently not variable) and abnormal (consistently variable).

It is possible that some older adults in our study are in transition, moving from normal to more abnormal variability but not yet considered consistently abnormal – thus, a group "in flux". As such, stance time variability values may *fluctuate* back and forth from "poor" to "good" within walking trials and between measurement sessions. This may be one of the reasons STV test-retest reliability measures are only "fair" compared to mean measures of gait characteristics. Older adults who are inconsistently variable may not demonstrate similar relationships with other variables (ie. stance time variability and energy cost) as those who belong to a more defined group (normal or abnormal). Future studies are needed to assess the variability of stance time variability and establish potential boundaries/cut-points that may identify individuals who are likely to be consistent and inconsistent.

3.4.4 Study Strengths

Our study demonstrated the feasibility of recording concurrent *overground* measures of stance time variability and oxygen consumption in older adults. Collecting overground measures

requires participants to wear portable equipment in order for data to be collected. All of the older adults in our sample were able successfully complete their 4-6 minute walking session without complaints pertaining to wearing the portable devices. Furthermore, we were able to demonstrate that gait speeds throughout the course of overground walking can be consistent among older adults, even those with slow gait speeds, as with our sample. We calculated the energy cost of walking using gait speeds recorded during various time-points during testing and found no significant difference in cost based on the different speeds. As a method of keeping gait speed consistent for studies collecting measures of gait and oxygen consumption, treadmills are often selected as the mode of activity. However, walking on the treadmill is not a common activity for many older adults and may not be as accurate of a measure of energy cost of walking as overground walking. Our study suggests that older adults *can* maintain a consistent preferred walking speed overground, offering a more natural mode of investigating gait and energy cost of walking for future studies.

3.4.5 Limitations

Our small sample of older adults was selected based on a narrow range of gait speeds, which is not representative of all older adults. Therefore our results should be generalized to like populations. The intent of limiting the range of gait speeds of our participants was to preemptively control for gait speed differences between participants; we also were interested in working with older adults with slow gait speed, as stance time variability and energy cost of walking tend to be higher in individuals who have a slower gait speed. As we understand more about the *why* and *how* gait changes within this "declining" group, we can better focus our interventions aimed at re-establishing efficient mobility in our older adults. Our study also had

too few participants who were considered to have variable gait and higher than normal energy cost of walking, potentially due to our decision to restrict participation to individuals with a narrow range in gait speed. Thus, we recognize that the relationship between stance time variability and energy cost of walking may be different in a sample with a greater range within these two variables.

3.4.6 Conclusion

Walking is a more costly activity for older adults to perform compared to young. Research suggests that the increase in energy demand likely arises from an increased demand placed on muscles involved during gait, particularly during stance phase. Although stance time variability appeared a likely candidate to account for the increase in energy cost of walking observed in older adults, our study did not find a relationship between stance time variability and energy cost of walking. While stance time variability did not show a direct relationship with energy cost of walking, its association with mobility disability, found in previous studies, suggests that it impacts gait in some manner. Future research is needed to explore potential moderating influences of stance time variability on factors known to have a relationship with energy cost of walking (ie. muscle coactivation, gait biomechanics, gait speed). Furthermore, it is likely that energy cost of walking has many contributors, thus research should begin to focus on the interplay between various factors and their shared influence on the energy cost of walking in older adults.

4.0 CHAPTER IV – THE RELATIONSHIP BETWEEN STANC TIME VARIABILITY AND ENERGY COST OF WALKING ACROSS DIFFERENT WALKING CONDITIONS

4.1 INTRODUCTION

Older adults expend more energy during walking than young adults, even at similar preferred walking speeds. [5, 6] When basic tasks such as walking exact such high energy cost, some older adults are left with very little "reserve" for performing the rest of their vital tasks; that limitation, in turn, may lead to reduced physical activity and potential loss of independence. [2, 9, 11, 40] To that end, Wert and colleagues (2009, 2010) found higher than normal energy cost of walking was related to poorer self-report of function in older adults with mobility disability.[7]

The greatest demand for energy (oxygen) during walking comes from muscle activity[1, 18], as muscles perform many energy-demanding tasks by operating as motors, tensile struts, and brakes. [85-88] The stance phase of gait appears to be the period of walking that demands the greatest amount of energy from muscles; Griffin et al (2003) found active muscle volume required to generate force on the ground and the rate of generating this force accounted for >85% of the increase in metabolic rate.[14] As such, we might suspect that higher energy cost of walking observed in older adults may be the result of an age-related change (increased) in muscle demand during the stance phase of gait. One gait characteristic known to: 1) occur during stance phase[42], 2) have higher (more abnormal) values among older adults compared to young[56], and 3) is suspected to result from age-related alterations in muscle function, is *stance time variability* - the fluctuation in stance time from one step to the next.

Although little is known about the neural control of stance time variability, some researchers have attributed the age-related progressive loss of motor abilities (exhibited in gait characteristics – like stance time variability) to "changes" in the input or signaling to spinal mechanisms responsible for automatic stepping. [47, 89, 90] Similarly, changes or disruptions to the peripheral signal for stepping, arising from sensory nerves of the anterior thigh[48], have also been implicated in altering the automatic stepping pattern via age-related changes in gait biomechanics. The combined "altered" contribution from central nervous system and peripheral input onto spinal mechanisms is thought to result in altered duration and frequency of muscle activity during gait, potentially creating an increase in demand (energy) from muscles during walking.[47] Thus, stance time variability may contribute to the energy cost of walking by way of fluctuating muscle force production and rate of muscle activity, a result of age-related alterations of central and peripheral signals for stepping. Despite the evidence suggesting stance time variability could be a viable contributor to the higher energy cost of walking in older adults, no studies, to our knowledge, have assessed the relationship between stance time variability and energy cost of walking in older adults.

Like energy cost of walking, stance time variability has been linked to poorer function in older adults. Higher values of stance time variability can be detrimental to older adults; Brach and colleagues (2007) reported a 13% higher incidence of mobility disability among older adults who differed in stance time variability by only .01s.[16] Furthermore, the same group also observed, older adults who reported an increase in mobility disability (decline in their walking ability) over the course of one year had an increase in their measure of stance time variability.[56] In contrast, older adults who reported no change in mobility (walking ability) over the course of one year had stable measures of stance time variability. [56] Although there are detrimental

consequences to having higher values of stance time variability for older adults, some preliminary research has suggested that stance time variability may be improved (reduced). In a small sample of community-dwelling older adults, with slow and variable gait, walking on the treadmill at a self-selected preferred walking speed resulted in lower values of stance time variability compared to walking overground at a preferred speed.[65] In fact, stance time variability was reduced to levels below .037 s, the value above which was associated with an increase in mobility disability in older adults.[16] Although there is some evidence to support that walking overground with a Rollator ™ improves kinetic and kinematic characteristics of gait in some older adults [74], no studies (to our knowledge) have specifically assessed its' ability to impact stance time variability. On the other hand, slow walking, like treadmill walking, has been reported to influence stance time variability; Brach et al (2006) found that older adults who were asked to walk slowly, had greater stance time variability compared to walking at preferred and fast speeds.[17]

Since stance time variability appears to differ depending on walking condition (fast-slow, overground-treadmill), it is possible a relationship between stance time variability and energy cost of walking could also differ across walking conditions (interaction) – or, regardless of condition-specific influences on stance time variability, it may remain consistent (no interaction). Furthermore, given the amenable nature of stance time variability to speed and mode of walking, subsequent changes (decreases or increase) in energy cost of walking may be expected when changes (decrease or increase) occur in stance time variability. Few studies, if any, have looked at the relationship between stance time variability and energy cost of walking across different walking conditions or have assessed whether a change in stance time variability leads to a subsequent (and like) change in energy cost of walking.

The purpose of this study is twofold; 1) to assess the relationship between stance time variability and energy cost of walking across different walking conditions in older adults and 2) to assess whether changes in stance time variability (increase or decrease) result in subsequent and similar changes in energy cost of walking. We hypothesize that all walking conditions will show a positive, linear relationship between stance time variability and energy cost of walking. Additionally, we hypothesize that a change (decrease) in stance time variability will result in subsequent and like change (decrease) in energy cost of walking; likewise, any condition-induced increase in stance time variability will result in a subsequent increase in energy cost of walking.

Since researchers, devoted to investigating gait characteristics and energy cost of walking in older adults, perform tests over a variety of walking speeds and conditions, it is important that any relationship explaining a portion of energy cost of walking is "consistent" across all conditions in which research is performed. Failure to do so would limit the generalizability of findings to specific walking conditions only, reducing the applicability of findings to fewer older adults. Similarly, determining whether changes in energy cost of walking are associated with changes in stance time variability will enhance support that the relationship is robust between the two measures; additionally, such evidence could provide another rehabilitative component for physical therapy interventions aimed at improving the efficiency of walking in older adults.

4.2 METHODS

4.2.1 Study Design

We used a cross-sectional study design to assess the relationship between stance time variability (varied by walking condition) and energy cost of walking.

4.2.2 Participants

Individuals were eligible to participate in this study if they: were 65 years of age or older, reported being able to walk 4-6 minutes non-stop at their preferred walking pace, without the use of an assistive device and without the assistance of another person, were willing to walk on a treadmill with upper extremity support, and had a usual overground walking speed in the range of 0.8-1.0 m/s.

Participants were excluded from participating in our study if they: were unable to provide informed consent, had concomitant neuromuscular disorders that impair movement, had a diagnosis of cancer with active treatment or severe pulmonary/cardiac disease, or were recently hospitalized for a life-threatening illness or major procedure. Forty participants were recruited from previous studies of mobility and balance in older adults that took place within the Pittsburgh Claude D. Pepper Older Adults Independence Center (Pittsburgh OAIC). Forty participants were recruited from previous studies of mobility and balance in older adults that took place within the Pittsburgh Claude D. Pepper Older Adults Independence Center (Pittsburgh OAIC). The following shows participants with complete data per walking condition: Overground = 30, Rollator = 32, Treadmill = 28, and Treadmill Slow = 22. Only those with

complete data were included in the analyses. The individuals not included in the analyses were missing either footswitch or oxygen consumption data due to equipment malfunction. All participants who agreed to participate in this study signed a consent form approved by the Institutional Review Board of the University of Pittsburgh.

4.2.3 Procedures

Participants completed a brief walking assessment to determine eligibility based on the gait speed inclusion criteria (gait speed between 0.8-1.0 m/s). Once eligibility was confirmed, participants had height, weight, blood pressure and heart rate assessed, and completed a series of questionnaires regarding confidence in walking and self-reports of fear of falling. Next, oxygen consumption and temporal gait characteristics were recorded during four different, randomized walking conditions, each lasting 4-6 minutes. Rest periods were provided between walking trials. Participants were also videotaped during a segment of each walking condition, which was later used for assessment of gait abnormalities and biomechanics.

Walking Conditions:

1) *Overground Walking* – preferred walking speed, around a flat 150 foot indoor walking track – without the use of any assistive devices.

2) *Rollator* – preferred walking speed, around a flat 150 foot indoor walking track using a Rollator (4-wheeled rolling walker). The rollator was adjusted to each subject in an upright standing position with the arms hanging down along the body and handles level with the ulnar styloid process. Instructions were provided for proper body positioning and hand placement.

The Rollator was selected as a possible means of decreasing gait variability overground, as it was speculated to act as a possible external cue for smooth, continuous walking or to facilitate improved biomechanics – which may in turn improve peripheral signaling for stepping.[48] Little research exists regarding the influence of using a rollator for walking on specific gait mechanics in older adults with slow gait. The few studies that exist have reported improvements in gait speed and step length.[74, 107]

3) *Treadmill Walking* – preferred (self-selected) walking speed on a treadmill. Speed was not matched to overground walking speed, but rather chosen by each participant based on how similar it felt to their overground pace. This was done in an attempt to limit the novelty (walking faster or slower than their "usual") of the walking condition which may influence our outcome measures. Each participant was encouraged to walk as close to their comparable overground speed as possible, however, participants ultimately chose the final speed based on their perception of what felt similar. Instructions were provided for proper body positioning and hand placement. The treadmill was selected as a means of assessing the effect of reduced stance time variability on energy cost of walking. Previous research has shown that walking on a treadmill reduces gait variability compared to overground walking[63-65].

4) *Slow Treadmill Walking* – performed at a speed of 0.4 mph (0.18 m/s) less than the self-selected, preferred pace selected for the Treadmill condition. Instructions were provided on proper body positioning and hand placement. Slow Treadmill Walking was used in this study to assess the effect of increased gait variability on the energy cost of walking. Dingwell et al. (2001, 2006) and others have reported an increase in gait variability when comparing usual overground walking to slow walking.[17, 62, 108] We were able to demonstrate similar findings in a small preliminary study using a treadmill; individuals demonstrated greater values of stance time

variability when the preferred walking speed was reduced by .4 mph, reductions in speed less than .4 mph did not result in higher values of stance time variability. Slow treadmill walking was selected, versus slow overground walking, as we wished to maintain a consistent "slowness" throughout the 4-6 minute duration of the walk – something we were uncertain could be achieved overground.

Participants were given a period of time to "practice" each of the four walking conditions while wearing the portable equipment prior to starting the first condition; this time also served as an opportunity to adjust and modify the equipment for optimal fit and comfort as well as an opportunity to determine the participant's preferred and slow speeds for the treadmill conditions.

4.2.4 Measures

4.2.4.1 Gait Characteristics

Temporal gait characteristics were recorded during each 4-6 minute walking condition using a portable footswitch system. Footswitches (1 inch x 1 inch membrane switches with a stainless steel metal dome that depresses with 20 ounces of force) were placed on the soles of the participants' shoes – at the first metatarsal head and laterally at site of heel strike (Figure 9).

Data were sampled at 100 Hz with 16 byte precision, using a Class 1 Bluetooth protocol to transmit data to a desk-top computer system. Temporal measures of gait (stance time) were determined using a custom written LabView program. Using MATLAB, heel-contact and toe-off instances were identified (Figure 10) using thresholding methods (minimum set at 2.5 Hz). Heel-contacts and toe-offs were then used to calculate stance time while the standard deviation

of all steps was used as the measure of stance time variability.[56] Stance time was data synced, using computer time, with oxygen consumption to allow for concurrent recording of both measures. Thus, stance time variability was determined during the same time period as oxygen consumption (and ultimately – cost of walking) at steady state. The reliability of our footswitch system to measure stance time variability during overground walking was consistent with values of stance time variability reported from the GaitMat (ICC = .51 vs. ICC = .50, respectively), the gold standard in gait measurement.

4.2.4.2 Condition Specific Gait Speed

Overground: For both Overground and Rollator walking conditions, two 4-meter distances were marked at opposite ends of a 150 foot indoor track. Time to complete each 4 meter distance was recorded for ~ 75% of the laps completed during the overground walking condition. On occasions where 4-meter times were not collected, researchers were monitoring collection of footswitch and oxygen consumption data. Gait speed (meters/second) was calculated by dividing the distance (4 meters) by the recorded time (in seconds) for each 4 meter pass; the mean gait speed for the condition was calculated from all gait speeds recorded during the condition. Individual gait speeds were consistent within the walking condition; in a sub-sample of our study (n=20), we calculated energy cost of walking (for each participant) using the (within subject) mean of all gait speeds collected during the overground condition, and compared it to the energy cost of walking calculated using a single measure of gait speed collected towards the end of the walking condition. There was no difference between the two measures of gait speed (mean difference = .005 m/s, t_{paired}= -.54, p=.59) or cost (mean difference = -.001 ml/kg m^{-1},

t_{paired}= -.49, p=.63), suggesting the gait speeds were similar. The reliability of overground gait speed for our study was, ICC (1,k) =.84, and for Rollator gait speed ICC (1,k) = .87.

Treadmill: Gait speeds for both treadmill conditions (Treadmill and Treadmill Slow) were recorded in miles per hour (mph); at the end of the familiarization period, participants were asked to select the speed that felt similar to their usual overground walking speed –this was recorded as their Treadmill condition speed. Speed was reduced by a standard 0.4 mph across all participants for the Treadmill Slow condition.

4.2.4.3 Oxygen Consumption

Oxygen consumption was determined by indirect calorimetry and analysis of expired gases.[22, 91] A portable metabolic measurement system (VO2000, MedGraphics Corp.™, Figure 7) was used to measure oxygen consumption (ml/kg min) for each of the four walking conditions previously described. Participants wore a light-weight, portable gas analysis system (VO2000, MedGraphics Corp. ™) attached to a nylon harness worn over the shoulders, and a neoprene mask which fit comfortably over the nose and mouth (total weight, 4 lbs).

Mean rate of oxygen consumption and carbon dioxide production was determined over one to three minutes after physiologic steady state was reached (approximately 2-3 minutes after the initiation of walking).[1, 41, 92] Parameters for the determination of steady state were defined as a combination of the following: the plateau of oxygen consumption (values not to fluctuate more than 3 ml/kg min.) and an RER value that remained stable and did not to exceed a value of 1.1. Mean oxygen consumption was determined by averaging oxygen consumption values recorded during the defined steady state period. The *energy cost of walking* (ml/kg m^{-1}), an estimate of energy expenditure per unit of gait speed[5], was calculated by dividing body mass-corrected mean

oxygen consumption (ml/kg min^{-1}) for each condition by the respective mean gait speed (m/min.). As previously described, mean gait speeds were calculated from all gait speeds recorded throughout the participant's walking trial. Energy cost is time independent and can be compared across individuals and over time, regardless of changes in gait speed.[22, 93] Portable gas analysis systems have been shown to be reliable and valid for measuring oxygen consumption. [21, 23-25, 76-78] The reliability of energy cost during walking in our study was determined to be ICC (1,k) = .72. All participants were provided time for walking familiarization around the indoor track with the portable gas analysis system prior to completing the recorded walking trial.

4.2.4.4 Gait Abnormality/Biomechanics

Gait biomechanics (abnormalities) were assessed as potential confounders for our study; certain measures (hip and trunk) have been shown to be related to energy cost of walking and stance time variability.[13, 33, 55] Video recordings of front, back and lateral views of participants were recorded during each walking condition, and used for the assessment of gait abnormality and mechanics. Assessors used the modified Gait Abnormality Rating Scale (GARS-m)[94], a 7 item criterion-based observational assessment of gait abnormality. We decided only to use the GARS-m hip item, as it is the only lower extremity GARS-m item shown to be significantly associated with energy cost of walking.[13] A hip score of 0-3 was recorded for each walking condition; a higher score represents greater biomechanical abnormality (reduced hip extension). The GARS-m has been shown to be a reliable and valid assessment tool for analyzing gait and the psychometric properties of individual item scores have been demonstrated.[94, 95]

4.2.4.5 Questionnaires

Each participant completed a demographic questionnaire and an assessment of co-morbidities; both measures were used to describe our specific sample of older adults while total number of positive co-morbidities also served as a potential covariate in the analyses. The presence of *co-morbidities* was ascertained using the Co-Morbidity Index[80], which included 18 different disorders, categorized to 8 domains. The total number of positive domains was recorded.

GES and SAFFE fear subscale were collected to describe the psychosocial well-being of our sample of older adults. Also, due to their potential to influence gait characteristics, both confidence in walking and fear of falling served as potential covariates for study analyses.

Confidence in walking was measured using the Gait Efficacy Scale (GES), an index of confidence in walking over various surfaces and conditions. Item scores (ranging from 1= no confidence to 10= complete confidence) for each of the 10 walking conditions are summed for a total GES score ranging from 10-100 (higher scores indicating greater confidence). The GES has been previously validated as a measure of confidence in walking.[96]

Fear of falling was assessed using the Survey of Activity and Fear of Falling in the Elderly (SAFFE) Fear subscale. The SAFFE Fear subscale is a mean score of fear (worry) of falling from 11 different activities rated on a 0-3 scale (0= not worried about falling during a specific activity to 3= very worried about falling during a specific activity). Scale validation and construct validity have been previously established. [97, 98]

4.2.5 Data Analysis

Descriptive statistics (mean and standard deviation) were used to describe primary outcome measures (stance time variability and energy cost of walking) and other physical performance and self-report measures considered possible covariates (gait speed, GARS-m hip, SAFFE fear, GES, Co-morbidity, and age). Bivariate correlations were performed for each walking condition in order to 1) identify covariates (variables significantly correlated with stance time variability and energy cost of walking) to be used in later regression analyses, and 2) to assess the strength and direction of the relationship between stance time variability and energy cost of walking. Regression analyses were completed to further explore unique contributions of primary variables and individual covariates to energy cost of walking for each walking condition. Only the variables significantly correlated with both stance time variability and energy cost of walking were included as covariates in the analyses. Mixed-effects models (unadjusted and adjusted for gait speed) were used to compare stance time variability and energy cost of walking across the four walking conditions. Energy cost of walking and stance time variability were used as the response variable, walking condition (Overground, Rollator, Treadmill, and Treadmill Slow) as a categorical fixed effect, and participants as a random effect. Post-hoc pair-wise mean contrasts were used to estimate between-condition differences of interest (Overground vs Rollator, Overground vs Treadmill, Rollator vs Treadmill, and Treadmill vs Treadmill Slow).

Separate bivariate and regression analyses were performed (not shown in results) to assess the need to include a quadratic term for gait speed, as it has a curvilinear relationship with energy cost. For Overground, Rollator, and Treadmill conditions, the quadratic term was not significant with other measures in our bivariate analysis, nor was it a significant contributor to energy cost of walking when added to the regression model (after the linear term for gait speed).

Thus, the quadratic term was not used in Overground, Rollator, and Treadmill regression analyses. For Treadmill Slow, the quadratic term significantly added to the model containing the linear term for gait speed (ΔR^2=.097, p=.004), suggesting a curvilinear relationship; thus, both terms were included in subsequent analyses for Treadmill Slow.

4.3 RESULTS

Our sample of older adults had a mean age of 76.3 years and slow gait (0.99 m/s). Participants were confident with walking (GES>70)[109] and had relatively low fear of falling (SAFFE fear < .40).[97, 110] Additionally, overground stance time variability (.026 s) was lower than the value associated with increased risk of mobility decline (.037 s)[16] and energy cost of walking overground was within the range of age-appropriate norms of .16-.17 ml/kg m^{-1}.[5, 6]

Table 7: Characteristics of Total Sample

Characteristic [a]	Value
Age , years	76.3 (6.8)
Gender, % female (n)	63.3 (19)
Race, % White (n)	90 (27)
Co-morbidities, 0-8	2.7 (1.2)
GES, 10-100	83.1 (15.5)
SAFFE Fear, 0-3	0.42 (.37)
GARS-m Hip [b], 0-3	1.0 (0-3)
Gait Speed [b], m/s.	.99 (.14)
Stance Time Variability [b], s	.026 (.01)
Energy Cost [b], ml/kg m^{-1}	.17 (.03)

[a] reported as mean (standard deviation) unless otherwise noted
[b] recorded during overground walking condition; GARS-m score for questions 5 (hip) – reported a median (range)

4.3.1 Assessment of Stance Time Variability – Energy Cost of Walking Relationship across Various Walking Conditions

Mean values for gait speed, stance time variability, and energy cost of walking were reported for each of the four walking conditions, Table 8. Due to equipment difficulties during data collection, some participants had incomplete measures of variability and/or oxygen consumption for some walking conditions – thus sample size varied per condition. However, individual sub-analysis (not shown) for each walking condition showed similarity of measures between older adults included in the analyses and their counterparts not included.

Table 8: Mean (SD) Measures per Walking Condition

Condition	Gait Speed m/s	Stance Time Variability s	Energy Cost ml/kg m^{-1}
Overground (n=30)	.99 (.14)	.026 (.01)	.17 (.03)
Rollator (n=32)	.93 (.16)	.028 (.01)	.17 (.03)
Treadmill (n=28)	.77 (.17)	.022 (.01)	.23 (.06)
Treadmill slow (n=22)	.61 (.18)	.027 (.01)	.25 (.07)

4.3.1.1 Correlations for Individual Walking Conditions

Overground walking correlations revealed no significant association between stance time variability and energy cost of walking, r=-.078, p=.68. Stance time variability was associated with three other variables, gait speed (r =-.512, p=.004), SAFFE fear (r =.423, p=.02), and Co-morbidity (r =.362, p=.049) (Table 9). Gait speed was correlated with energy cost of walking (r=-.280), in the expected direction [4, 13, 29], but did not achieve significance (p=.13).

Table 9: Unadjusted Correlations for Overground Walking Condition

Variable	Correlation Coefficient (p) for Overground Walking:							
	Energy Cost	Stance Time Variability	Gait Speed	GARS-m Hip	SAFFE Fear	GES	Co-Morbidity	Age
Energy Cost	1.00							
Stance Time Variability[a]	-.078 (.683)	1.00						
Gait Speed	-.280 (.134)	-.512 (.004)*	1.00					
GARS-m Hip	.241 (.199)	-.037 (.845)	-.401 (.028)*	1.00				
SAFFE Fear	-.055 (.773)	.423 (.020)*	-.290 (.120)	-.011 (.953)	1.00			
GES[a]	-.228 (.226)	-.164 (.385)	.547 (.002)*	-.308 (.098)	-.317 (.088)	1.00		
Co-Morbidity	.189 (.318)	.362 (.049)*	-.304 (.102)	.132 (.487)	.288 (.123)	-.299 (.224)	1.00	
Age	.182 (.336)	.241 (.200)	-.355 (.054)	.237 (.207)	.198 (.294)	-.350 (.058)	.176 (.352)	1.00

[a]Spearman Rank Correlation (all others are Pearson Correlations); * significant at <.05

Similarly, no associations were found between stance time variability and energy cost of walking during the Rollator condition (Table 10). Likewise, no other variables were associated with energy cost of walking. However, gait speed (r_{SR}= -.83, p=.001), GES (r_{SR}= -.380, p=.032), and Co-morbidity (r_{SR}= .462, p=008) were related to stance time variability.

Table 10: Unadjusted Correlations for Rollator Walking Condition

Variable	Correlation Coefficient (p) for Rollator Walking:							
	Energy Cost	Stance Time Variability	Gait Speed	GARS-m Hip	SAFFE Fear	GES	Co-Morbidity	Age
Energy Cost	1.00							
Stance Time Variability[a]	.006 (.975)	1.00						
Gait Speed	-.105 (.568)	-.783 (.001)*	1.00					
GARS-m Hip	.053 (.775)	.019 (.918)	-.310 (.084)	1.00				
SAFFE Fear	.136 (.458)	.278 (.124)	-.225 (.216)	-.106 (.563)	1.00			
GES[a]	-.084 (.649)	-.380 (.032)*	.478 (.006)*	-.028 (.880)	-.246 (.175)	1.00		
Co-Morbidity	-.103 (.575)	.462 (.008)*	-.396 (.025)*	.187 (.305)	.221 (.224)	-.337 (.060)	1.00	
Age	.048 (.794)	.199 (.275)	-.202 (.268)	.082 (.656)	.063 (.732)	-.413 (.019)*	.294 (.103)	1.00

[a]Spearman Rank Correlation (all others are Pearson Correlations); * significant at <.05

No main relationship was found between stance time variability and energy cost of walking (r_{SR}=.312, p=.106) for Treadmill walking. However, gait speed, SAFFE fear, and GES were all associated with energy cost [r= -.427 (p=.023); r= .394 (p=.038); and r= -.432 (p=.022), respectively]. Gait speed and SAFFE fear also had similar associations with stance time

variability (Table 11). This would suggest that as gait speed decreases, energy cost of walking and stance time variability would increase; whereas lower levels of fear would yield lower cost of walking and stance time variability.

Table 11: Unadjusted Correlations for Treadmill Walking Condition

Variable	Correlation Coefficient (p) for Treadmill Walking:							
	Energy Cost	Stance Time Variability	Gait Speed	GARS-m Hip	SAFFE Fear	GES	Co-Morbidity	Age
Energy Cost	1.00							
Stance Time Variability[a]	.312 (.106)	1.00						
Gait Speed	-.427(.023)*	-.688 (.001)*	1.00					
GARS-m Hip	.285 (.141)	.288 (.137)	-.174 (.376)	1.00				
SAFFE Fear	.394(.038)*	.519 (.005)*	-.388 (.041)*	.403(.034)*	1.00			
GES[a]	-.432(.022)*	-.149 (.451)	.376 (.048)*	-.151 (.443)	-.285 (.142)	1.00		
Co-Morbidity	.072 (.718)	.525 (.004)*	-.249 (.201)	.156 (.427)	.272 (.162)	-.247 (.204)	1.00	
Age	.076 (.700)	.020 (.919)	-.182 (.355)	.423(.025)*	-.028 (.886)	-.373 (.051)	.138 (.484)	1.00

[a]Spearman Rank Correlation (all others are Pearson Correlations); * significant at <.05

Finally, Treadmill Slow was the only condition to show a relationship between stance time variability and energy cost of walking (r_{SR}= .676, p=.001), suggesting that older adults with greater stance time variability exhibited higher values of energy cost during walking. Also, faster gait speeds (r=-.856), lower self-report of fear (r=.622), greater confidence in walking (r=-.466), and lower degrees of hip abnormality (r=.561) were related to a lower energy cost of walking (Table 12); while gait speed, SAFFE fear, and GES showed a similar association with stance time variability.

Table 12: Unadjusted Correlations for Treadmill Slow Walking Condition

Variable	Correlation Coefficient (p) for Treadmill Slow Walking:							
	Energy Cost	Stance Time Variability	Gait Speed	GARS-m Hip	SAFFE Fear	GES	Co-Morbidity	Age
Energy Cost	1.00							
Stance Time Variability[a]	.676(.001)*	1.00						
Gait Speed	-.856(.001)*	-.746(.001)*	1.00					
GARS-m Hip	.561(.007)*	.267(.230)	-.551(.008)*	1.00				
SAFFE Fear	.622(.002)*	.649(.001)*	-.586(.004)*	.203 (.364)	1.00			
GES[a]	-.466(.029)*	-.580(.005)*	.666(.001)*	-.405(.062)	-.272 (.220)	1.00		
Co-Morbidity	-.035(.877)	.310(.160)	-.178(.429)	.256 (.249)	.256 (.250)	-.428(.047)*	1.00	
Age	.340(.121)	.017(.940)	-.306(.167)	.245 (.272)	.060 (.791)	-.542(.009)*	.111 (.622)	1.00

[a]Spearman Rank Correlation (all others are Pearson Correlations); * significant at <.05

Since gait speed was highly correlated with many variables, a separate partial-correlation analysis (not shown) was performed for variables associated with both energy cost of walking and gait speed (stance time variability, GARS-m hip, SAFFE fear, and GES). Once gait speed was controlled for, no significant correlation remained between these variables and energy cost of walking [*Treadmill*: SAFFE fear r= .274 (p=.17); GES r=-.236 (p=.24), and *Treadmill Slow*: stance time variability r_{SR}=-.108 (p=.64); GARS-m hip r=.208 (p=.37); SAFFE fear r=.287 (p=.207); and GES r_{SR}=.307 (p=.18)].

4.3.1.2 Regression Models for Individual Walking Conditions

Individual regression analyses were performed for each walking condition in order to better assess the unique contribution of stance time variability in explaining energy cost of walking, while controlling for covariates. Only variables related to energy cost of walking and stance time variability (bivariate analysis) were entered into the regression models as covariates. Since gait speed has been shown to have a strong, well-documented relationship with energy cost of walking, it was automatically entered into all of the regression models for each walking condition (both as a linear and curvilinear variable). Each model regressed energy cost of

walking on stance time variability after controlling for gait speed and appropriate covariates as identified in bivariate analysis; due to sample size limitations, models were limited to 2-3 variables.

Adding stance time variability to the model did not further explain energy cost of walking, beyond that explained by gait speed, for any of the walking conditions [Overground ΔR^2=.073, p=.14; Rollator ΔR^2=.047, p=.24; Treadmill $\Delta R^2_{(Model\ 1)}$=.025,p=.28, $\Delta R^2_{(Model\ 2)}$ =.032, p=.21; Treadmill Slow $\Delta R^2_{(Model\ 4)}$=.003, p=.64]. As such, stance time variability was not a unique contributor to energy cost of walking (Table 13-16).

Conversely, gait speed was a strong independent contributor to energy cost of walking for Overground (sr^2= 15%), Treadmill ($sr^2_{Model\ 1}$ = 28%, $sr^2_{Model\ 2}$ = 27%), and Treadmill Slow ($sr^2_{Model\ 4}$ = 22%) conditions. Treadmill Slow was the only condition where the quadratic term for gait speed was significant (Table 16).

Similar to our partial-correlation, after accounting for gait speed, GARS-m hip, SAFFE fear and GES all failed to add to the models or provide unique contributions to energy cost for Treadmill or Treadmill Slow conditions (Tables 15, 16).

Table 13: Regression of Energy Cost for Overground Walking Condition

Model 1 (n=30)	B	SEB	β	sr^2	R^2	ΔR^2
Step 1					.078	-
Gait Speed	-.001	.001	-.280	8%		
Step 2					.151	.073
Gait Speed	-.001	.001	-.453*	15%		
STV	-1.042	.685	-.321	7%		

*p<.05, STV=stance time variability

Table 14: Regression of Energy Cost for Rollator Walking Condition

Model 1 (n=32)	B	SEB	β	sr^2	R^2	ΔR^2
Step 1					.011	-
Gait Speed	.000	.001	-.105	1%		
Step 2					.058	.047
Gait Speed	-.001	.001	-.354	5%		
STV	-1.002	.830	-.331	5%		

STV=stance time variability

Table 15: Regression of Energy Cost for Treadmill Walking Condition

Model 1 (n=31)	B	SEB	β	sr^2	R^2	ΔR^2
Step 1					.402***	-
Gait Speed	-.004	.001	-.634***	40%		
Step 2					.412**	.010
Gait Speed	-.004	.001	-.577**	26%		
GES	-.001	.001	-.117	1%		
Step 3					.437**	.025
Gait Speed	-.004	.001	-.648**	28%		
GES	-.001	.001	-.156	2%		
STV	-1.637	1.496	-.185	2%		
Model 2 (n=31)						
Step 1					.402***	-
Gait Speed	-.004	.001	-.634***	40%		
Step 2					.448***	.046
Gait Speed	-.004	.001	-.535**	24%		
SAFFE fear	.052	.034	.236	5%		
Step 3					.480***	.032
Gait Speed	-.004	.001	-.619***	27%		
SAFFE fear	.060	.034	.275	6%		
STV	-1.841	1.434	-.208	3%		

***P<.001; **p<.01; STV=stance time variability

Table 16: Regression of Energy Cost for Treadmill Slow Walking Condition

Model 1 (n=32)	B	SEB	β	sr^2	R^2	ΔR^2
Step 1					.615***	-
Gait Speed	-.009	.001	-.784***	61%		
Step 2					.712***	.097**
Gait Speed	-.006	.001	-.560***	21%		
Centered GS^2	.000	.000	.384**	10%		
Step 3					.713	.001
Gait Speed	-.006	.001	-.561***	21%		
Centered GS^2	.000	.000	.385**	10%		
GARS-m Hip	-.007	.023	-.029	.08%		
Model 2 (n=32)						
Step 1					.615***	-
Gait Speed	-.009	.001	-.784***	61%		
Step 2					.712***	.097**
Gait Speed	-.006	.001	-.560***	21%		
Centered GS^2	.000	.000	.384**	10%		
Step 3					.717***	.005
Gait Speed	-.006	.001	-.585***	21%		
Centered GS^2	.000	.000	.390**	10%		
SAFFE fear	-.021	.030	-.077	.5%		
Model 3 (n=32)						
Step 1					.615***	-
Gait Speed	-.009	.001	-.784***	61%		
Step 2					.712***	.097**
Gait Speed	-.006	.001	-.560***	21%		
Centered GS^2	.000	.000	.384***	10%		
Step 3					.727***	.014
Gait Speed	-.007	.001	-.616***	22%		
Centered GS^2	.000	.000	.362**	8%		
GES	.001	.001	.129	1%		
Model 4 (n=22)						
Step 1					.733***	
Gait Speed	-.006	.001	-.856***	73%		
Step 2					.762***	.030
Gait Speed	-.005	.001	-.770***	48%		
Centered GS^2	.000	.000	.193	3%		
Step 3					.782***	.019
Gait Speed	-.007	.002	-1.010***	24%		
Centered GS^2	.000	.000	.263	5%		
STV	-1.586	1.254	-.311	2%		

***p<.001; *p<.05; **p<.01; STV=stance time variability; Centered GS^2= quadratic term for gait speed

4.3.2 Changes in Stance Time Variability across Walking Conditions and Subsequent Influence on Energy Cost of Walking

The second purpose of our study was to assess whether increases or decreases in STV (via slow walking and treadmill walking respectively) resulted in subsequent and similar changes in energy

cost of walking. We expected that walking with a rollator (compared to overground) would decrease variability and energy cost, while walking on a treadmill would yield an even greater decrease in both measures. Lastly, we expected slow walking on a treadmill to provide the greatest values of stance time variability and thus the highest costs of walking.

Unadjusted differences for each paired condition are summarized in Table 17. Rollator vs Treadmill walking was the only paired condition to differ in stance time variability (.006 s, p =.03); the positive value of the difference reveals that the mean stance time variability for Treadmill walking was less than that for Rollator walking. Differences in energy cost of walking were significant for three of the four paired conditions (OG-TM, ROL-TM, and TM-TMS; p<.001); no difference was noted for OG-ROL (p=.94). Gait speed differed significantly between all four paired conditions (OG-ROL dif. =3.17 m/min, p=.002; OG-TM dif. =14.1 m/min, p=.0001; ROL-TM dif. =10.9 m/min, p=.0001; TM-TMS dif. =10.9 m/min, p=.0001).

Since gait speed is associated with stance time variability and energy cost, adjusted differences (controlling for gait speed) were also reported (Table 18). Independent of gait speed, stance time variability was significantly lower for the Treadmill condition compared to walking Overground (.013 sec, p<.0001) and with a Rollator (.012 sec., p<.0001). Changes in energy cost of walking for the same two paired conditions failed to differ (OG-TM dif. = -.008, p=.63; ROL-TM dif. = -.02, p=.23), nor did the remaining paired conditions (Table 18). Since the paired walking conditions failed to show adjusted differences for *both* stance time variability and energy cost, we were unable to further assess whether a change in stance time variability accounted for a change in energy cost.

Table 17: Unadjusted Comparison of Variables across Walking Conditions

Paired Walking Conditions	Paired Differences								
	Gait Speed Difference (m/min)			Stance Time Variability Difference (seconds)			Energy Cost of Walking Difference (ml/kg m^{-1})		
	Mean	95% CI	t-value	Mean	95% CI	t-value	Mean	95% CI	t-value
OG vs ROL	3.17	1.2-5.2	3.12**	-.002	-.006-.003	-.82	-.001	-.035-.033	-.08
OG vs TM	14.14	12.1-16.1	13.95***	.004	-.002-.009	1.39	-.065	-.099- -.032	-3.83**
ROL vs TM	10.97	8.9-13.0	10.59***	.006	.001-.011	2.18*	-.064	-.097- -.030	-3.78**
TM vs TMS	10.97	8.9-13.0	10.49***	-.004	-.008-.002	-1.26	-.043	-.077- -.009	-2.50*

*<.05;**<.001; ***<.0001; CI = confidence interval; OG = Overground; ROL = Rolator; TM = Treadmill; TMS = Treadmill Slow

Table 18: Comparison of Variables across Walking Conditions - Adjusted for Specific Gait Speed

Paired Walking Conditions	Paired Differences					
	Stance Time Variability Differences (seconds)			Energy Cost of Walking Differences (ml/kg m^{-1})		
	Mean	95% CI	t-value	Mean	95% CI	t-value
OG vs ROL	.0003	-.003-.004	.21	.011	-.018-.039	.75
OG vs TM	.013	.009-.017	6.18***	-.008	-.039-.024	-.48
ROL vs TM	.012	.008-.016	6.19***	-.018	-.048-.012	-1.21
TM vs TMS	.003	-.001-.007	1.43	.0004	-.030-.031	.03

*<.05;**<.001; ***<.0001; CI = confidence interval; OG = Overground; ROL = Rolator; TM = Treadmill; TMS = Treadmill Slow

4.4 DISCUSSION

Our study set out to examine whether a relationship between stance time variability and energy cost of walking existed across various walking conditions; additionally, we wanted to determine whether a condition-induced change in stance time variability (increase or decrease) would result in a subsequent change (increase or decrease) in energy cost of walking. No significant correlation was found between stance time variability and energy cost of walking across any walking conditions (after adjusting for gait speed). Although stance time variability was found to differ for two paired walking conditions, no subsequent differences in energy cost of walking were observed.

Several factors, such as narrowed range of gait characteristics (gait speed, stance time variability, and energy cost), lack of impact of walking conditions on stance time variability, and contributions from variables not accounted for in our study, may have impacted our ability to show an association between stance time variability and energy cost of walking.

4.4.1 Possible Influence of a Narrowed Range in Gait Speed

As discussed previously (Chapter III, Influence of Narrow Gait Speed Range), the decision to include older adults with overground gait speeds between .8-1.0 m/s, as a way to preemptively manage the confounding influence of gait speed on energy cost of walking, subsequently reduced the range in stance time variability and energy cost in our sample. Selecting our sample of older adults based on a narrow gait speed range may have, in fact, reduced our ability to find a significant relationship between stance time variability and energy cost of walking; only four individuals (13%, 4/30) had values above .20 ml/kg m^{-1}, of which only one would be considered

having a cost value significantly different from the age-appropriate norm (.17 ml/kg m^{-1}). Likewise, only 17% (5/30) of study participants had values of stance time variability considered to be abnormal (≥ .037 s, related to future disability)[16]. It would appear that restricting the range of gait speeds to between 0.8 and 1.0 m/s limited the number of our participants with "higher" values of STV and energy cost, which in turn resulted in a homogenous group of older adults with regards to these two primary variables. Having a homogenous sample with a restricted variable range reduces our ability to detect the strength of association between the two variables and in turn decreases the likelihood of observing any existing relationship between the two variables[100] (Figure 12).

There is some evidence to suggest if we had expanded our range of gait speeds, we may have had more individuals who differed significantly from age-related norms in energy cost - as well as from one another. VanSwearingen et al (2009) selected their sample of older adults based on a slow preferred walking speed of 0.6 to 1.0 m/s (0.2 m/s greater range than our range), and consequently, 73% (35/48) of the sample were considered as having high energy cost of walking values – compared to our sample of only 13% (4/30). The consequence of a narrower range for gait speed was seen across all walking conditions in our study.

4.4.2 The Influence of Different Walking Conditions on Stance Time Variability

The purpose behind testing the relationship between stance time variability and energy cost of walking was to show that the relationship was consistent across a variety of walking conditions, even though variability may "vary". Demonstrating a consistency in the relationship, across various modes of walking commonly used in assessing the gait of older adults, would provide

further support that a robust relationship exists between the two measures. Additionally, as energy cost has been shown to vary over different gait speeds in a curvilinear fashion, it is also possible that energy cost may vary in the same manner with different values of stance time variability. Assessing the response of energy cost to different values of stance time variability would help to further establish the type of relationship that exists (linear or curvelinear).

We selected walking conditions for which we expected to see changes in stance time variability compared to usual overground walking. However, only one of the conditions elicited the change we expected. Our results support the hypothesis we made for the Treadmill condition, stance time variability will be reduced during treadmill walking compared to overground walking. After controlling for gait speed, a significant difference (reduction) in stance time variability was shown for both OG-TM and ROL-TM paired conditions. These differences were not only statistically significant but also consistent with a substantial clinically meaningful differences (.01 s), as related to increased risk for mobility disability.[16] While differences held true for the Treadmill condition, no other paired differences in stance time were significant. Although we expected to see a slight decrease in variability during Rollator walking compared to overground, older adults walked similarly for the two conditions. This is not surprising given the baseline characteristic of our sample; they had a mean gait speed at the low end of normal, normal values of energy cost, and low values of stance time variability. In short, we had a sample of rather "efficient" older adults. Perhaps if our older adults had gait speeds towards the low end of our gait range (0.8 m/s), then stance time variability and energy cost measures would have reflected a more "inefficient" sample of older adults.

It may be that benefits derived from using a Rollator (in terms of reducing variability) only occur at higher values of stance time variability; until such values are achieved, no substantial benefits are received.

Reducing gait speed has been shown to increase measures of variability in older adults.[17, 62, 108] In order to ensure a consistency in "slowness", we selected slow walking on the treadmill as a way to increase stance time variability while ensuring that a slow speed was maintained. Preliminary work in our lab suggested that a 0.4 mph reduction in gait speed was adequate to elicit a substantial change (.01 s) in stance time variability. However, in our study the difference between Treadmill and Treadmill Slow variability was only -.004 (p=.21). Although the direction was consistent with our hypothesis, it was insignificant and below the value reported for a small change (.005).[56]

The inability to induce a change in stance time variability across OG-ROL and TM-TMS conditions contributes to the inability to observe a relationship between stance time variability and energy cost of walking across these conditions.

For the two conditions that did show a significant difference in variability, no subsequent difference was noted in energy cost. The selection of the treadmill as a mode of walking may have presented its own unique problem. Some researcher have argued that treadmill walking, matched on overground speed, is associated with greater energy cost, compared to overground walking.[35] It is possible that the expected "benefit" of treadmill walking (hypothesized as a reduction in stance time variability) on reducing energy cost, may have been overshadowed by the increase in cost incurred from using the treadmill. Parvataneni (2008) noted a .038 ml/kg m^{-1} greater energy cost of walking while on the treadmill compared to overground walking. Although less than a moderate meaningful difference (.5 ES x (sd) of .10) of .05 ml/kg m^{-1}, it

may have been great enough to overshadow the reduction expected from the change in stance time variability. It is also possible that participants were anxious while walking on the treadmill and energy cost values were raised from the start. Although every attempt was made to ensure that our older adults were comfortable and competent with walking on the treadmill, walking on the treadmill may have still caused a greater increase in energy demand (see later section for discussion on anxiety and novelty of task). Selecting all overground conditions for eliciting a change (decrease/increase) in stance time variability may have reduced the influence of using the treadmill, however, maintaining consistent slow and fast walking in older populations may present other study difficulties.

4.4.3 Implications of Other Contributors to Energy Cost of Walking

As previously discussed (Chapter III, Contributions of Other Factors Towards Energy Cost), our inability to find an association between stance time variability and energy cost of walking may be attributed to the multifaceted nature of energy cost of walking, of which, stance time variability is not a primary contributor. Other studies have implicated increased muscle coactivation, altered gait biomechanics, changes in spatial and temporal gait characteristics, and decline in cellular health as possible factors related to the age-associated increase in energy cost of walking. [3, 10, 11, 40] It is possible the more subtle energy demands placed on the muscles of gait related to stance time variability are overshadowed by the greater energy demands requested from other factors listed above, or stance time variability may be a moderator of some of the factors above.

Increased muscle coactivation has been reported to be higher in older adults with poorer postural control, in standing and during walking, compared to older adults with "high" balance ability.[68] Furthermore, increased muscular coactivation has been associated with fear of falling in older adults; individuals who are fearful of falling (compared to their non-fearful counterparts) have higher muscular coactivation (co-contraction index = 46.7 and 59.5, respectively).[67] The increased coactivation associated with fear of falling and poorer postural control may explain, in part, the associations we found between SAFFE fear and energy cost of walking – and possibly between GES and energy cost, although confidence in walking and fear of falling have been considered different constructs. Our correlations between SAFFE fear and energy cost ($r=.394$, $p=.04$ and $r=.622$, $p=.002$ for TM and TMS, respectively) suggest that older adults with higher report of fear of falling during activity have a higher energy cost of walking. It may be that our older adults with greater fear of falling subsequently have higher coactivation of antagonists – thus, higher cost of walking. In this case, SAFFE fear may be a moderator of energy cost through its relationship with muscle coactivation. It is also possible, however, that fear of falling is a moderator of gait speed, which in turn is related to energy cost of walking. Higher fear is associated with slower gait speeds in our study, and slower gait speeds are associated with higher energy cost of walking. Since we did not perform any EMG measures of muscle activity, we are unable to assess which variable might play a greater role in cost of walking in our sample.

As more studies begin to find associations with energy cost of walking, the feasibility of combining such multifaceted variables, in order to assess more comprehensive contributions, becomes more of a reality. A broader understanding of energy cost takes us closer to developing more effective therapeutic interventions aimed at improving the efficiency of walking in older adults.

4.4.4 Are Older Adults Consistent in Measures of Stance Time Variability?

Little is known about the "variability" of variability; are individuals consistently variable or will their variability change in magnitude from session to session? It would benefit us to know whether our participants were stable (consistent) or unstable (inconsistent) in their level of stance time variability. Participants with unstable measures of stance time variability may have had low values during our data collection, but had higher values the next day, or even during another trial. We discuss, in greater detail, the potential implications of an inconsistent measure of variability in an earlier chapter (Chapter III, Is Stance Time Variability Variable?)

It is possible that some older adults in our study are in transition, moving from normal to more abnormal variability but not yet considered consistently abnormal – thus, a group "in flux". As such, stance time variability values may *fluctuate* back and forth from "poor" to "good" within walking trials and between measurement sessions. Older adults who are inconsistently variable may not demonstrate similar relationships with other variables (ie. stance time variability and energy cost) as those who belong to a more defined group (normal or abnormal). Future studies are needed to assess the variability of stance time variability and establish potential boundaries/cut-points that may identify individuals who are likely to be consistent and inconsistent.

4.4.5 Study Strengths

Our study findings are consistent with previous works which reported a strong relationship between gait speed and energy cost of walking (Figure 14).[3, 4, 29] Ralston (1960) was one of the first to report that as an individual's gait speed deviates away from "normal", the cost of walking

becomes greater – visually depicted as a J-curve.[29] Thus, speeds that were slower than 1.1 m/s and faster than 1.4 m/s, had *greater* subsequent values of energy cost.[29] Martin (1992) and Malatesta (2003) repeated this earlier work, and have increased the breadth of understanding gait speed contribution to energy cost by showing that the relationship exists across various ages. Although general shape of the J-curve remained the same, the vertical placement of the curve varied for different ages; the curve for older adults was positioned above all other age-groups, showing a higher cost of walking compared to young adults- even at similar walking speeds.

When working with older adults with only slow gait, we would expect to see the left end of the speed-energy cost curve, a negative association, as the energy cost-speed relationship "plateaus" across the range of normal gait speeds (approximately .9-1.4 m/s)[3, 4, 29]. Our current findings further support the negative relationship between gait speed and energy cost, as our correlations for individual walking conditions ranged from r= -.105 (p=.57) for the Rollator condition to r= -.856 (p=.001) for Treadmill Slow. As a note: we did include a quadratic term for gait speed to account for the curvilinear relationship between speed and energy cost in our correlations; however, the quadratic terms were not significant with respect to cost in any of the conditions.

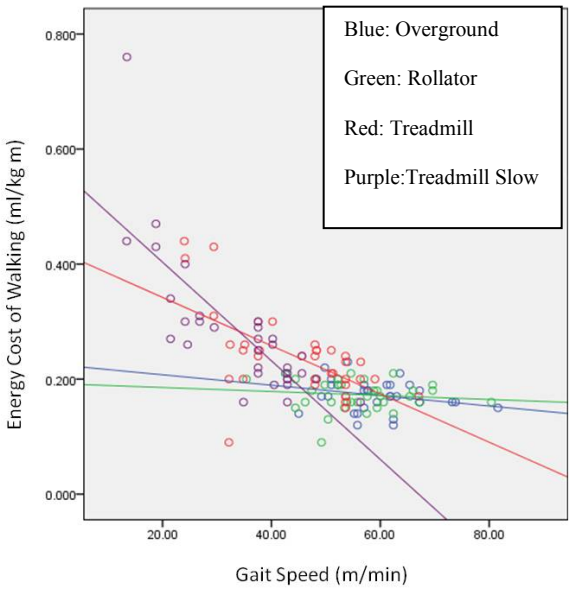


Figure 14: Speed-cost relationship across walking

Our study demonstrated the feasibility of recording concurrent *overground* measures of stance time variability and oxygen consumption in older adults. Collecting overground measures requires participants to wear portable equipment in order for data to be collected. All of the older adults in our sample were able successfully complete their 4-6 minute walking session without complaints pertaining to wearing the portable devices. Furthermore, we were able to demonstrate that gait speeds throughout the course of overground walking can be consistent among older adults, even those with slow gait speeds, as with our sample. We calculated the energy cost of walking using different measures of gait speed for each participant and found no significant difference in cost based on the different speeds. As a method of keeping gait speed consistent for studies collecting measures of gait and oxygen consumption, treadmills are often selected as the mode of activity. However, walking on the treadmill is not a common activity for many older adults and may not be as accurate of a measure of energy cost of walking as overground walking. Our study suggests that older adults with low variability and slow gait

speed (between .8 and 1.0 m/s) *can* maintain a consistent preferred walking speed overground, offering a more natural mode of investigating gait and energy cost of walking for future studies.

Finally, our studied showed that walking on the treadmill decreases stance time variability in older adults with slow gait. As stance time variability is associated with future mobility disability in some older adults[16], treadmill training may be a beneficial addition to therapy interventions aimed at improving mobility in older adults. Although our lab has completed preliminary work demonstrating feasibility for carry-over (unpublished), further studies (with larger sample sizes) are required to assess whether the reductions in stance time variability induced by the treadmill can be transferred to overground walking.

4.4.6 Study Limitations

As previously mentioned our study sample of older adults is restricted in their range of gait speeds and therefore are not representative of all older adults. Thus our study findings should be generalized only to like populations.

Four different walking conditions were selected to be used in this study; each condition was hypothesized to be unique in its "level" of stance time variability (see Measures). However, two conditions were performed overground, while two others were performed on a treadmill. It is possible that walking on a treadmill may have influenced energy cost of walking in different ways (anxiety, greater muscle co-activation) compared to overground walking. Such increases may have elevated values of energy cost to a greater extent than the benefit (reduce energy cost) of treadmill walking could reduce. Despite providing practice time on the treadmill, in order to establish familiarity with the task and reduce novelty and anxiety, participants may still have had anxiety and used compensatory mechanisms not accounted for by the observational assessment

of biomechanics (increased toe push-off, increased antagonist coactivation, increased grip on railings). Although we had measures of fear and confidence in our study, these measures were not condition specific and therefore would not be able to account for any "anxiety" that may have been present for the treadmill condition. It may benefit future studies to consider using different walking conditions within a single mode of walking (all overground or all treadmill).

We also recognize that gait speeds varied significantly between walking conditions. Participants were allowed to select their preferred walking speed for each condition. Despite efforts to keep gait speeds consistent across the three usual-paced conditions (ie practice trials across various speeds, ensuring proper calibration of the treadmill), participants consistently selected slower gait speeds on the Treadmill compared to Overground. More subtle, but significant, reductions were observed between overground and Rollator walking as well. Matching treadmill gait speed to overground gait speed may introduce a novel effect, such that participants feel they are walking faster than their normal – and may result in greater energy cost. When working with older adults, especially those not accustomed to walking on a treadmill, researchers may need to provide longer and more frequent practice trials prior to data collection. This may allow participants to establish more similar speeds between varying walking conditions. Additional attention is also needed in the area of statistical analysis and controlling for gait speed. As gait speed shows strong relationships between many concurrently studied research variables (stance time variability, energy cost of walking, biomechanics), important consideration should be given when controlling for gait speed in the analyses – so as not to "wash out" the underlying effects/relationship of variables. Since energy cost is a measure of oxygen consumption "standardized" by gait speed, additional control of gait speed in analyses

may “over-control” for speed and subsequently reduced any small, but significant relationship it may have with other variables.

4.4.7 Conclusion

For reasons not completely understood, older adults expend more energy during walking than young. For adults with lower aerobic capacity, such increased demands on an already taxed system may significantly reduce their likelihood of maintaining functional independence. Greater cost of walking may also discourage older adults from participating in physical activity due to an increased feeling of exertion or fatigue - further increasing their risk for mobility disability. In addition to greater cost of walking, older adults also show greater levels of stance time variability, which is also linked to a greater risk of developing mobility disability. Although these two age-related characteristics share a common threat to future mobility, so far there remains no evidence to suggest that the two are related. Future research should continue to focus on assessing multiple “potential” contributors to energy cost of walking, across larger samples of older adults, in order to establish a more comprehensive look into identifying factors attributing to the age-related increase in cost of walking.

5.0 CHAPTER V: CLINICAL SIGNIFICANCE AND DIRECTION OF FUTURE RESEARCH

Walking plays an integral role in undertaking activities of daily living and in the ability to complete instrumental activities of daily living among older adults.[2] Interestingly, older adults have a greater energy cost associated with walking than do young adults[4, 6], which is of concern as higher cost of walking has been related to poorer reports of function among older adults[7, 8]. Factors responsible for the greater energy cost of walking observed in older adults have not been clearly identified.

In an attempt to provide greater insight into the cause of costly gait among older adults, which in turn could provide greater focus to rehabilitation programs aimed at improving the mobility of older adults, we explored the relationship between stance time variability (known to be associated with mobility disability) and energy cost of walking in a cohort of older adults with slow gait. Specifically, this project explored the potential influence of wearing a portable gas-analysis system, used to record oxygen consumption, on the gait characteristics of older adults. As studies begin to use portable devices to assess oxygen consumption during more natural conditions (overground walking versus treadmill walking), it is important that we fully understand the impact that wearing such devices may have on other measures of interest to researchers (such as gait characteristics). Additionally, we explored the relationship between stance time variability and energy cost of overground walking. To our knowledge, this is one of the first studies to collect *concurrent* measures of oxygen consumption and gait variability during *overground* walking. Finally, in an effort to comprehensively explore the stance time variability – energy cost of walking relationship, we assessed the relationship over a variety of

walking conditions thought to influence (increase/decrease) stance time variability. Demonstrating that reductions in stance time variability result in a subsequent reductions in energy cost of walking, would provide further support as to the robust nature of the relationship, and provide evidence that substantiates the use of stance time variability as a rehabilitation target for interventions aimed at improving efficiency of gait in older adults.

By exploring the potential influence of wearing a portable gas-analysis system on the gait characteristics of older adults, we found no evidence to suggest that wearing such a device results in any significant change to temporal or spatial gait characteristics in our sample of older adults. For researchers devoted to examining the relationship between gait characteristics and energy cost of walking in older adults, our findings provide confidence that gait data is not adversely altered while trying to collect concurrent measures of oxygen consumption via a portable system. Since our findings only report on the impact of one brand of portable gas analysis systems (Medgraphics, VO2000), and on a very specific population of older adults, we recommend further research in other preferably larger samples of older adults – and across different portable devices, to substantiate our findings. Such work would add to the strength supporting the use of this flexible and less restrictive method of recording measures of oxygen consumption.

Our investigation of the relationship between stance time variability and energy cost of overground walking indicates that stance time variability is not a significant contributor to the energy cost of walking in our sample of older adults. We suggested that one explanation for our inability to identify a relationship between the two measures was an inadvertent narrowing of the range in stance time variability and energy cost of walking – limiting the number of participants who were variable and inefficient walkers. We also recognized that energy cost of walking is

likely influenced by multiple factors, of which stance time variability may not be a primary contributor. Energy demands exacted from muscles due to stance time variability may be more subtle, and overshadowed by the larger energy demand placed on muscles due to other factors acting on the muscles (coactivation).

Despite not being able to demonstrate a relationship between stance time variability and energy cost of walking, we were able to demonstrate the feasibility of collecting concurrent measures of gait characteristics and oxygen consumption via portable devices, in older adults. Older adults may be more susceptible to alterations in gait, increased unsteadiness, and elevated anxiety associated with wearing portable data collection equipment while walking, possibly moreso if equipment is worn on the face. Our older adults were able to successfully complete all walking conditions requested of them, without complaint of increased anxiety or fear of falling related to wearing the equipment; 160/160 walking trials were completed during the study. Furthermore, as RER values were monitored as part of oxygen consumption data collection, all values remained below 1.0-1.1, demonstrating that older adults were below the aerobic threshold and were not physically taxed during the condition (often reported as values above 1.1).

In addition to demonstrating the feasibility of portable data collection involving older adults, we were also able to demonstrate the feasibility of collecting overground measures of gait and oxygen consumption. Typically, the treadmill has been used as the primary mode of data collection as gait speed can be precisely controlled during longer periods of walking, compared to overground walking. However, researchers continue to debate the differences and similarities between treadmill walking and overground walking. We were able to collect two 4-meter gait speeds for approximately 75% of the laps completed by each participant during the overground conditions. We were able to show that the mean gait speed did not vary from speeds recorded

towards the end of each walking trial, and that measures of energy cost of walking failed to differ when calculated with the different speeds. Furthermore, our reliability for Overground was ICC=.8 and for Rollator walking was ICC=.87. Our findings offer evidence, for researchers interested in investigating gait and energy cost measures in older adults during more natural (real life) activities, that older adults with slow gait can maintain consistent preferred walking speeds overground. Since some researchers have suggested that treadmill walking can increase energy cost of walking at speeds similar to overground speeds, and that gait characteristics can also vary, studies which use overground walking as the mode of activity can circumvent these "short-comings" of using the treadmill, and confidently collect data during overground trials.

Historically, gait characteristic data has been collected using force plates, motion-capture analyses on treadmills, or overground using computerized walkways of limited length (ie. 4-8 meters). Although consistent in their measures, such methods don't allow for the collection of many steps over a continuous period of walking in a natural setting (overground). Portable footswitch systems have provided a method for researchers to record hundreds of footsteps during overground walking. Our study reported the reliability of a portable system to collect overground gait variability measures in older adults; our reliability (ICC=.51) was similar to the gold standard measurement system (computerized walkway), which was ICC=.50.

We hypothesized that certain walking conditions would vary in their measures of stance time variability by acting on either (or both) central or peripheral factors associated with automatic stepping. Our study was able to provide evidence that stance time variability is reduced while walking on the treadmill, compared to overground walking. Since stance time variability is known to be associated with future mobility disability, reducing stance time variability in older adults with higher than normal values may help reduce future risk of

disability - treadmill training may serve as an appropriate intervention to reduce stance time variability. However, future research is needed to confirm the carryover of such reductions to overground walking. Unpublished work in our lab, with a small sample of older adults, suggests that treadmill training can reduce overground values of stance time variability below values associated with future disability, and for some older adults – in as few as 7 training sessions. Future studies should focus on assessing such change in larger samples of older adults, with subsequent follow up visits to evaluate the length of time such benefits are maintained.

Although our study was unable to provide evidence supporting a relationship between stance time variability and energy cost of walking, the fact remains that stance time variability is associated with mobility disability, specifically a decline in walking ability. Thus, stance time variability is impacting some component of gait – but how and where remains unclear. Possibly, stance time variability is a precursor to other factors, like coactivation, and only makes an impact on such factors once certain "thresholds" are crossed. Likewise, contributions to energy cost of walking may come from many factors, alluded to in more recent research, and as such, complex relationships may exist between such factors. However, we are not aware of any studies that have taken a multifactorial approach to explaining the higher cost of walking in older adults, nor have many begun to explore mediating/moderating effects of factors on one another, as they relate to energy cost of walking. Although researchers have identified muscle as the greatest consumer of oxygen during walking [14], and research has focused on identifying factors that could account for greater muscle demand in older adults; more basic science research may be warranted to identify other mechanisms that might contribute to greater oxygen in older adults. Age-related changes in oxygen delivery to body tissues, changes to blood and cellular

composition, and altered efficiency of cellular production of energy (ATP) may provide additional insights into higher energy cost of walking in older adults.

Despite the increasing amount of research devoted to identifying contributors to the higher energy cost of walking in older adults, few intervention studies have been presented, demonstrating the ability to improve the efficiency of gait in older adults. One group of researchers have shown that energy cost of walking was decreased in a sample of older adults who received 12 weeks of physical-therapist supervised exercise, with specific emphasis on principles of motor learning that enhance "skill" or smooth and automatic movement control.[41] More intervention studies are needed to provide insight into mechanisms that may influence (restore) more normal central and peripheral signaling to spinal stepping mechanisms, or improve the efficiency with which oxygen is transferred or utilized by the body tissues.

Walking is a vital function for the performance of many activities of daily living and other routine tasks of independent living among older adults. Aging appears to exact a greater demand from the body during walking, possibly exposing some older adults to greater risk of developing future disability and loss of independent function. Research must continue to focus on identifying factors which contribute to the higher cost of walking observed in older adults, while additional efforts are put forth to develop and test intervention-related hypotheses aimed at re-establishing more efficient walking; such efforts will bring us closer to enhancing the mobility of many older adults – one step at a time.

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